The Biomechanics of Exercise Countermeasures
NAGW-4421 (Years 1 & 2) & NAG5-6199 (Year 3)

FINAL REPORT

PERSONNEL

Prinicipal Investigator	**Department**
Peter R. Cavanagh, Principal Investigator	Center for Locomotion Studies (CELOS)

Faculty Support (Other)	**Department**
Steven Arnold, Statistician	Department of Statistics
Janice Derr, Statistician	Statistical Consulting Center, Department of Statistics
Neil Sharkey, Co-Investigator	Center for Locomotion Studies
Ge Wu, Co-Investigator	Center for Locomotion Studies

Staff Support	
Mary Becker	CELOS
David Lemmon	CELOS
Diane Plummer	CELOS
Douglas Tubbs	CELOS
Ronald Welling	CELOS

Graduate Student Support	**Majors**
Jean McCrory	CELOS and Kinesiology
Sandy Balkin	Statistics
John O'Gorman	Statistics
Marc Peterman	CELOS and Kinesiology
Andrew Hamel	CELOS and Mechanical Engineering
John Schwass	CELOS and Kinesiology
John Dingwell	CELOS and Kinesiology

Undergraduate Student Support	**Majors**
Heidi Baron	Kinesiology
Aquilah Couvson	Science and Engineering - MURA Program (Minority Undergraduate Research Assignment)
Brian D'Archangelo	Mechanical Engineering
Jeff Yablonovitz	Pre-medicine

BACKGROUND

The Penn State Zero-gravity Simulator (PSZS) is a device developed by the Center for Locomotion Studies (CELOS) to enable ground studies of exercise countermeasures for the bone loss that has been shown to occur during long-term exposure to zero gravity (0G). The PSZS simulates 0G exercise by providing a suspension system that holds an individual in a horizontal (supine) position above the floor in order to enable exercise on a wall-mounted treadmill. Due to this orientation, exercise performed in the PSZS is free of the force of gravity in the direction that would normally contribute to ground reaction forces. In order for movements to be more similar to those in 0G, a constant force suspension of each segment (equal to the segment weight) is provided regardless of limb position. During the preliminary development of the PSZS (supported by NASA grant NAG 9-379), CELOS researchers also designed an optional gravity-replacement simulation feature for the PSZS. This feature was a prototype tethering system that consisted of a spring tension system to pull an exercising individual toward the treadmill. The immediate application of the tethering system was to be the provision of gravity-replacement loading so that exercise in 0G- and 1G-loading conditions could be compared, and the PSZS could then be used to evaluate exercise countermeasures for bone loss during space flight. This tethering system would also be a model for the further refinement of gravity-replacement systems provided for

astronaut usage while performing prescribed exercise countermeasures for bone loss during long-term space flights.

OBJECTIVES

The objectives of the research conducted with this award have been:

1) to redesign the gravity replacement loading configuration to increase the load tolerance of individuals exercising in the PSZS with the desired end point being full body weight replacement,

2) to quantify the biomechanical similarities between fully loaded (100%BW) PSZS exercise and 1G exercise, and

3) to make preliminary determinations of localized bone strains in human cadaveric tibia during simulated loading conditions.

YEAR 1 PROGRESS

Phase 1

The first phase of the project was designed to evaluate alternate tethering configurations for the gravity replacement aspects of the PSZS. Eight males who were either students or instructors in the Penn State Navy ROTC program volunteered to be participants in the first phase of this study. They had an average age of 23.62±3.96 years, an average height of 159.12±2.97 cm, and an average mass of 72.33±4.26 kg. These subjects were recruited because they had similar body mass indices and physical activity levels to members of the astronaut corps. Constraints imposed by the PSZS limited subject body mass to less than 80 kg. The braces used to support subjects in the PSZS were not adaptable to most female body types; thus, male subjects were chosen. [Note: The suspension system was later redesigned and women subjects were tested in phase 3 experiments.]

The objective of the first phase was to quantify the ground reaction forces, tethering spring tensions, and subjective comfort ratings from subjects wearing one of four restraint harness designs with a 60% body weight load from the spring tensions while walking or running in the PSZS. The four spring conditions tested were: 1) "no springs," 2) "shoulder springs," 3) "waist springs," and 4) "both (waist and shoulder) springs." The waist harness consisted of a standard mountain climbing harness to which four springs were attached. The shoulder harness, to which four springs were attached, consisted of a pair of modified football shoulder pads. The "waist and shoulder" springs harness consisted of the combination of these two harnesses. Comfort was assessed by a modified Borg-like scale where "0" indicated no discomfort, "4" indicated moderate discomfort, and "10" indicated excruciating pain.

There were great variations in ratings of comfort ($p<0.001$) from subject to subject, which was not surprising due to the fact that people perceive pain differently. The order in which the conditions were presented had little bearing on comfort ($p>0.05$). This implies that subjects could well tolerate being hung in the PSZS because their final condition was no more uncomfortable than the first condition to which they were exposed. In addition to rating overall discomfort, the subjects were also asked how much discomfort they were feeling in their legs, hips, back, and shoulders. Tethering subjects by the waist caused discomfort in their legs, hips, and back. Not being tethered at the shoulders reduced discomfort in the shoulders. Similarly, not being tethered at the waist reduced discomfort at the waist. Not surprisingly, tethering subjects by the waist and shoulders, i.e. the "both springs" condition, was the most comfortable condition.

Spring tension variability was also evaluated during this phase of research. The spring tensions appeared to remain similar across conditions, regardless of the harness configuration worn by the subject. To allow for comparison between subjects, the spring tension variables from each subject were divided by that subject's body weight, thus "normalizing" the variables. The only factor that was significantly different was tension fluctuation, which increased in the "both springs" condition ($p = 0.004$). It was apparent during the testing of the "both springs" condition that there were times in a typical gait cycle when the springs lost all tension. This observation was verified on the videotapes of the spring experiments. Because eight springs were used, this condition was such that the springs did not have to be stretched far in order to reach a tension of 60% of body weight. Thus, the large fluctuation is a result of this loss in tension in the springs, whereas the springs never lost their tension in the "shoulder springs" and "waist springs" conditions. The order in which treatments were administered did not affect the tension in the springs.

The maximum ground reaction forces were highest in the "shoulder springs" condition ($p=0.002$). Likewise, the impulse, or the area under the ground reaction force curve, was also significantly higher in the "shoulder springs" condition ($p=0.001$). The following variables were significantly different between subjects ($p < 0.05$) and were not affected by the condition or the order in which the conditions were administered: contact time, impact force, time to maximum impact peak, and loading rate.

In summary, it was determined during Year 1 that the ground reaction forces were highest in the "shoulder springs" condition, but the reason for this is not clear. There were no differences in the average tensions between any of the spring conditions, although the average maximum tension and the amount of tension fluctuation was highest in the "both springs" condition. The comfort data showed that subjects complained of back, hip, and leg pain when the waist was tethered. Although no kinematic analysis was performed of the motion of the subject walking or running on the treadmill, it can be speculated that greater ground reaction forces were produced in the "shoulder springs" condition because the subjects felt freer to move their legs, taking longer, bigger strides, and thus producing higher ground reaction forces. Because the subjects frequently objected to the discomfort associated with waist spring attachment sites at the hips and because this fact clearly did affect the ground reaction forces, the location of the attachment of the carabiners were moved to another area in the hip region so that the subject was not in such discomfort. The new attachment site for the carabiners was also selected to make certain that the legs are unencumbered.

The unanswered question arising from these preliminary experiments was a determination of the reason that the "both springs" condition was the most comfortable. The "both springs" condition caused the tethering load to be distributed over a larger area of the body, but it also resulted in small time intervals during which certain springs were not stretched such that smaller tethering force might be felt. Because the average tensions were the same in all of the conditions, the first possibility is most likely the one that is true. However, this could not be proven until another experiment was performed in which the load in the tethering springs was increased to a point when the springs always have a tension.

The order in which the treatments were given was not a significant factor in the comfort or spring tension data, but it was significant in some of the ground reaction force data. Subjects apparently needed time to get acclimated to the unusual task being required of them (i.e. being lifted by the PSZS, being pulled by the spring harness system, and having to run on a treadmill on the wall), and the ground reaction force data thus shows a minor order effect.

In summary, the comfort data, spring tensions, and ground reaction forces would have lead to different conclusions as to which harness design was the most effective if considered separately. Therefore, it was determined that all of the factors have to be carefully considered in concert with one another when choosing a harness design for a gravity replacement system. However, bone demineralization is thought to occur in response to a lack of impact forces or the lack of high loading rates applied to the lower extremity. Thus, ground reaction forces were considered to be an integral part in the decision to choose a particular harness design.

Phase 2

The objective of the second phase of the experiment was to measure the ground reaction forces, tensions in the tethering springs, and subjective ratings of comfort from subjects running with gravity replacement loads (GRL) 60%, 80%, or 100% body weight applied from either the "waist and shoulder" springs harness (WSO) or the "shoulder springs" harness (SSO). This objective was met with two successive experiments.

Harness comfort was first evaluated using eight subjects running at 2.0 ms-1 for 3 minutes at each level of the GRL in the PSZS wearing each harness design. Subjective ratings of harness comfort, ground reaction forces, and GRL data were collected during the final minute of exercise. The results showed that 100% BW loading conditions were comfortably tolerated, although discomfort increased as the GRL increased. There were no differences in perceived comfort between harnesses using the newest harness design. The weight acceptance rate and the first and second peaks of the ground reaction force increased with increasing levels of the GRL, and subjects were able to tolerate a GRL of 100% BW well. The magnitude of the ground reaction force peaks and the weight acceptance rate were found to be related directly to the magnitude of the GRL.

