The Effect of Increasing Inertia upon Vertical Ground Reaction Forces during Locomotion

John K. De Witt, Ph.D.[1], R. Donald Hagan, Ph.D.[2], Ronita L. Cromwell, Ph.D.[3]

[1]Bergailia Engineering Services, Houston, Tx, USA; [2]NASA – Johnson Space Center, Houston, Tx, USA; [3]Department of Physical Therapy, The University of Texas Medical Branch, Galveston, TX, USA

Corresponding Author:

John K. De Witt, Ph.D.

Wyle Labs

1290 Hercules, Ste 120

Houston, Tx 77058

Phone: (281) 483-8939

Fax: (281) 483-4181

Email: john.k.dewitt@nasa.gov

Address reprint requests to the corresponding author

Keywords: biomechanics, locomotion, ground reaction forces, inertia, gravity, weight, mass, human

Short Title: Adaptations to inertia during locomotion

Summary

The addition of inertia to exercising astronauts could increase ground reaction forces and potentially provide a greater health benefit. However, conflicting results have been reported regarding the adaptations to additional mass (inertia) without additional net weight (gravitational force) during locomotion. We examined the effect of increasing inertia while maintaining net gravitational force on vertical ground reaction forces and kinematics during walking and running. Vertical ground reaction force was measured for ten healthy adults (5 male/5 female) during walking (1.34 $m \cdot s^{-1}$) and running (3.13 $m \cdot s^{-1}$) using a force-measuring treadmill. Subjects completed locomotion at normal weight and mass, and at 10, 20, 30, and 40% of added inertial force. The added gravitational force was relieved with overhead suspension, so that the net force between the subject and treadmill at rest remained equal to 100% body weight. Peak vertical impact forces and loading rates increased with increased inertia during walking, and decreased during running. As inertia increased, peak vertical propulsive forces decreased during walking and did not change during running. Stride time increased during walking and running, and contact time increased during running. Vertical ground reaction force production and adaptations in gait kinematics were different between walking and running. The increased inertial forces were utilized independently from gravitational forces by the motor control system when determining coordination strategies.

Introduction

During locomotion, the forces occurring between the foot and ground reflect the acceleration patterns of the body's center of mass (Munro et al., 1987). The magnitudes of these ground reaction forces (GRFs) therefore are influenced by gravity and the inertial properties of the body. While walking and running in microgravity or a reduced-gravity environment, the amount of force necessary to overcome gravitational forces will vary. However, since the mass of the body will not change, the force necessary to overcome inertia of the body should remain unaffected.

During long-term space flight, astronauts experience losses in bone mineral density in addition to the loss of muscle (Schneider et al., 1995; LeBlanc et al., 2000; Iwamoto et al., 2005). Locomotive exercise is performed during spaceflight by astronauts, resulting in GRFs that may create bone strains hypothesized to be an osteogenic stimulus (Rubin and Lanyon, 1985). An external load is applied through a waist and shoulder harness that anchors the astronaut to the treadmill. McCrory et al. (2002) suggest that achieving an external load which is equivalent to 100% of body weight is beneficial to generating GRFs similar to those experienced in normal gravity. However, in actual microgravity conditions, increasing external loads may not be sufficient to recreate all components of GRF trajectories experienced in normal gravity (Schaffner et al., 2005). Inertia is dependent upon mass and independent of gravity level. Therefore, the addition of inertial mass to subjects exercising in microgravity may enhance GRF magnitudes and prove beneficial for bone health.

The addition of inertial mass could cause adaptations in locomotion kinematics and kinetics that may mitigate increased GRFs under some conditions. In general, increasing inertial and gravitational force has been shown to increase metabolic cost and joint forces during walking (Griffin et al., 2003). However, investigations examining the effect of increasing inertia while maintaining gravitational force during locomotion in normal gravity (Grabowski et al., 2005; Chang et al., 2000) report conflicting findings. Grabowski et al. (2005) found that increasing inertia without increasing gravitational force during treadmill walking at 1.25 $m{\cdot}s^{-1}$ resulted in an increased oxygen consumption, suggesting that the expenditure of metabolic energy used to overcome inertial forces is independent of that used to overcome gravitational forces. Chang et al. (2000) examined GRF, temporal variables, and the orientation of the GRF vector during treadmill running at 3.0 $m{\cdot}s^{-1}$ at various gravitational and inertial conditions. They reported that an increase in inertia without a corresponding increase in gravitational force did not result in an increase in GRF or changes in locomotion kinematics.

However, the problem with these findings is how does one explain the contradiction between increase in metabolic rate and lack of change in GRF? An increase in mass resulted in an increase in metabolic cost during walking, but kinematics and GRF were not affected during running. Adaptations in muscular activity may have occurred in response to the increase in mass to maintain optimal locomotion kinematics, resulting in greater energy expenditure. It is also possible that there could be fundamental differences between these two types of locomotion that result in differing adaptation to inertial modifications.

The purpose of this investigation was to examine the affect of added inertial force (AIF) while maintaining gravitational force on GRFs during walking and running. Specifically, we hypothesized that there will be kinematic and GRF adaptations to increasing inertial mass while maintaining gravitational force and that the adaptations will differ between the two modes of locomotion. The results of this investigation will help to better explain the affects of inertial forces on locomotion independent of gravitational forces.

Materials and Methods

SUBJECTS

Ten experienced treadmill runners (five men and five women) volunteered to participate in this study (age 34.4 ± 6.9 years, mass 68.4 ± 11.7 kg; mean ± S.D.). All subjects were healthy and had previously passed a yearly United States Air Force Class III-equivalent physical examination. In addition, because the vest used for adding mass to the subject had a maximum capacity of 38.1 kg, all subjects had to weigh less than 95.3 kg. This investigation was reviewed and approved by the NASA Johnson Space Center Committee for Protection of Human Subjects. Subjects provided written informed consent prior to participation in the study.

INSTRUMENTATION

Vertical GRF data were collected using a force-measuring treadmill (Kistler Gaitway, Amherst, NY). GRF data were sampled at a rate of 481.2 Hz with two force plates beneath the running tread. Force plates were arranged so that one plate rested in front of the other. Each plate contained four piezoelectric load cells that measured vertical GRF and allowed for determination of the center of pressure.

Inertial force was added to each subject using a weighted exercise vest (X-Vest, Perform Better, Cranston, RI). The vest had pockets located around the upper and lower trunk for the addition of weights (Figure 1). Each pocket was fitted with slots in which up to twenty-one individual 0.45 kg masses could be placed. Slots for weights were located on both inner sides of each pocket (10 on the outer side; 11 on the inner side). During trials, masses were added equally to the front and rear of the vest. The masses were always added to the inner-lower slots first, followed by inner-upper, outer-lower, and outer-upper slots.

Gravitational force was maintained with an overhead unweighting system (H/P/Cosmos Airwalk, Nussdorf, Germany). The system provided a constant upward force via a pneumatic pump that unweighted subjects through use of a harness worn about the waist and thighs (Figure 1).

Place Figure 1 about here

PROCEDURES

Prior to data collection, each subject participated in a familiarization session during which they had the opportunity to practice each test condition at each speed until they were comfortable. Subjects then completed one walking and one running data collection session within one week after the familiarization session. Walking trials occurred during a separate session than running trials. Seven days separated each session. The speed order for the data collection trials was randomized for each subject using a coin flip during the familiarization session.

Treatment randomization occurred independently for each speed. To assure that there was a balance of AIF conditions between subjects, a balanced Latin square random assignment was used (Portney and Watkins, 2000). The design allowed for a balance of treatment orders so that no two testing sequences were the same for different subjects within each speed. Each subject was randomly assigned a sequence from the table with only one subject completing each specific order. Trial order assignment occurred separately for each speed. The subjects wore the unweighting harness during all conditions, including the 0% AIF trial.

Upon arrival to the laboratory, each subject was provided with standardized running shoes (Xccelerator TR, NIKE, Inc, Beaverton, OR) and completed a general health questionnaire. Once the unweighting harness had been donned, the subject's weight was measured by the force treadmill. This weight was used to compute the amount of AIF required to achieve each condition.

Data were collected at two speeds during five AIF treatments. Subjects walked at 1.34 $m \cdot s^{-1}$ and ran at 3.13 $m \cdot s^{-1}$. In addition to a control condition of no added inertial force (0% AIF), inertial force was added while body weight was maintained. We added an additional 10%, 20%, 30%, and 40% of body weight and mass to each subject. For each AIF condition, the added weight was relieved with the unloading system so that the net force between the subject and treadmill remained equal to 100% body weight.