A second experiment investigated ground reaction forces in both harness conditions. Ground reaction forces were measured during overground walking ($1.35\ \mathrm{m{\cdot}s^{-1}}$) and running ($2.68\ \mathrm{m{\cdot}s^{-1}}$) and fully-loaded locomotion at the same speeds in two restraint harness designs in the PSZS. Load cells measured the magnitude of the gravity replacement load in the PSZS conditions. It was determined from these experiments that maximum active forces were greater in overground walking and running than in the PSZS; however, weight acceptance rates were greater in the PSZS running conditions than in overground running. Large loads and rates of change of load were found to be generated at the feet during simulated zero gravity exercise, and further investigation was needed to determine whether these loads would be sufficient countermeasures to bone loss when used in an appropriate exercise program in space. The refinement of the gravity replacement system to provide a constant 1G load was, thus, a factor that was identified as requiring continued consideration during future design modifications.

YEAR 2 PROGRESS

The next phase of the study was designed to quantify the biomechanical similarities between fully loaded (100%BW) PSZS exercise and 1G exercise. Ground reaction forces, lower extremity joint movements, and electromyographic data were collected on 16 subjects during walking and running in four experimental conditions. Normal overground locomotion and locomotion on a conventional treadmill were the two "1G" conditions. Loads in the tethering harnesses were compared between the "shoulder springs only" and "waist and shoulder springs" conditions.

Each subject was instrumented with electrogoniometers aligned to the ankle, knee, and hip to quantify joint motion. Electromyographic electrodes were placed on the subject's left tibialis anterior, gastrocnemius, rectus femoris, vastus lateralis, biceps femoris, and gluteus maximus muscles to quantify the muscular activations. These electromyographic and joint kinematic data were then combined to determine the phases of isometric, concentric, and eccentric activations of each of the six muscles.

The average tensions in the tethering springs were slightly greater than body weight during walking in the two PSZS conditions and slightly less than body weight in the PSZS running conditions. The tension fluctuations were 13% and 18% of body weight in the "shoulder springs only" condition during walking and running respectively, and 22% and 36% of body weight in the "waist and shoulder springs" condition. It appears as if these fluctuations were related to many of the biomechanical differences between the 1G and PSZS conditions in the kinematic, ground reaction force, and electromyographic variables.

The major finding of the kinematic data was that in the PSZS, subjects had a tendency to "groucho" walk and run with their knees significantly more flexed ($p<0.05$) than during the "1G" conditions. In the stance phase of overground running, the knee was flexed to an average of 6.96° whereas in the PSZS, the average knee stance phase flexion was 11.72° (0° was considered a straight leg). This was most likely an attempt to reduce the discomfort of the tethering springs pulling on the body. The more the subjects' knees were flexed, the less the springs were stretched, thus reducing the tension.

The maximum ground reaction force ($p<0.05$) and the impulse ($p<0.005$) were significantly less in the PSZS conditions, although the rate at which the initial ground reaction force was applied in the PSZS conditions was greater than in the "1G" conditions ($p<0.05$). However, if the ground reaction force data was normalized to subject load instead of body weight, the ground reaction force curves were remarkably similar, with no differences in the maximum ground reaction force or the impulse. At the time of the maximum ground reaction force, the loads in the tethering harness were near their minimum. In the PSZS, the subjects did not have to push off with as much force as was required during the overground condition in order to propel themselves forward.

The muscular activations, in terms of the areas under the full-wave rectified EMG, of the tibialis anterior, gastrocnemius, rectus femoris, vastus lateralis were significantly greater in the PSZS conditions than in the 1G conditions ($p<0.004$). The timing of the muscular activations was similar between conditions, but the intensity of activation was greater in the PSZS. The activations of the biceps femoris and gluteus maximus were not affected by the experimental condition. It is therefore reasonable to assume that tethered treadmill exercise provides a sufficient countermeasure to the atrophy of these 6 muscles during spaceflight. The patterns of isometric, concentric, and eccentric activations were similar between conditions.

Researchers do not agree on the necessary stimulus for maintaining bone mass in space, and this uncertainty prevents clear conclusions to be drawn on the effectiveness of treadmill exercise in space. If high loading rates applied to the lower extremity are the crucial stimulus, then tethered treadmill exercise under 100% BW tethering load should prove to be a sufficient exercise countermeasure. However, if it is the maximum amount of force applied to the lower extremity that maintains bone homeostasis, then tethered treadmill exercise will prove ineffective unless the maximum ground reaction forces seen during overground running can be replicated during tethered exercise in spaceflight. However, it is clear thatmore attention must be paid to the manner in which the subject is tethered to the treadmill in order to provide the maximum benefits of treadmill exercise during spaceflight. Future work will be aimed at minimizing the fluctuation in the tethering springs so as to more closely resemble the constant pull of gravity, thus requiring the exercising person to "push off" the treadmill with a force of approximately 2-3 times body weight in running. Also, an attempt to minimize the harness discomfort in order to lessen the subjects' desire to "groucho" walk and run would also result in increased maximum ground reaction forces.

YEAR 3 PROGRESS

Assuming that bone homeostasis is modulated by strain-related factors, it is likely that the success of treadmill exercise countermeasures will be dependent on the magnitude of the gravity replacement loads (GRLs) used. It is possible that subject discomfort may prevent the use of 1G load substitution. However, the relationship between GRL and bone strain is also unknown. Thus, a new protocol was developed in the final year of the project to allow information gathered from human studies using the PSZS to be compared with bone strain data. The dynamic gait simulator (DGS), a unique cadaver modeling system, was used to measure dynamic bone strains at the human distal tibia under a representative range of GRLs in order to better predict the dose-response relationship between treadmill exercise in micro-gravity and localized bone strains. The DGS was used to provide dynamic loading of cadaveric limb preparations and mimic the kinetics and kinematics of the tibia, foot, and ankle during the stance phase of gait from heel-strike to toe-off. Physiologic actions of the extrinsic foot muscles are simulated using force feedback controlled linear actuators interfaced with the tendons of the specimen. Previous studies have shown that the ground reaction forces and plantar pressures under the specimens are well within physiologic ranges and demonstrate inter-subject variation consistent with that seen in live subjects. The timing and force output of six separate muscle groups (constituting the major dorsi- and plantar-flexors of the foot) were independently controlled with stepper motor-powered linear actuators. Thus, the PSZS and the DGS were combined to enable studies of the implications of exercise for maintaining bone quality during space flight.

Using the DGS system, we found bone strains to be linearly related to external GRLs (R^2>0.75) and established equations whereby strain stimuli at given locations can be calculated from external loads. The distribution of strains indicates that the primary mode of tibial loading is in bending, with little variation in the neutral axis over the stance phase of gait. The greatest maximum (tensile) and minimum (comprehensive) principal strains were found to develop on the anterior crest and posterior aspect of the tibia, respectively. The relative strain response to a given GRL dosage was found to be site-specific, with the largest dose-response gradient occurring at the anterior and posterior sites furthest removed from the neutral axis of bending.

Thus, using a robust cadaver model, we have been able to show in a small group of cadaver legs that decreased GRLs produce proportional linear decrements in tibial strains. More than 75% of the variance in internal bone strain response was explained by external gravity replacement load (which was also proportional to external ground reaction force). Predictive dose-response equations have been developed through which a strain stimulus at a given site may be computed for any GRL. The overall peak strain for each condition was compressive and occurred along the posterio-medial border of the tibial cross section. Maximum tensile strains occurred anteriorly on the tibial crest. The distribution of tensile and compressive strains and the relative invariance of the neutral axis orientation over the stance phase indicate that the primary mode of loading at the distal tibia is bending, with muscle forces (notably the triceps surae) acting as the principal modulators of bone strain. Additionally, peak strains occur at sites other than those typically measured in live human subjects.

Our data suggest that *in-situ* bone strains can be reliably related to the external-loading environment. It is reasonable to assume, therefore, that previously proposed equations could be use to calculate a theoretical bone maintenance stimulus using only GRL as input. This relationship is important for understanding the theoretical potential and most effective implementation of various exercise countermeasures to bone loss during space flight.

RELATED PRODUCTIVITY & SUPPORT

Year 1 Intramural Funding:

Dr. Cavanagh and Dr. Derr acquired additional support during Year 1 of the project through two intramural awards from the Pennsylvania Space Grant Consortium.

The first intramural award was obtained under the STIR (Stimulating Interdisciplinary Research) program. This program is designed to strengthen interdisciplinary collaborations in research, and the small award made to Drs. Cavanagh and Derr enabled the formation of a solid interdisciplinary student support team for the PSZS project. The award provided funds which enabled Jean McCrory (who was partially supported under main NASA project award) to be assigned to this research on a year-round basis. In addition, it provided a semester of support for Sandy Balkan (Graduate Assistant, Statistics) and provided hourly summer wages for two undergraduate students, Heidi Baron (Exercise and Sports Science) and Brian D'Archangelo (Mechanical Engineering). Because this project provided a single source of funding for students from three different units representing three different Penn State colleges, faculty and students from extremely diverse disciplines were provided a forum for direct interaction and at the same time enhanced ongoing efforts.

Year 2 Intramural Funding

An intramural award was obtained through the Minority Undergraduate Research Assignments (MURA) Program funded by NASA at Penn State, which is designed to increase involvement of minority freshmen in NASA-related projects with the ultimate goal of increasing retention of these students in science and engineering programs. MURA has provided hourly wages and related support to enable the involvement of Aquilah Couvson, a Penn State freshman, to participate in this project and gain research experience and exposure.

Jean McCrory received a $200 dissertation grant from the Penn State College of Health and Human Development, which covered additional supplies and materials for aspects of the project, that contributed specifically toward her associated dissertation research.