Subjects completed approximately one minute of treadmill locomotion at each AIF condition. Data collection began once the subjects had achieved a steady walking or running pace. Immediately following one minute of data collection, the weighted vest was removed and the unweighting harness was released. The subject then completed three minutes of walking at 1.34 $m\ s^{-1}$ to eliminate any adaptation to gait that may have occurred during the test condition. The subjects were given additional rest of approximately three to four minutes until they felt that they were ready to continue with the next AIF condition.

DATA PROCESSING

The first ten strides of the left leg were analyzed in each one-minute trial. The left side only was analyzed with the assumption that gait kinematics were symmetrical within subjects (Karamanidis et al., 2003). The chosen epoch began with the first heel strike of the left foot, and ended with the eleventh heel strike of the left foot. Data analyses were performed using software written in Visual Basic for Applications interfaced with Microsoft Excel 2003 SP1 (Redmond, WA) and MATLAB Version 7.2.0.232 (R2006a) (Mathworks, Natick, MA).

Custom software converted the output from each force sensor to vertical GRF and center of pressure location. Raw voltage data from the eight load sensors in the treadmill force platforms were transformed into forces using calibration factors. The total vertical GRF during each sample was then calculated as the sum of the vertical forces measured by each sensor. Center of pressure during each sample was determined relative to the force platform reference frame using the force outputs from each sensor along with the dimensions of the force sensors relative to one another. Center of pressure locations were used to determine which foot was in contact with the treadmill during each step.

DATA ANALYSES

The time that heel strike occurred for each stride was determined using GRF data according to the criterion of Chang et al. (2000). An automated algorithm determined heel strikes as the samples at which a positive change in the force greater than 1 N s^{-1} occurred when the force magnitude was less than 100 N. The time of toe off was computed in a similar manner. Toe off samples were defined as the samples at which a negative change in the GRF less than 1 N s^{-1} occurred when the magnitude of the force was less than 100 N.

GRF data were used to find contact time, stride time, peak vertical impact force, loading rate, peak vertical propulsive force and impulse for each trial. All analyses were completed using raw GRFs to ensure that peak values were not dampened. Visual inspection of each footfall was used to ensure that there were no anomalous data.

Contact time was the length of time that the left foot was in contact with the treadmill during each stride, and was calculated as the duration between heel strike and toe off for each left footfall. Stride time was the length of time between successive heel strikes of the left foot. Peak vertical impact force was the magnitude of the first distinct peak in the GRF trajectory. Peak vertical propulsive force was the magnitude of the second distinct peak. Loading rate was the peak vertical impact force divided by the time between heel strike and time of peak vertical impact force. The impulse for each footfall was computed as the integral of the GRF trajectory over contact time. Peak vertical impact force, loading rate, peak vertical propulsive force and impulse were all normalized to actual body weight found prior to the data collection session to allow for inter-subject comparisons.

STATISTICAL ANALYSIS

There were six dependent variables computed for each stride during each trial. A trial mean for each variable was calculated from all ten strides. Statistical analyses were conducted utilizing NCSS 2004 statistical software (Kaysville, UT). Trial means for all dependent variables were tested using repeated measures analysis of variance (ANOVA) with AIF level as a single factor. Walking and running were analyzed separately because they are two different tasks that require different motor patterns. Tukey-Kramer Multiple Comparisons tests were used to determine differences between AIF levels when a significant main effect was found. Differences were considered to be statistically significant when $p<0.05$.

Results

TEMPORAL PARAMETERS

Contact time did not change during walking, but did increase with added inertia during running. However, stride time was affected by inertia during both walking and running (Table 1). While there was no effect of AIF condition on contact time during walking, running produced increased contact time as inertial force was added to the subject ($p<.001$). The 10%, 20% 30% and 40% AIF conditions had greater contact times than the 0% AIF condition. The 30% and 40% AIF conditions had greater times than the 10% AIF condition, and the 40% AIF condition had a greater time than the 20% AIF condition.

There was a main effect of AIF condition during both walking ($p=.047$) and running ($p<.001$). During walking, stride time was less during the 20% AIF condition than the 40% AIF condition. During running, stride times during the 40% AIF condition were longer than the 0% AIF, 10% AIF and 20% AIF conditions.

Place Table 1 about here

GROUND REACTION FORCES

During walking, peak vertical impact forces increased and peak vertical propulsive forces decreased with the addition of inertial force. During running, peak vertical impact forces decreased as inertial force was added. In general, the trajectories of the GRF curves appeared to be a similar shape to one another regardless of the AIF level. Normalized, ensemble averaged vertical GRF trajectories for walking and running over all trials for all subjects are shown in Figures 2 and 3.

Peak vertical impact force and loading rate were significantly affected by the addition of inertial force (Table 2). There were significant effects of AIF on peak vertical impact force during both walking ($p<.001$) and running ($p<.001$). During walking, the peak vertical impact force increased with the addition of inertial force, with the force at 20%, 30% and 40% AIF greater than during 0% AIF. Peak vertical impact force at 30% AIF was significantly greater than that at 10% AIF.

During running, peak vertical impact forces decreased as AIF increased. Peak vertical impact force during running for the 20%, 30% and 40% AIF conditions was less than that of the 0% AIF condition. In addition, peak vertical impact force at 40% AIF was less than the 10% AIF condition.

Place Table 2 about here

Similar to peak vertical impact force, there were effects on loading rates during both walking ($p<.001$) and running ($p<.001$). During walking, loading rate was significantly greater during the 20%, 30% and 40% AIF conditions than during the 0% AIF condition. Loading rate was also greater at 30% AIF than at 10% AIF. During running, loading rate was significantly greater during the 0% AIF than during 20%, 30% and 40% AIF. Loading rate during the 10% AIF was significantly greater than during 40% AIF.

Peak vertical propulsive forces were affected by additional inertial force only during walking, while impulse was affected by added inertial force only during running (Table 3). There was no effect of AIF effect during running for peak vertical propulsive force. During walking, peak vertical propulsive force decreased as inertial force was added ($p<.001$). The peak vertical propulsive force during 30% and 40% AIF conditions was less than during the 0% AIF condition. In addition, the peak vertical propulsive force during the 40% AIF condition was less than all other AIF conditions. There was no effect of additional inertial force upon impulse during walking. During running, the impulse during 40% AIF condition was greater than during the 0% AIF condition (p=.027).

Place Table 3 about here

Place Figure 2 about here

Place Figure 3 about here

Discussion

We found that adaptations to AIF occur in the kinematics and GRF of walking and running. The addition of inertial force without a corresponding increase in gravitational force resulted in increased peak vertical impact forces and loading rates during walking, but decreased peak vertical impact forces and loading rates during running. In contrast, peak vertical propulsive forces during walking decreased with added inertial force but did not change during running.

This study was the first to investigate walking and running with increased inertial forces using identical methodology for each speed. We observed that the adaptations to additional inertial force during walking were different than those during running.

LOCOMOTIVE ADAPTATIONS TO ADDED INERTIAL FORCE DIFFER BETWEEN WALKING AND RUNNING

During walking gait kinematics generally were maintained while impact GRF increased as inertial force was increased. Surprisingly propulsive GRF decreased. The opposite occurred during running. Kinematics were affected by AIF while propulsive GRFs were unaffected.

We define adaptation to occur when the kinetic or kinematic variable of interest remains unaffected by the addition of inertia. However, if a change occurs as a result of the treatment, then no adaptation is evident. Adaptation strategies to AIF during walking may possibly prioritize the preservation of gait kinematics while generating greater GRF. During running, however, kinematic adaptations occur to gait patterns that result in GRF maintenance. Perhaps the GRF magnitude is used by the control system during the production of movement, such that increases in the forces are mitigated to prevent potential injury.

During walking, subjects maintained contact time and stride time, except during the most extreme loading condition (40% AM), during which stride time increased. Kinetic adaptations included increases in peak impact GRF and loading rate coupled with a decrease in propulsive GRF. The increases followed a linear dose-response relationship up to 30% AIF, and may explain the increases in metabolic costs reported by Griffin et al. (2003) and Grabowski et al. (2005). There was a slight decrease in impact forces and loading rates from 30% to 40% AIF, suggesting that a threshold effect may occur at additional inertia greater than 30% of normal. The threshold might occur as a protective mechanism against injury, since increased loading could increase the risk of bone or muscle damage.