The Pennsylvania Space Grant Consortium has also awarded an $8,000 grant to CELOS through the STIR Program (Stimulating Interdisciplinary Research) to provide wages for two undergraduate senior honors students and two graduate assistants from the Departments of Kinesiology and Statistics to receive compensation to work during the Summer of 1997 on evaluations of various exercise countermeasures performed by Russian cosmonauts during space flight.

Year 2 Extramural Funding

The Center for Locomotion Studies concurrently performed contracted evaluations for Wyle Laboratories (formerly Krug Life Sciences) under a proposal entitled, "Verification Experiments for the Treadmill Vibration Isolation and Stabilization System (TVIS)." This contract included four tasks as follows: Task 1: SLD and Harness Verification - Ground Based Studies; Task 2: Stability of TVIS - Ground Based Studies; Task 3: Stability of TVIS - Flight Expert Analysis; and Task 4: Biomechanics of Locomotion on TVIS - Ground Based.

Year 3 Extramural Funding

A proposal that was submitted in response to NRA 96-HEDS-04 from the NASA Gravitational Biology and Biomedical Research and Countermeasures Program was funded during the final period of support for this grant. That proposal will provide funding to continue the current line of investigation to enable ground validation and subsequent implementation in the International Space Station of biomechanical experiments designed using the PSZS to provide further insight into reasons for loss of bone mineral during prolonged exposure to microgravity.

Other Activities

Dr. Peter R. Cavanagh is member of the Science Working Group for the International Space Station Human Research Facility and a member of the NASA US-Russian Working Group on the International Space Station Treadmill.

BIBLIOGRAPHY

Refereed publications:

Davis BL, Cavanagh PR, Sommer HJ, III, Wu G. Ground reaction forces during locomotion in simulated microgravity. Aviation, Space and Environmental Medicine 1996; 67(3):235-242.

McCrory JL, Baron HA, Derr JA, Davis BL, Cavanagh PR. Subject load-harness interaction during zero-gravity treadmill exercise (Abstract). Presented to the 20th Annual Meeting of the American Society of Biomechanics (ASB), October 17-19, 1996, Georgia Institute of Technology, Atlanta, GA 1996.

McCrory JL, Schwass JP, Connell RB, Cavanagh PR. In-shoe force measurements from locomotion in simulated zero gravity during parabolic flight (Abstract). Program of the 5th EMED User Meeting, Penn State University, University Park, PA, 1996.

Dingwell JB, Cavanagh PR, Lloyd T. Quantifying daily load bearing activity: results of ground reaction forces measured over a ten hour day (Abstract #13). Gait & Posture 1997; 5(2):p. 172. Abstracts of the 2nd Annual Meeting of the North American Society of Gait and Clinical Movement Analysis, April 1997, Chicago

McCrory JL, Baron HA, Derr JA, Davis BL, Cavanagh PR. Subject load-harness interaction during zero-gravity treadmill exercise. In: Conference Proceedings of the 21st Annual Meeting of the American Society of Biomechanics. 1997: 235-236. Presented to the 21st Annual Meeting of the American Society of Biomechanics, September 24-27, 1997, at Clemson University, Clemson, SC

McCrory JL, Derr J, Cavanagh PR. Ground reaction forces in 1G and simulated zero-gravity running. In: Conference Proceedings of the 21st Annual Meeting of the American Society of Biomechanics. 1997: 234-235. Presented to the 21st Annual Meeting of the American Society of Biomechanics, September 24-27, 1997, at Clemson University, Clemson, SC

Peterman M, McCrory JL, Sharkey NA, Piazza S, Cavanagh PR. Exercise countermeasures for bone loss during space flight (Abstract). In: First Biennial Space Biomedical Investigators' Workshop. 1998: 96-9. Abstracts prepared for a workshop to take place January 11-13, 1999, League City, Texas. National Aeronautics and Space Administration (NASA) and Universities Space Research Association (USRA)

Thornton WE, Cavanagh PR, Buczek FL, Milliron MJ, Davis BL. The kinematics of treadmill locomotion in space. In: Allard P, Cappozzo A, Lundberg A, Vaughan CL Eds. Three-Dimensional Analysis of Human Locomotion. Chichester: John Wiley & Sons, 1998: 375-88.

McCrory JL, Lemmon DR, Sommer HJ, III, Prout B, Smith D, Korth DW, Lucero J, Greenisen M, Moore J, Kozlovskaya I, Pestov I, Stepansov V, Miyakinchenko Y, Cavanagh PR (The TVIS Study Group). Evaluation of a treadmill with vibration isolation and stabilization (TVIS) for use on the International Space Station. Journal of Applied Biomechanics 1999; 15: 292-302.

Peterman MM, Hamel AJ, Sharkey NA, Piazza SJ, Cavanagh PR. The implications of reduced ground reaction forces during space flight for bone strains (Abstract). Abstract Book, 23rd Annual Meeting of the American Society of Biomechanics. 1999; 156-7. A podium presentation to the 23rd Annual Meeting of the American Society of Biomechanics (ASB), October 21-23, 1999, University of Pittsburgh, Pittsburgh, PA

Peterman MM, Hamel AJ, Cavanagh PR, Piazza SJ, Sharkey NA. In vitro modeling of human tibial strains during exercise in micro-gravity. Submitted (December 22, 1999) to the Journal of Biomechanics.

Non-refereed publications:

Faking weightlessness. Clues to better space exercise. Health & Human Development News, Penn State University, College of Health and Human Development, Fall 1992.

Doing the Locomotion. Research/Penn State, Vol. IV, 1995.

Academic dissertations or theses:

McCrory JL. A Biomechanical Comparison of 1-G and Fully-Loaded Simulated Zero-Gravity Locomotion. (PhD Thesis), Summer 1997)

Baron HA. Investigating phase two of the NASA research grant on tethered exercise in simulated zero-gravity (Honors Thesis), Spring 1996

ORIGINAL RESEARCH

Ground Reaction Forces During Locomotion in Simulated Microgravity

B. L. DAVIS, P. R. CAVANAGH, H. J. SOMMER, III, and G. WU

DAVIS BL, CAVANAGH PR, SOMMER HJ III, WU G. *Ground reaction forces during locomotion in simulated microgravity.* Aviat Space Environ Med 1996; 67:235–42.

Background: Significant losses in bone density and mineral, primarily in the lower extremities, have been reported following exposure to weightlessness. Recent investigations suggest that mechanical influences such as bone deformation and strain rate may be critically important in stimulating new bone formation. ***Hypothesis:*** It was hypothesized that velocity, cadence, and harness design would significantly affect lower limb impact forces during treadmill exercise in simulated zero-gravity (0G). ***Methods:*** A ground-based hypogravity simulator was used to investigate which factors affect limb loading during tethered treadmill exercise. A fractional factorial design was used and 12 subjects were studied. ***Results:*** The results showed that running on active and passive treadmills in the simulator with a tethering force close to the maximum comfortable level produced similar magnitudes for the peak ground reaction force. It was also found that these maximum forces were significantly lower than those obtained during overground trials, even when the speeds of locomotion in the simulator were 66% greater than those in 1G. Cadence had no effect on any of the response variables. The maximum rate of force application ($DFDT_{max}$) was similar for overground running and exercise in simulated 0G, provided the "weightless" subjects ran on a motorized treadmill. ***Conclusions:*** These findings have implications for the use of treadmill exercise as a countermeasure for hypokinetic osteoporosis. As the relationship between mechanical factors and osteogenesis becomes better understood, results from human experiments in 0G simulators will help to design in-flight exercise programs that are more closely targeted to generate appropriate mechanical stimuli.

A VARIETY OF STUDIES on both humans and animals have shown that many systems of the body undergo adaptation on exposure to a hypogravic environment. Among these changes are decreases in muscle mass, loss of blood volume, decreased cardiovascular fitness, and loss of bone mineral. With respect to bone loss, Kakurin (11) analyzed bones of three cosmonauts tragically killed by accidental decompression of their Salyut-1 space station. Examination of these tissues showed foci of resorption that are typically seen after acute immobilization, and X-ray densitometry measurements showed reductions in calcaneal density of about 17%. What makes this bone loss all the more remarkable is that it occurred during a flight that lasted only 24 d.

Wheton (25) reports a gradual increase in urinary calcium excretion for the first 2–4 wk, followed by a plateau level varying from 60–100% more than the control level. Also of concern is that a long-term follow-up of Skylab bone demineralization found that spaceflight leads to a statistically significant loss of bone mineral as much as 5 yr post-flight (22). In addition, there is general consensus that the "weightbearing" bones of the lower extremity are at greater risk during spaceflight (7,26). Various countermeasures have been suggested, including anabolic hormones, thiazide diuretics and lowered oxygen partial pressure (10), nutritional supplements (20), skeletal compression with penguin suits (7) and, most frequently, exercise (1,14). With the exception of exercise, these countermeasures have generally shown limited potential in overcoming bone loss (1,21).

In an attempt to prevent disuse osteoporosis, Brannon et al. (2) showed that only a small amount of isometric exercise was needed to prevent bone rarefaction in bedridden patients. In another bed rest study, however, seven different physical interventions were used in an attempt to prevent disuse osteoporosis: exercise with a pulley system, static compression, intermittent compression, lower body negative pressure (LBNP), cyclic LBNP, 20 lbs impact, and 35 lbs impact on the heel (21). Of these interventions, only quiet standing for 3 $h \cdot d^{-1}$ was partially effective as a countermeasure. In the case of space travel, it has been suggested that daily weightbearing exercise equivalent to 4 h of walking on Earth should be investigated (1).

More recent investigations suggest that mechanical influences may indeed be critically important in reducing the severity of spaceflight-induced osteoporosis. Rubin and Lanyon (19) showed that peak bone strains greater than a threshold of 0.001 can result in substantial new bone formation. Other mechanical stimuli that are considered important are joint forces (24), strain duration (9), rate of loading (8,9), and fluid flow within bone (18). Many of these stimuli are related--for instance, a large impact force will produce high strains and strain rates as well as sudden changes in fluid movement within bone.