The decrease in vertical propulsive GRF when walking suggests that subjects adapted during the latter phases of contact. Chang et al. (2000) defined the impact phase of foot-ground contact to occur during the initial 20% of the contact phase. Increased inertia during the AIF conditions resulted in greater GRF necessary to decelerate the body as the foot contacted the ground. The lack of adaptation to the increased inertia resulted in a dose-response relationship between impact GRF and AIF. However, the decrease in GRF during the propulsive phase of ground contact suggests, that some adaptation had occurred that affected the latter phases of stance. Therefore, adaptations to increased inertia may occur during specific phases of the gait cycle rather than to the entire pattern.

The increase in walking impact force coupled with the decrease in propulsive force could be a result of a decreased need to generate vertically-directed forces to propel the center of mass upward. During walking, the trajectory of the body's center of mass can be approximated as an inverted arc (Alexander, 1976; Lee and Farley, 1998). The center of mass is highest during single-limb support and lowest during double-limb support (Chou, et al., 2001). If the maximum height of the center of mass does not change, but the downward displacements of the endpoints of the arc decrease, the arc becomes flatter. The downward displacement during double-support will be dependent upon the upward force applied and the time the impact force acts. If the time the force acts does not change, but the magnitude of the force increases, the downward displacement of the center of mass will be reduced. This could result in less propulsive force necessary at the end of ground contact to attain the same center of mass height.

It also is possible that as inertial force increased, there was an amplified requirement for control of the upper body to maintain balance and posture. As inertial force was added, the relative mass of the trunk increased with respect to the rest of the body's segments. Subjects may have decreased propulsive GRF in order to increase the control of the trunk.

Because our equipment did not allow us to measure shear forces between the foot and treadbelt, it is impossible to determine if the reduction in vertical GRF was accompanied by an increase in horizontal GRF. If this were to occur, the magnitude of the GRF would be unaffected, but the orientation of the GRF vector would change. This could be another explanation for the reduced vertical GRF with increased inertial forces.

However, Chang et al. (2000) found that during running with increased inertial forces, the orientation of the GRF vector did not change. Since our experimental setup was similar to theirs, we have no reason to believe that a chance in GRF vector occurred.

In contrast to walking, both contact time and stride time during running increased with added inertia. Peak vertical impact forces and loading rates decreased, and peak vertical propulsive forces did not change. Horizontal GRFs may have increased, but we were not able to measure shear force magnitudes with our apparatus. Vertical impulse increased with greater inertia. The impulse of the center of mass is directly related to GRF magnitude and the times the forces act. The adaptation of longer contact time to increased inertia may occur to allow peak vertical impact forces to decrease. The longer contact time allowed the vertical GRF to act for a longer period of time. Although peak vertical impact forces decreased as inertia was added, the relatively small proportion of the entire GRF trajectory due to impact forces may not have been enough to cause a decrease in impulse.

The larger GRF throughout the stance phase during running compared to walking will create greater accelerations of the center of mass. We felt it reasonable to hypothesize that greater GRF would occur to maintain normal locomotive kinematic patterns when mass was added. We found the opposite to occur, where kinematic patterns were altered and vertical impact GRF decreased.

Similar to walking, the lack of increase in propulsive GRF during running may have resulted in decreased vertical acceleration of the center of mass as inertial force increased. A flatter trajectory of the center of mass during the flight phase may have resulted from the lack of increase in vertical propulsive GRF with the accompanying increase in inertial force. Because the center of mass trajectory may have been flatter, there was a decrease in impact GRF due to a decreased need to decelerate the center of mass.

The control strategy utilized during running also may be to adapt kinematics in order to increase control, since increases in GRF would affect the accelerations of the center of mass. Most strikingly, peak impact forces decreased as mass was added. During normal locomotion, vertical GRF, including impact forces, will increase in proportion to gravitational forces (Cham and Redfern, 2004). A protective mechanism may be employed to reduce injury risk. These adaptations were automatic, and did not appear to be consciously chosen by the subject, and may have resulted from increased inertia without increased net gravitational forces.

Perhaps the most interesting finding is that the adaptations to inertia during walking were different than during running. This suggests that walking and running should be thought of as two distinct tasks, rather than alternate forms of locomotion. Our results may also explain the seemingly disparate findings between the research of Grabowski et al. (2005) and Chang et al. (2000).

Grabowski et al. (2005) found that increasing inertia without increasing gravitational force resulted in an increased metabolic cost during treadmill walking at 1.25 m s^{-1}. Metabolic cost, as determined from oxygen consumption, increased as inertial force was added. This finding suggests that there is a cost of metabolic energy used to overcome inertial forces that is independent of that used to overcome gravitational forces. Chang et al. (2000) examined GRF, temporal variables, and the orientation of the GRF vector during treadmill running at 3.0 m s^{-1} at various gravitational and inertial force conditions. They reported that an increase in inertial force without a corresponding increase gravitational force did not result in an increase in GRF or changes in locomotion kinematics.