The common feature among most exercises that have been performed in space is that they are virtually devoid

From the Center for Locomotion Studies (P. R. Cavanagh, G. Wu) and the Department of Mechanical Engineering (H. J. Sommer, III), Penn State University, University Park, PA; and the Department of Biomedical Engineering, Cleveland Clinic Foundation, Cleveland Park, OH (B. L. Davis).

This manuscript was received for review in October 1994. It was revised and accepted for publication in June 1995.

Address reprint requests to: Peter R. Cavanagh, Center for Locomotion Studies, Room 10, Intramural Building, Penn State University, University Park, PA 16802.

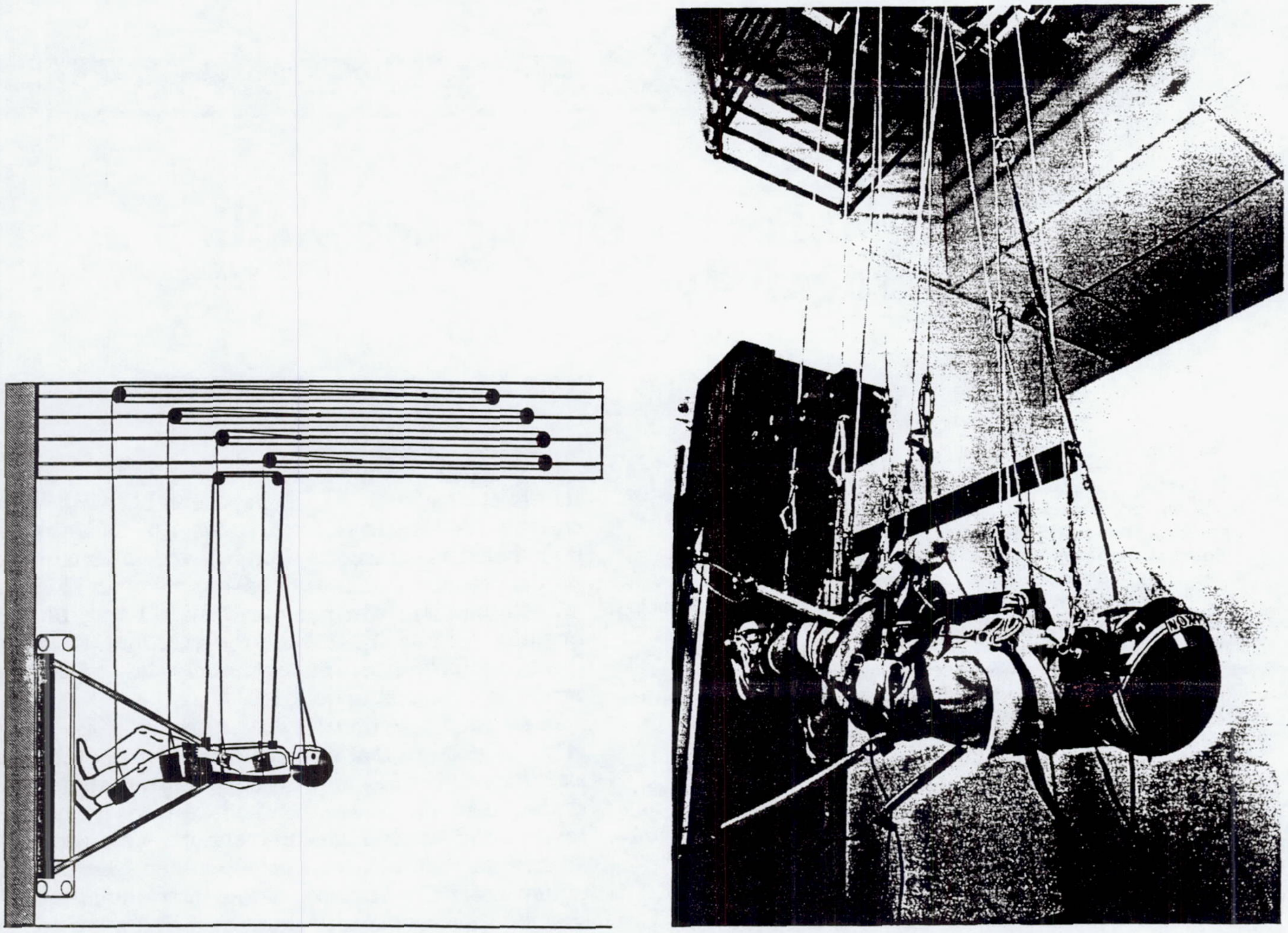


Fig. 1. Supine suspension system used to simulate microgravity: a) schematic drawing; b) photograph of system in use.

suspension system and the subject adopted a supine position as shown in **Fig. 1a and b.** The table was then elevated using a customized jack until the subject was approximately in line with the center of the wall-mounted forceplate. At this stage the pulleys in the ZLS closest to the subject's feet were positioned directly above the appropriate centers of mass. Specially made limb supports molded from a thermoplastic orthotic material were fitted to each forearm, upper arm, thigh and shank in such a way that the latex cords could be attached at points coinciding with the centers of mass. After attaching the cords, the pulleys towards the rear of the ZLS needed to be positioned such that the tension in each cord balanced the weight of the limb segment being suspended. Before lowering and removing the examination table, the torso and head were fitted with a harness and helmet respectively and these were suspended using 6 mm diameter mountaineering rope. Once the table had been removed and a safety mat had been placed below the subject, further refinements to the tension in the cords could be performed to ensure that the subject's limbs were aligned with the head and torso in a horizontal position. In the event of the subject being too low (or high), the position of the upper body was altered by adjusting mountaineering ascenders that controlled the length of the 6 mm rope.

Experimental Protocol

Each subject was tested under various conditions. The factors that were controlled during trials in the ZLS included speed, cadence, treadmill mode (motorized vs. passive), and tension in the elastic springs that tethered each subject to the treadmill. Three levels of tension were set: 60% body weight (BW), 60% BW plus 80N, and 60% BW minus 80N (i.e., approximately 51% BW, 60% BW, and 69% BW). At the highest setting most subjects felt that further increases in the tether force would have been too uncomfortable for exercise to continue. Four levels of speed were studied in the ZLS: slow walking (0.5 $m \cdot s^{-1}$), normal walking (1.0 $m \cdot s^{-1}$), slow running (3 $m \cdot s^{-1}$), and faster running (4 $m \cdot s^{-1}$). Each of these speeds were associated with three cadence levels. **Table II** summarizes the conditions studied.

In the case of overground locomotion, one walking and one running speed were selected. These were chosen to be in the ratio of 0.6:1 (overground: ZLS). The rationale for this was that for each subject the tether tension was considerably less than the force of one bodyweight (i.e., comparing identical speeds when the effective "weight" was unequal would not have been as valid as comparing trials in which the product of weight and speed was kept constant). In addition, our object was to determine if 1G-like forces could be elicited during gait in reduced

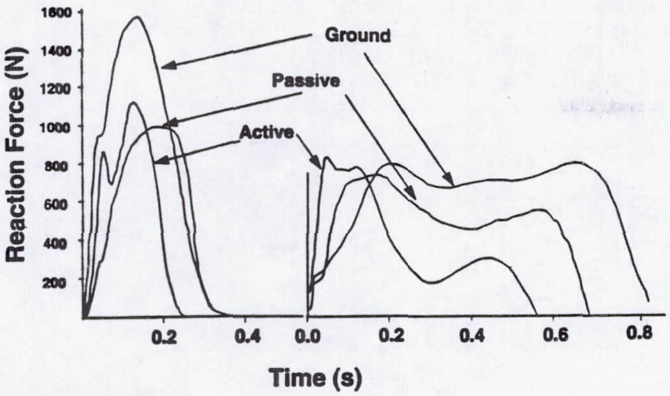


Fig. 2. Typical results for different modes of locomotion for one of the subjects. The right graph shows data for walking (in 1G, velocity = 1.1 $m \cdot s^{-1}$, and in simulator, velocity = 1.7 $m \cdot s^{-1}$) and the left graph refers to running trials (in 1G, velocity = 2.3 $m \cdot s^{-1}$, and in simulator, velocity = 4 $m \cdot s^{-1}$).

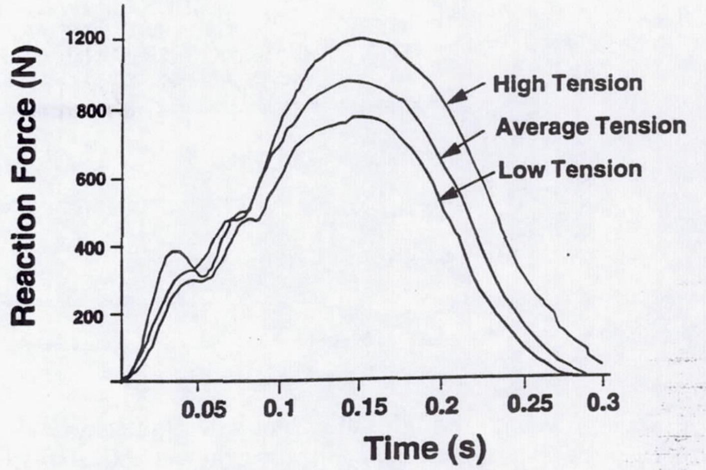


Fig. 3. Typical example of the dependence of ground reaction force profiles on the tension in the gravity replacement system for running at 4 $m \cdot s^{-1}$ on an active (motorized) treadmill in simulated reduced gravity.

level and average level of tether tension in common. Similarly, to make comparisons between passive and motorized treadmill exercise, for the 1.7 and 4 $m \cdot s^{-1}$ trials, all subjects were tested at all 3 tether tensions at the average cadence setting. The end result is that each subject had 16 trials in the ZLS.

In summary, the dependent variables included GRF_{max}, $DFDT_{max}$, impulse, stance time, tibial acceleration, torso, hip and knee angles. The independent variables were gravity (1G, 0G), velocity (slow walking, normal walking, slow running, faster running), load on harness (low tension, average tension, high tension), treadmill type (active, passive), and cadence (minimum, average, maximum). ANOVA models were used to test whether there were significant differences when walking or running under various gravitational or treadmill conditions.

RESULTS

Overground locomotion: Each subject performed overground walking and running trials at speeds of 1.0 and 2.3 $m \cdot s^{-1}$, respectively. Typical results are shown in **Fig. 2** (graphs labelled "Ground"), and the overall means of several dependant variables are listed in **Table III.**

Supine locomotion on a passive treadmill: In the same way that subjects were required to both walk and run in 1G, they performed trials at two speeds in the ZLS with the treadmill motor switched off. It should be noted that walking or running on a passive treadmill was quite exhausting. Therefore, it would have been impractical to request subjects to repeat the trials over and over until they achieved a certain belt speed. Hence, there was

TABLE III. MEAN RESULTS FOR 1G OVERGROUND WALKING AND RUNNING.

	Run (2.3 $m \cdot s^{-1}$)	Walk (1 $m \cdot s^{-1}$)	Ratio (Run/Walk)
GRFmax (N)	1576	785	2:1
DFDTmax ($N \cdot s^{-1}$)	42900	10200	4.2:1
impulse ($N \cdot s$)	298	445	1:1.5
hip range (deg)	34.1	33.8	1:1
knee range (deg)	63.8	60.3	1.05:1

some latitude in the speeds that were deemed acceptable for the 12 subjects. The mean walking speed was 0.836 $m \cdot s^{-1}$ ($\pm$0.136 SD) and for running, 1.375 $\pm$ 0.244 $m \cdot s^{-1}$.

During locomotion on the passive treadmill, it was found that the subject's mass category had no significant effect on any of the features of the GRF profile. Further, different tether tensions only affected impulses under the foot, with the highest tension causing a significantly ($p \leq 0.05$) higher value (284 N.s) than for the lowest setting (244 N.s). One possible reason for other response variables not being significantly affected by tether tension is that, for passive trials, subjects tended to lean forward and this affected the tensions in the springs that tethered them to the treadmill. Since these changes depended on a host of inter-related variables (e.g., subject posture and stature), it is likely the tension levels were not as controlled as they were for trials on an active treadmill.

By far the most dramatic effects were caused by the different velocity settings. Compared to walking on the passive treadmill, running on the device caused significantly higher readings for $DFDT_{max}$, GRF_{max}, tibial acceleration, and significantly lower values for stance duration, impulse and hip range of motion ($p = 0.0005$). The fact that running on the passive treadmill was very taxing means that it is unlikely astronauts could run at a tether tension level of 60% BW for any length of time if the resistance of the treadmill was similar to that used in this study. Even in a typical 1G situation, it is unlikely that a person would freely choose to exercise on a passive treadmill with this level of resistance.

Supine locomotion on active treadmill: As was the case for passive trials, neither subject mass category nor cadence had any significant ($p < 0.05$) effect on the dependent variables. However, the force in the tether significantly affected the maximum loading rate ($DFDT_{max}$), reaction force (GRF_{max}), and impulse. This can be ascertained by inspecting typical GRF's for running at 4 $m \cdot s^{-1}$, shown in **Fig. 3.**

In analyzing the effects of velocity, follow-up tests using Tukey's multiple range procedure shown in **Table IV** were obtained. It can be seen that although there are

TABLE VI. ANALYZING THE RESPONSE VARIABLES FOR WHICH THERE WERE SIGNIFICANT INTERACTIONS BETWEEN MODE AND VELOCITY. G, A, AND P REFER TO OVERGROUND, ACTIVE AND PASSIVE TRIALS. NUMERICAL CHARACTERS "1" AND "4" REFER TO WALKING AND RUNNING, RESPECTIVELY. HORIZONTAL LINES INDICATE MEAN RESPONSES THAT ARE NOT SIGNIFICANTLY DIFFERENT.

	G1	P1	P4	A1	A4	G4
DFDTmax ($N \cdot s^{-1}$)	10229	13424	21584	24888	28688	42952
	P1	A1	G1	A4	P4	G4
GRFmax (N)	688	729	785	946	1002	1576
	A4	G4	P4	A1	P1	G1
Support (s)	0.233	0.326	0.347	0.517	0.743	0.754
	A4	A1	P4	G4	P1	G1
Impulse ($N \cdot s$)	114	165	201	298	335	446

active running, though both were still significantly less than for the 1G situation. Running on the active treadmill, however, could be done for a longer duration since it required less effort on the part of the person exercising. On the other hand, one could also argue that a passive treadmill is lighter, simpler, causes less vibration than an active device, and, if animal data (13) are considered appropriate for the human skeleton, one could maintain that only a few cycles of high load are necessary.

A major factor distinguishing active from passive exercise was the rate of change of force. Running on an active treadmill produced DFDTmax values that were no different ($p \leq 0.05$) from those found during overground running, whereas passive running resulted in values that could be matched by simply walking on an active treadmill. This issue of rate of change of force is an extremely important one, since Rubin and Lanyon (24) found that low strain magnitudes do not prevent bone deterioration *except at high strain rates*. The problem in extrapolating their research to the problem of prescribing exercise for long-term spaceflight is that, to date, there is no direct way of relating bone strain rate to the rate of change of DFDT. Thus, although the minimum threshold for which an externally applied load is osteogenic still needs to be determined, one can state that if passive exercise met this threshold, then so would exercise on an active treadmill. The converse is not true, since active treadmill running may result in a certain threshold being met that passive exercise could not achieve.

At this stage one might ask whether the continuous application of shock loads during running would predispose some form of injury. The work that is usually cited to support this notion is that of Radin and colleagues (16) who have suggested that in animals, cartilage deterioration can be caused by mechanical impacts. Studies on humans, however, (12) have shown no correlation between long-distance running and degenerative joint disease. In fact, the study by Lane et al. (12) showed the cartilage thickness to slightly favor runners, and, of particular relevance to astronauts, the bone density of runners was 40% higher than control subjects.

CONCLUSIONS

The ZLS offered a unique opportunity to study the biomechanics of tethered locomotion in "weightlessness." The kinematic measurements of running on a passive treadmill matched those obtained from actual inflight film data very closely. With regard to recommending either motorized or passive treadmill exercise, the data support the notion that whatever can be achieved with a passive device can at least be equalled, or in the case of DFDTmax and tibial accelerations, exceeded by using a motorized device. Since logistic considerations would favor a smaller, lighter device, further investigations are needed to ascertain whether a passive device can result in the, as yet unknown, threshold for "acceptable" external loads being met. Until such time, the conservative option would be to recommend active exercise as a countermeasure to spaceflight-induced osteoporosis.

It can also be concluded that neither active nor passive running at a tether tension setting of 60%BW produced similar values for GRFmax when compared to slow running on the ground, although exercise on an active treadmill produced responses for DFDTmax that were not statistically different from those found for overground running. Although the exact relationship between bone strain and external loads still needs to be investigated, it is possible that the reasonably high DFDTmax responses

TABLE VII. COMPARISON OF GAIT KINEMATICS IN THE SHUTTLE AND IN THE SIMULATOR DURING RUNNING AND WALKING.

Speed Variables	Max. Hip	Min. Hip	Hip Range	Max. Knee	Min. Knee	Knee Range
Fast run						
ZLS, avg. tension	62.9	11.5	50.5	155.5	89.8	65.7
ZLS, low tension	59.1	11.9	47.2	153.0	90.6	62.4
Shuttle	55.5	11.9	45.5	156.0	94.1	61.9
Walk						
ZLS, avg. tension	57.6	3.9	53.7	162.2	97.4	64.8
ZLS, low tension	60.3	5.4	54.9	161.8	96.9	64.8
Shuttle	46.4	10.8	35.7	172.1	120.3	51.8

green 105

SUBJECT LOAD-HARNESS INTERACTION DURING ZERO-GRAVITY TREADMILL EXERCISE

Jean L. McCrory, Heidi A. Baron, [1]Janice A. Derr, [2]Brian L. Davis, and Peter R. Cavanagh

Center for Locomotion Studies , Department of Kinesiology and [1]Statistical Consulting Center
Penn State University, University Park, PA 16802

[2]Department of Biomedical Engineering, The Cleveland Clinic Foundation, Cleveland, OH

INTRODUCTION

When astronauts exercise on orbit, a subject load device (SLD) must be used to return the subject back to the supporting surface. The load in the SLD needs to be transferred to the body by a harness which typically distributes this load between the pelvis and the shoulders. Through the use of a zero-gravity simulator, this research compared subject comfort and ground reaction forces during treadmill running at three levels of subject load (60%, 80%, and 100% of body weight) in two harness designs ("shoulder only" and "waist and shoulder ").

REVIEW AND THEORY

Exercise will almost certainly play an integral part in minimizing the adverse effects of space travel on the body, particularly bone mineral loss and muscular atrophy. It is hypothesized that an effective exercise regimen would elicit loads on the lower extremities that resemble those encountered on Earth (Cavanagh, 1986; Convertino and Sandler, 1995). No testing has been done in space to quantify the ground reaction forces to which the lower extremities are exposed, but it is believed that these forces are much less than those experienced in 1-G (Cavanagh, 1987).