Modification in metabolic cost can be reflected by increases in muscular work (Griffin et al., 2003). The differences between Grabowski et al. (2005) and Chang et al. (2000) may be explained by the fact that the former tested walking while the latter tested running. Lack of adaptation to increased inertia during walking resulted in an increased peak impact force. The increased impact force may result in greater ankle, knee and hip extensor activity, leading to an increased metabolic cost during walking. During running, however, there is no increase in GRF, resulting in no additional metabolic cost to increased inertia. It must be noted that we were unable to measure horizontal forces, hence there is the possibility that horizontal forces changed as inertial forces were modified, which could potentially affect our results. However, given that we tested all subjects in the same manner, we believe any affects of the horizontal GRF would be systematic. Chang et al. (2000) did find an increase in horizontal impulse when inertial forces were increased and gravitational forces were held constant. However, the increases were not linearly related to the amount of added inertia.

Finally, our results suggest that adaptations to inertia may occur during the impact or propulsive phases as opposed to causing systematic adaptations to the entire gait pattern. In other words, adaptations to inertial forces during the stance phase of locomotion may be specific to impact or propulsion rather than to the entire period of stance.

ADAPTATIONS TO GRAVITATIONAL AND INERTIAL FORCES

Gravitational force is a product of mass and gravity level. The gravitational force, by our definition, is the same as Chang et al. (2000), and relates strictly to body weight.

Inertia relates strictly to mass. Our findings suggest that adaptations during locomotion to altered inertia differed from adaptations to altered net gravitational forces.

Others have studied the affect of altering gravitational and inertial forces on locomotion. Increased gravitational forces and inertia result in an increase in contact time and decrease in stride time during walking and running (LaFiandra et al., 2003; Chang et al., 2000). Griffin et al. (2003) found increases in peak GRFs during walking and Chang et al. (2000) found similar increases in GRFs during running. The increase in GRF is intuitive because of the need to decelerate and accelerate a larger mass during stance.

Other authors have investigated adaptations that occur when gravitational forces are decreased and inertia is maintained. Donelan and Kram (1997) found that contact time decreased and step length shortened during walking at speeds between 0.75 and 1.75 m s^{-1}, yet swing time did not change. Since stride time is the sum of contact time and swing time, a decrease in contact time coupled with a constant swing time results in a decreased stride time. Griffin et al. (1999) and Finch et al. (1991) also found that stride frequency, which is the inverse of stride time, did not significantly change during walking at 1.0 m s^{-1} with the reduction of gravity.

Those who investigated running with decreased body weight found opposite adaptations in kinematics. He et al. (1991) found that contact time decreased while stride time increased as gravitational force decreased. Millslagle et al. (2005) found similar increases in stride time. Farley and McMahon (1992) also found no increases in contact time with the reduction of gravity during running.

Both Chang et al (2000) and Newman et al. (1994) found decreases in peak vertical GRFs during running with decreased gravitational force and constant inertia. Similar to increased gravitational force results, the decrease in peak GRF when body weight is reduced makes sense since there is a reduced need to generate forces to decelerate and accelerate the center of mass.

Taken together, these studies suggest that an increase in inertial and gravitational force results in kinematic adaptations during walking and running that include increased contact time, decreased stride times, and increased GRF. Decreases in gravitational force with constant inertia resulted in decreased contact time, no change in stride time, and decreased GRF. Contact time and vertical GRF during walking and running may be directly influenced by gravitational forces. Stride time, however, may only adapt when gravitational force is decreased.

We found that adding inertia while maintaining gravitational force increases contact time and stride time during running. In the literature, it has been shown that maintaining inertia and decreasing body weight decreased these times. Therefore, gravitational force may not be the sole controlling factor of footfall patterns during running. The increased contact time that occurs with an increase in inertial force may allow the lower body musculature more time to develop force to propel the body upward. Rather than increasing force, contact time is increased to allow the vertical propulsive forces to act for a longer period. The mass of the person, which affects the inertial forces acting upon the body, may be the critical control factor.

The vertical GRF during walking and running is directly related to the gravity level (Chang et al., 2000). Our results suggest that during walking, gravitational force is the critical factor utilized for selection and execution of the gait pattern, although adaptations to increased inertia do occur. However, during running, inertial forces play a larger role in the control process. Bernstein (1967) theorized that motion requires the interaction between the central nervous system and the state of the position, velocity, and weight of the effected limbs. Our findings are consistent with this theory, because if gravitational forces were the main input when determining the motion patterns during locomotion, no adaptations to increased inertial force on the trunk should occur.

LIMITATIONS

While our original question that initiated this study was to investigate a potential enhancement to exercise countermeasures performed in microgravity, we tested our hypothesis in normal gravity using an overhead suspension system. Our testing location allowed data collection from multiple subjects in a controlled environment. The subjects could familiarize themselves with the testing environment, and testing sessions were not limited by factors that influence experiments in microgravity, such as limited sample size and availability to collect data. However, because we tested subjects in normal gravity, it is possible that our results would differ in a microgravity environment. The AIF were applied in a manner that could be used during spaceflight. However, in our experiment, the limbs were subject to normal gravity, and the gravitational forces resulting from the AIF were reduced by suspension with a harness. The harness could have influenced the adaptations that we measured.

APPLICATIONS TO SPACEFLIGHT EXERCISE

One of our intents in this investigation was to determine if locomotive exercise performed in reduced gravity could be enhanced with the addition of inertial force. Astronauts currently suffer from bone mineral density losses during long-duration spaceflight (Schneider et al., 1995; LeBlanc et al., 2000; Lang et al., 2004). Locomotive exercise is performed to create GRFs that may be beneficial for bone remodeling (Cavanagh et al., 2005; Schaffner et al., 2005). The GRFs developed during locomotion in weightlessness are less than those occurring at similar speeds in normal gravity

(Schaffner et al., 2005; De Witt et al., 2004). We hypothesized that adding inertial force may help to increase the vertical GRFs that occur during treadmill exercise, and thus enhance the current countermeasure. Our findings in normal gravity suggest that the addition of inertial force may increase GRFs during walking during space flight. However, adaptations in the gait pattern during running likely would mitigate increases in GRF. Confirmation of these suppositions could be gained only during a microgravity experiment.

List of symbols and abbreviations

AIF - added inertial force

GRF ground reaction force

ANOVA – analysis of variance

Acknowledgements

We wish to thank the members of the Exercise Physiology Laboratory at the NASA – Johnson Space Center for their help during this project. This project was funded by the Exercise Countermeasures Project at NASA – Johnson Space Center.

References

Alexander, R. McN. (1976). Mechanics of bipedal locomotion. In *Perspectives in experimental biology, vol. 1* (ed. P. S. Davies), pp. 493–504. Oxford: Pergamon Press.

Bernstein, N.A. (1967). The problem of the interrelation of coordination and localization. In *The coordination and regulation of movements* (ed. N.A. Bernstein), pp. 15-59. Oxford: Pergammon Press.

Cavanagh, P.R., Licata, A.A. and Rice, A.J. (2005). Exercise and pharmacological countermeasures for bone loss during long-duration space flight. *Grav. Space Biol*. **18**, 39-58.

Cham, R. and Refern, M.S. (2004). Gait adaptations during load carrying on level and inclined surfaces. *Occup. Ergonomics* **4**, 11-26.

Chang, Y.H., Huang, H.W.C., Hamerski, C.M. and Kram, R. (2000). The independent effects of gravity and inertia on running mechanics. *J. Exp. Biol.* **203**, 229-238.

Chou, L.S., Kaufman, K.R., Brey, R.H. and Draganich, L.F. (2001). Motion of the whole body's center of mass when stepping over obstacles of different heights. *Gait Posture* **13**, 17-26.

DeWitt, J.K., Schaffner, G., Laughlin, M.S., Loehr, J. and Hagan R.D. (2004). External load affects ground reaction force parameters non-uniformly during running in weightlessness. Presented at the 2004 Annual Meeting of the American Society of Biomechanics, Portland, OR.

Donelan, J.M. and Kram, R. (1997). The effect of reduced gravity on the kinematics of human walking: a test of the dynamic similarity hypothesis for locomotion. *J. Exp. Biol.* **200**, 3193-3201.

Farley, C.T. and McMahon, T.A. (1992). Energetics of walking and running: insights from simulated reduced-gravity experiments, *J. Appl. Physiol.* **6**, 2709-2712.