The Penn State Zero-Gravity Simulator (PSZS Davis et al. 1996), is a device which suspends subjects horizontally from multiple latex cords, with each cord negating the weight of a different limb segment. A treadmill mounted on the wall under the PSZS enables subjects to run in simulated zero-gravity. The SLD has, in the past, consisted of a set of 4 springs attached to a harness, with the waist of the subject feeling the entire pull of the SLD. With this system, the subjects could only tolerate an artificial gravity of 60% of 1-G (Davis et al. 1996). Astronauts currently wear a harness system in which the SLD pulls both at the waist and shoulders (Greenisen and Edgerton, 1994), although the tension in these springs has not been quantified. However, it is likely that previous SLDs have only provided loads less than Earth gravity (Cavanagh 1986, 1987).

The purpose of this study was to quantify ground reaction forces, subject load, and subjective ratings of comfort from subjects wearing one of two harness designs under loads of 60%, 80%, and 100% of body weight while running in the PSZS. The objective was to gain insight into the effectiveness of the present countermeasures against bone mineral loss and muscular atrophy in space.

PROCEDURES

Eight subjects (mean age 29.4 ±4.5 years, mean height 176.6 ±9.0 cm and mean mass 73.3±5.3 kg) participated in this study. Two harness configurations were assessed: a "shoulder only" design, in which 4 springs were attached to shoulder pads worn by the subject, and "waist and shoulder " design, in which 4 springs were attached to the shoulder pads and 4 to a waist harness. Three levels of load (60%, 80%, and 100% of body weight) were randomly administered in each harness design. Ground reaction forces were measured via a Kistler force plate mounted within the treadmill belt. Load cells measured tension in the SLD. A modified Borg scale was used to assess the levels of discomfort. Subjects ran at a speed of 1.96 m/s for 3 minutes during each condition. A period of 3 minutes rest was given between conditions. Data were collected at 500 Hz.

RESULTS

The level of discomfort increased significantly ($p<0.05$) as the subject load increased from 60% to 80% to 100% body weight. Also, on a scale from 0 (no discomfort) to 10 (excruciating pain), the maximum levels of discomfort at 100% BW load averaged 2.3 ± 0.6 in the "shoulder only" condition and 2.5 ± 0.8 in the "waist and shoulder " condition ($p<0.05$).

The following subject load variables were measured: maximum load, time to maximum load, minimum load, time to minimum load, average load, and load fluctuation. By definition, the subject loads were significantly different between the loading conditions of 60%, 80%, and 100% of body weight ($p<0.05$). When comparing the two harness designs, the "waist and shoulder " design resulted in a lower minimum load and average load, while the load fluctuation was greater ($p<0.05$). Selected load variables for the full body weight loading conditions are shown in Table 1.

Table 1: Subject Load variables during a 100% BW load.

Load Variables (% BW)	Shoulder Only	Waist & Shoulder
Maximum	112.6 ± 2.6	110.5 ± 3.0
Minimum	90.0 ± 2.5	70.8 ± 2.8
Average	101.4 ± 2.3	91.7 ± 2.7
Fluctuation	22.5 ± 1.3	39.7 ± 1.4

The following ground reaction force variables were measured: contact time, maximum impact force, time to maximum impact force, maximum propulsive force, time to maximum propulsive force, impulse, and loading rate. In both harness designs, the maximum impact force, maximum propulsive force, impulse, and loading rate were significantly different between loading conditions ($p<0.05$). Results for the "waist and shoulder " conditions are shown in Figure 1.

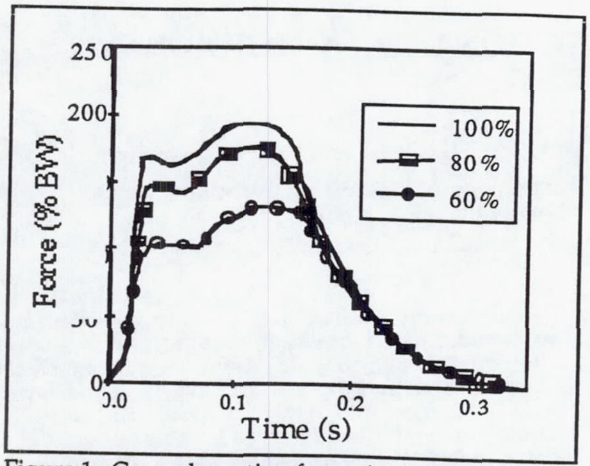


Figure 1: Ground reaction forces in the "waist and shoulder " condition in each of the subject loads.

When comparing the two harness designs, the maximum propulsive peak and the impulse were significantly greater in the "shoulder only " harness configuration ($p<0.05$). The ground reaction force curves at 100% load are shown in Figure 2.

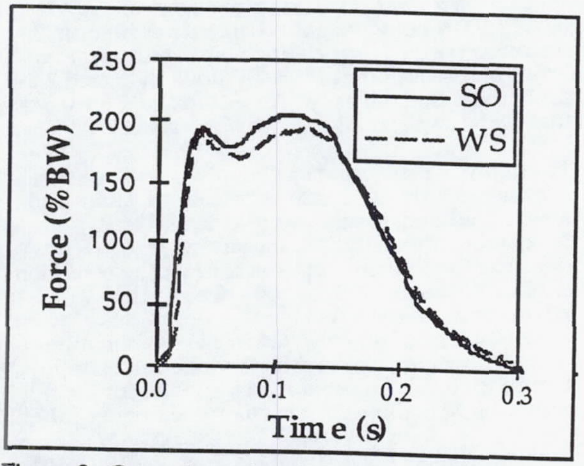


Figure 2: Ground reaction force curves at a load of 100% BW. (Harness design: SO= shoulder only, WS= waist and shoulder)

DISCUSSION

The clear dependence of peak reaction force on subject load is apparent from Figure 1. This highlights the importance of maximizing subject load if countermeasures are to generate 1-G like loads on the lower extremity. At 100% load, the peak ground reaction force was significantly greater for the "shoulder only" harness configuration. The amount of discomfort from the SLD and harness was perceived to be in the slight to moderate range at 100% loading. However, in both harness configurations, the shapes of the ground reaction force curves of subjects running in the PSZS were characteristic of the "groucho running" force curves reported by McMahon et al. (1987), indicating that the subjects were running in a slightly crouched (hips and knees flexed) position. This was most likely an attempt to reduce the discomfort caused by the SLD pulling on the body.

The effectiveness of tethering running as a countermeasure against bone mineral loss and muscle atrophy is believed to be dependent upon the presence of 1-G type forces. Unless a harness can be designed which will alleviate pressure felt at the SLD attachment sites, astronauts will tend to do "groucho running" to lessen the pain of the harness, thereby also attenuating the ground reaction forces. Another possibility is that the altered gait patterns result from subject loads which are locally applied (at the hips and shoulders) compared to the more global action of gravitation force.

REFERENCES

Borg, G.V. MSSE, 14, 377-387,1982.

Cavanagh, P.R. et al. A Final Report to Krug International, 1-126, 1987.

Cavanagh, P.R. Workshop on Exercise Prescription for Long-Duration Space Flight, NASA Conference Publication 3051, 61-67, 1986.

Cavanagh, P.R. and Lafortune, M.A. J. Biom., 13, 397-406, 1980.

Convertino, V.A., and Sandler, H. Acta Astro., 35, 253-270, 1995.

Davis, B.L. et al. Av. Space Env. Med. 67, 235-242, 1996.

Greenisen, M.C. and Edgerton, V.R. In: Space Physiology and Medicine, (pp. 194- 210), Lea & Febiger, 1994.

McMahon, T.A. et al. J. App. Phys., 62, 2326-2337, 1987.

ACKNOWLEDGMENTS

This research was supported by NASA grant NAGW-4421 and by a STIR award under the Pennsylvania Space Grant Consortium.

IN-SHOE FORCE MEASUREMENTS FROM LOCOMOTION IN SIMULATED ZERO GRAVITY DURING PARABOLIC FLIGHT

Jean L. McCrory, John P. Schwass, Richard B. Connell*, and Peter R. Cavanagh
Center for Locomotion Studies, Penn State University, University Park, PA 16802, USA
*Krug - Life Sciences Division, Houston, TX 77058, USA

Introduction. No effective countermeasure for space-induced bone loss has yet been identified. It has been hypothesized that an effective exercise regimen would elicit loads on the lower extremity which resemble those encountered on Earth[1]. Although a treadmill has been used on shuttle flights, the loads to which the lower extremity was exposed have not yet been quantified. It is believed that these loads are much less than the loads experienced in 1G[2]. The purpose of this study was to determine the magnitude of lower extremity loading during tethered treadmill exercise in a 0G environment.

Methods. Data were collected on five subjects (avg. ht. 177.3±10.1cm, avg. mass 78.3±18.0kg) onboard the KC-135, a NASA airplane used to simulate periods of zero gravity through parabolic flight. Subjects ambulated at 4 speeds: a walk (1.56m/sec), fast walk (2.0m/sec) slow jog (2.75m/sec), and jog (3.35m/sec) on the NASA treadmill operated in either a passive or motorized mode. Each subject wore a harness connected to the Subject Load Device (SLD) to tether them to the treadmill. The tension in the SLD was subjectively adjusted for comfort by each subject. Force data were collected at 60Hz using Pedar insoles. The number of parabolas per subject was variable due to motion sickness and hardware problems.

Results. Analysis of the insole data showed that the average SLD load was only 35.2% BW, although the values ranged from 20.1% to 56.6%. Maximum ground reaction force values increased with increasing speed and were not affected by treadmill mode. The impulse was higher during walking with the treadmill in the passive mode than in the active mode, but this difference diminished with increasing speed. Subjects tended to run on their forefeet, as shown from the extremely small heel impulse values. At higher speeds, heel contact was absent, while forefoot impulse became more pronounced.

Discussion. All force values were lower than those reported from 1G studies, where typical peak ground reaction forces are 1.2xBW and 2.5xBW for walking and running, respectively[3]. At every speed, the ratio of the rearfoot to forefoot impulse was much lower than reported from 1G studies, and this ratio decreased with increasing speed[4].

Conclusions. If the exposure to forces similar to those in 1G is a requirement for countermeasures against space-induced osteoporosis, the loads in the SLD must be greatly increased and should be directly measured before exercise.

Table 1. Results of selected variables

	Walk		**Fast Walk**		**Slow Jog**		**Jog**	
	Motor	*Passive*	*Motor*	*Passive*	*Motor*	*Passive*	*Motor*	*Passive*
n (# parabolas)	5	6	4	1	10	3	24	3
Maximum GRF (% BW)	0.81± 0.16	0.90± 0.10	1.06± 0.34	0.77	1.05± 0.29	1.49± 0.04	1.46± 0.13	1.44± 0.06
Impulse(%BW*sec)	0.20± 0.05	0.38± 0.04	0.21± 0.02	0.26	0.16± 0.04	0.22± 0.01	0.21± 0.03	0.21± 0.01
Heel Impulse (%BW*sec)	0.06± 0.03	0.07± 0.01	0.05± 0.05	0.1	0.02± 0.01	0.00± 0.00	0.01± 0.01	0.00± 0.00
Forefoot Impulse (%BW*sec)	0.12± 0.03	0.27± 0.04	0.13± 0.05	0.14	0.14± 0.03	0.21±0.00	0.17± 0.02	0.21± 0.01

References

1. Convertino VA, Sandler H. *Acta Astronautica* 1995; 35: 253-270.
2. Cavanagh PR, et al. *Aviat., Space, and Env. Med* 1992; 63: 482-485.
3. Ounpuu S. *Clinics in Sports Med.* 1994; 13: 843-863.
4. Hennig EM, Milani TL. *J. Appl. Biom.* 1995; 11: 299-310.

QUANTIFYING TEN-HOUR LOAD BEARING ACTIVITIES IN A GROUP OF ADOLESCENT WOMEN

Jonathan B. Dingwell[1], Thomas Lloyd[2], and Peter R. Cavanagh[1]

[1] Center for Locomotion Studies, Penn State University, University Park, PA 16803
[2] Department of Obstetrics and Gynecology, Hershey Medical Center, Hershey, PA 17033

INTRODUCTION

Many applications of gait analysis require an examination of more than the single foot-ground contact that is often sampled. For example, load bearing activities (LBA's) play an important role in the development and maintenance of bone strength (Smith et al., 1989, and Chilibeck et al., 1995) but typical laboratory analysis gives no indication of cumulative loading, or of patterns of loading over the course of one or many days. This is due in large part to the lack of a convenient, accurate, and reproducible means of measuring LBA's during daily activities. A load monitoring device originally described by Breit and Whalen (1994) offers many new possibilities for long term monitoring during unrestricted activities of daily living. The first aim of the present study was to replicate this device and use it to collect continuous ground reaction force data from both feet for a ten hour period for a group of adolescent women. The second aim of this study was to use this data to quantify the daily loading patterns in these women and estimate their "daily stress stimulus" as defined by Beaupre et al. (1990).

REVIEW AND THEORY

It has been shown that decreased physical activity levels are associated with decreases in bone mineral content, and likewise that increased physical activity levels are associated with increases in bone mineral content (Smith et al., 1989, and Chilibeck et al., 1995). However, the exact nature of the relationship between long term lower extremity loading and bone remodeling is still poorly understood. Beaupre et al. (1990) defined a relationship for relative changes in ground reaction forces and "daily stress stimulus" (1) under two different conditions.

$$\frac{\Psi_2}{\Psi_1} = \left[\frac{\sum n_2 (GRF_2)^M}{\sum n_1 (GRF_1)^M}\right]^{1/M} \qquad (1)$$

Where Ψ_1 and Ψ_2 are the two daily stress stimulus values, M is an experimentally determined exponent, and n is the number of loading cycles at a given load level, GRF. It has been hypothesized that values of M between 4 and 6 represent "normal active baselines" (Whalen et al., 1988).

PROCEDURES

The Personal Force Monitoring Device (PFMD) consists of a pair of capacitative force monitoring shoe insoles and signal amplifier (Electronic Quantification, Inc.), a Tattletale microprocessor (Onset Computer, Inc.) with a four megabyte PCMCIA card for data storage, and an elastic waist belt to which these electronics are attached. The entire system is powered by four 9-volt batteries, weighs just 1.5 kg, and can be worn comfortably throughout the course of a full day's activities.

Ten hours of continuous ground reaction force data were sampled at 25 Hz from both feet of two subjects during unrestricted activities. Subject 1 was a sedentary subject, and Subject 2 was an athletic subject who ran cross-country in the afternoons. Data were downloaded from the PFMD and analyzed on a Pentium PC. To quantify the loading history for each subject, raw voltage data were converted to percent body weight (%BW), and peak force values above 15 %BW were extracted and plotted in a histogram distribution. Histograms were generated by dividing the external loads into equally spaced categories (or "bins") from 15 %BW to 300 %BW, and counting the number of peak forces that occurred within each bin. A ten hour "Stress Stimulus" measure (Ψ) was calculated for each subject according to the following equation adapted from Beaupre et al. (1990):

$$\Psi = \left[\sum_{i=1}^{k} n_i (GRF_i)^M \right]^{1/M} \tag{2}$$

Where k was the number of histogram bins, n was the number of above threshold force peaks within each bin, and the value of M was varied from 4 to 6.

RESULTS

Each subject demonstrated distinctly different patterns of activity and distributions of peak loads over the course of the ten hour day (Figures 1 and 2). Distinct periods of activity and inactivity can be seen in the Load-Time History plots of both subjects. The increased level of forces encountered by the athletic Subject 2 compared to the sedentary Subject 1 are evident from both the Load-Time History and Peak Histogram plots.

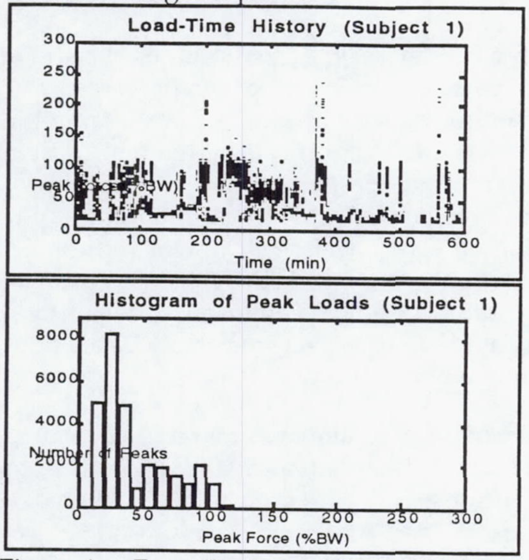


Figure 1 - Force Peaks and Peak Histogram from a 10 Hr Trial for an "Inactive" Subject.

While the total number of above threshold peaks was much greater for Subject 1 (33,216) than for Subject 2 (19,411), the number of force peaks greater than 125 %BW experienced by Subject 1 was only 64, compared to 2980 such peaks for Subject 2. This increase in the number of peaks at high stress levels is reflected in the 10 hr. Stress Stimulus (Ψ) values for each subject (Figure 3), where Ψ was greater for Subject 2 across all values of the exponent M despite the lower number of total force peaks.

DISCUSSIONS

Although the PFMD is less accurate than in-ground force platforms, it can collect data during extended real life activities. Thus, data collected using this device have the potential to provide considerable insight into the mechanisms of bone remodeling. It has been hypothesized (Whalen et al., 1988, and Beaupre et al., 1990) that the response of bone to mechanical load is more a function of the magnitude than the number of peak loads. Recent evidence suggests that this theory may not be sufficient to distinguishing the daily stimulus required for bone maintenance from that required for bone remodeling (Adams et al., 1997). The large difference in higher magnitude peaks seen between these two subjects, and the subsequent differences in Ψ values would allow these hypotheses to be tested in groups of subjects with diverse profiles of physical activity. This would represent an important step in gaining a better understanding of the relationship between lower extremity loading and bone remodeling.

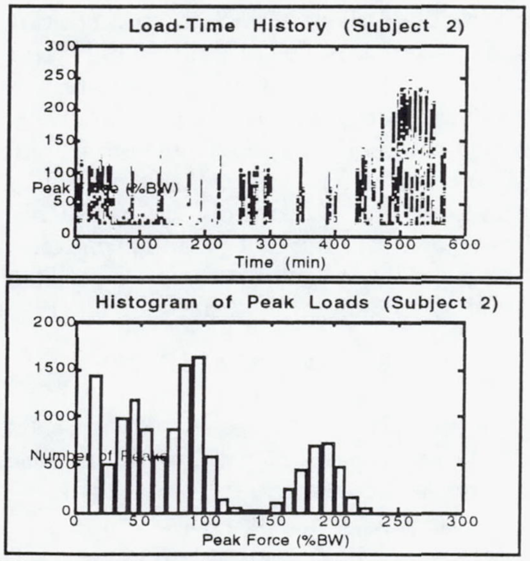


Figure 2 - Force Peaks and Peak Histogram from a 10 Hr Trial for an "Active" Subject.

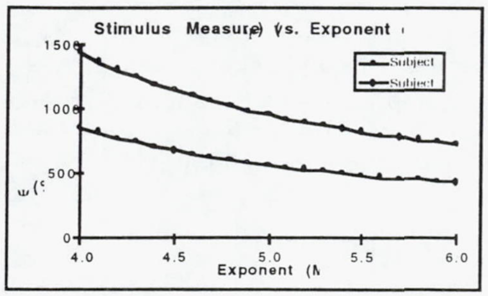


Figure 3 - Daily Stress Stimulus (Ψ) computed for exponent values (M) ranging from 4 to 6.