Finch, L., Barbeau, H. and Arsenault, B. (1991). Influence of body weight support on normal human gait: development of a gait retraining strategy, *Phys. Ther.* **71**, 842-855.

Grabowski, A., Farley, C.T. and Kram, R. (2005). Independent metabolic costs of supporting body weight and accelerating body mass during walking. *J. Appl. Physiol.* **98**, 579-583.

Griffin, T.M., Roberts, T.J. and Kram, R. (2003). Metabolic cost of generating muscular force in human walking: insights from load-carrying and speed experiments. *J. Appl. Physiol.* **95**, 172-183.

Griffin, T.M., Tolani, N.A. and Kram R. (1999). Walking in simulated reduced gravity: mechanical energy fluctuations and exchange. *J. Appl. Physiol.* **86**, 383 - 390.

He, J.P., Kram, R. and McMahon, T.A. (1991). Mechanics of running under simulated low gravity. *J. Appl. Physiol.* **71**, 863-870.

Iwamoto, J., Takeda, T. and Sato, Y. (2005). Interventions to prevent bone loss in astronauts during space flight. *Keio J. Med.* **54**, 55-59.

Karamanidis, K., Arampatzis, A. and Bruggemann, G.P. (2003). Symmetry and reproducibility of kinematic parameters during various running techniques. *Med. Sci. Sports Exerc.* **35**, 1009-1016.

LaFiandra, M., Wagenaar, R.C. and Obusek, J.P. (2003). How do load carriage and walking speed influence trunk coordination and stride parameters?, *J. Biomech.* **36**, 87-95.

Lang, T., LeBlanc, A., Evans, H., Lu, Y., Genant, H. and Yu, A. (2004). Cortical and trabecular bone mineral loss from the spine and hip in long-duration spaceflight. *J. Bone Miner. Res.* **19**, 1006-1012.

LeBlanc, A., Lin, C., Shackelford, L., Sinitsyn, V., Evans, H., Belichenko, O., Schenkman, B., Kozlovskaya, I., Oganov, V., Bakulin, A., Hedrick, T. and Feeback, D. (2000). Muscle volume, MRI relaxation times (T2), and body composition after spaceflight. *J. Appl. Physiol.* **89**, 2158-2164.

Lee, C.R. and Farley, C.T. (1998). Determinants of the center of mass trajectory in human walking and running. *J. Exp. Biol.* **201**, 2935-2944.

McCrory, J.L., Baron, H.A., Balkin, S. and Cavanagh, P.R. (2002). Locomotion in simulated microgravity: Gravity replacement loads. *Aviat. Space Environ. Med.* **73**, 625-631.

Millslagle, S., Levy, M. and Matack, N. (2005). Comparative kinematic measures of treadmill running with or without body weight support in runners. *J. Sports Sci.Med.* **4**, 583-589.

Munro, C.F., Miller, D.I. and Fuglevand, A.J. (1987). Ground reaction forces in running: a reexamination. *J. Biomech*, **20**, 147-155.

Newman, D.J., Alexander, H.L. and Webbon, B.W. (1994). Energetics and mechanics for partial gravity locomotion. *Aviat. Space Environ. Med.* **65**, 815-823.

Portney, L.G. and Watkins, M.P. (2000). Foundations of clinical research: applications to practice. 1st ed. Upper Saddle River, NJ: Prentice Hall.

Rubin, C.T. and Lanyon, L.E. (1985). Regulation of bone mass by mechanical strain magnitude. *Calcif. Tissue Int*. **37**, 411-417.

Schaffner, G., DeWitt, J.K., Bentley, J.R., Yarmanova, E., Kozlovskaya, I. and Hagan, R.D. (2005). Effect of subject loading device load levels on gait. NASA Technical Report TP-2005-213169.

Schneider, V., Oganov, V., LeBlanc, A., Rakmanov, A., Taggart, L., Bakulin, A., Huntoon, C., Grigoriev, A. and Varonin, L. (1995). Bone and body mass changes during space flight. *Acta Astronautica* **36**, 463-466.

Figure Legends

Figure 1. Data collection procedures showing the unweighting system and weighted vest.

Figure 2. Mean ensemble GRF trajectories during walking at all AM levels.

Figure 3. Mean ensemble GRF trajectories during running at all AM levels.