This information is important for understanding the role of daily activity in the acquisition of bone early in life, which is believed to be important for preventing osteoporosis later in life (Johnston et al., 1992). It is also important for understanding the relationship between daily loading patterns and bone loss that occurs in the weightless environment of space (Whalen et al., 1988, Beaupre et al., 1990), or the development of stress fractures in at risk populations such as athletes of military recruits (Grimston et al., 1991). The exploration of these and other conditions appears to be within the realm of the PFMD.

REFERENCES

Adams et al., *J. of Biomech..*, In Press, 1997.

Beaupre GS et al., *J. of Orthop. Res.*, 8: 651 - 661, 1990.

Breit GA and Whalen RT, *Proc. of the 18th Ann. Meeting of the Amer. Soc. of Biomech.*, Ohio State U., pp. 231 - 232, 1994.

Chilibeck PD et al., *Sports Medicine*, 19: 103 - 122, 1995.

Grimston S et al., *Intl. J. of Sports Biomech.*, 7: 293 - 309, 1991.

Johnston CC et al., *Osteoporosis Intl.*, Suppl. 1: S54-S55, 1992.

Smith EL et al., *Bone and Mineral Res.*, 6: 139 - 173, 1989.

Whalen RT et al., *J. of Biomech.*, 21 (10): 825 - 837, 1988.

SUBJECT LOAD-HARNESS INTERACTION DURING ZERO-GRAVITY TREADMILL EXERCISE

Jean L. McCrory, Heidi A. Baron, [1]Janice A. Derr, [2]Brian L. Davis, and Peter R. Cavanagh

Center for Locomotion Studies , Department of Kinesiology and [1]Statistical Consulting Center
Penn State University, University Park, PA 16802

[2]Department of Biomedical Engineering, The Cleveland Clinic Foundation, Cleveland, OH

INTRODUCTION

When astronauts exercise on orbit, a subject load device (SLD) must be used to return the subject back to the supporting surface. The load in the SLD needs to be transferred to the body by a harness which typically distributes this load between the pelvis and the shoulders. Through the use of a zero-gravity simulator, this research compared subject comfort and ground reaction forces during treadmill running at three levels of subject load (60%, 80%, and 100% of body weight) in two harness designs ("shoulder only" and "waist and shoulder ").

REVIEW AND THEORY

Exercise will almost certainly play an integral part in minimizing the adverse effects of space travel on the body, particularly bone mineral loss and muscular atrophy. It is hypothesized that an effective exercise regimen would elicit loads on the lower extremities that resemble those encountered on Earth (Cavanagh, 1986; Convertino and Sandler, 1995). No testing has been done in space to quantify the ground reaction forces to which the lower extremities are exposed, but it is believed that these forces are much less than those experienced in 1-G (Cavanagh, 1987).

The Penn State Zero-Gravity Simulator (PSZS Davis et al. 1996), is a device which suspends subjects horizontally from multiple latex cords, with each cord negating the weight of a different limb segment. A treadmill mounted on the wall under the PSZS enables subjects to run in simulated zero-gravity. The SLD has, in the past, consisted of a set of 4 springs attached to a harness, with the waist of the subject feeling the entire pull of the SLD. With this system, the subjects could only tolerate an artificial gravity of 60% of 1-G (Davis et al. 1996). Astronauts currently wear a harness system in which the SLD pulls both at the waist and shoulders (Greenisen and Edgerton, 1994), although the tension in these springs has not been quantified. However, it is likely that previous SLDs have only provided loads less than Earth gravity (Cavanagh 1986, 1987).

The purpose of this study was to quantify ground reaction forces, subject load, and subjective ratings of comfort from subjects wearing one of two harness designs under loads of 60%, 80%, and 100% of body weight while running in the PSZS. The objective was to gain insight into the effectiveness of the present countermeasures against bone mineral loss and muscular atrophy in space.

PROCEDURES

Eight subjects (mean age 29.4 ±4.5 years, mean height 176.6 ±9.0 cm and mean mass 73.3±5.3 kg) participated in this study. Two harness configurations were assessed: a "shoulder only" design, in which 4 springs were attached to shoulder pads worn by the subject, and "waist and shoulder " design, in which 4 springs were attached to the shoulder pads and 4 to a waist harness. Three levels of load (60%, 80%, and 100% of body weight) were randomly administered in each harness design. Ground reaction forces were measured via a Kistler force plate mounted within the treadmill belt. Load cells measured tension in the SLD. A modified Borg scale was used to assess the levels of discomfort. Subjects ran at a speed of 1.96 m/s for 3 minutes during each condition. A period of 3 minutes rest was given between conditions. Data were collected at 500 Hz.

RESULTS

The level of discomfort increased significantly ($p<0.05$) as the subject load increased from 60% to 80% to 100% body weight. Also, on a scale from 0 (no discomfort) to 10 (excruciating pain), the maximum levels of discomfort at 100% BW load averaged 2.3 ± 0.6 in the "shoulder only" condition and 2.5 ± 0.8 in the "waist and shoulder " condition ($p<0.05$).

The following subject load variables were measured: maximum load, time to maximum load, minimum load, time to minimum load, average load, and load fluctuation. By definition, the subject loads were significantly different between the loading conditions of 60%, 80%, and 100% of body weight ($p<0.05$). When comparing the two harness designs, the "waist and shoulder " design resulted in a lower minimum load and average load, while the load fluctuation was greater ($p<0.05$). Selected load variables for the full body weight loading conditions are shown in Table 1.

Table 1: Subject Load variables during a 100% BW load.

Load Variables (% BW)	Shoulder Only	Waist & Shoulder
Maximum	112.6 ± 2.6	110.5 ± 3.0
Minimum	90.0 ± 2.5	70.8 ± 2.8
Average	101.4 ± 2.3	91.7 ± 2.7
Fluctuation	22.5 ± 1.3	39.7 ± 1.4

The following ground reaction force variables were measured: contact time, maximum impact force, time to maximum impact force, maximum propulsive force, time to maximum propulsive force, impulse, and loading rate. In both harness designs, the maximum impact force, maximum propulsive force, impulse, and loading rate were significantly different between loading conditions ($p<0.05$). Results for the "waist and shoulder " conditions are shown in Figure 1.

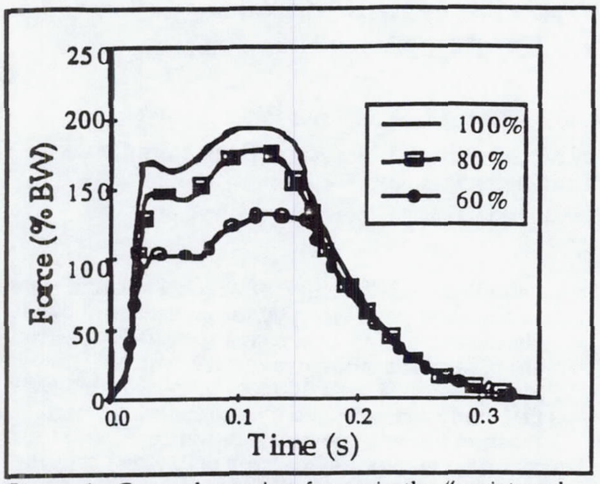


Figure 1: Ground reaction forces in the "waist and shoulder " condition in each of the subject loads.

When comparing the two harness designs, the maximum propulsive peak and the impulse were significantly greater in the "shoulder only " harness configuration ($p<0.05$). The ground reaction force curves at 100% load are shown in Figure 2.

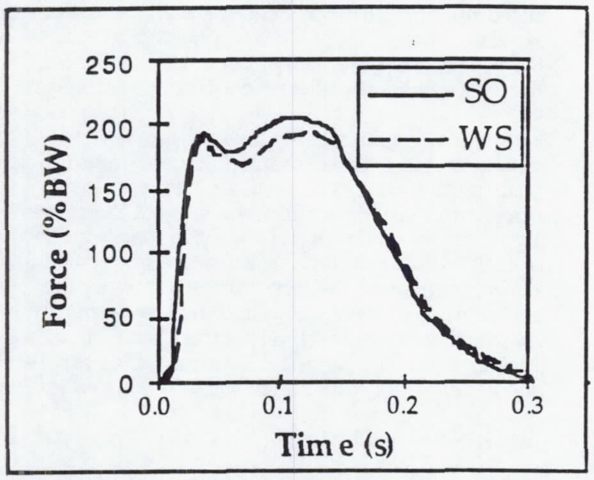


Figure 2: Ground reaction force curves at a load of 100% BW. (Harness design: SO= shoulder only, WS= waist and shoulder)

DISCUSSION

The clear dependence of peak reaction force on subject load is apparent from Figure 1. This highlights the importance of maximizing subject load if countermeasures are to generate 1-G like loads on the lower extremity. At 100% load, the peak ground reaction force was significantly greater for the "shoulder only" harness configuration. The amount of discomfort from the SLD and harness was perceived to be in the slight to moderate range at 100% loading. However, in both harness configurations, the shapes of the ground reaction force curves of subjects running in the PSZS were characteristic of the "groucho running" force curves reported by McMahon et al. (1987), indicating that the subjects were running in a slightly crouched (hips and knees flexed) position. This was most likely an attempt to reduce the discomfort caused by the SLD pulling on the body.

The effectiveness of tethering running as a countermeasure against bone mineral loss and muscle atrophy is believed to be dependent upon the presence of 1-G type forces. Unless a harness can be designed which will alleviate pressure felt at the SLD attachment sites, astronauts will tend to do "groucho running" to lessen the pain of the harness, thereby also attenuating the ground reaction forces. Another possibility is that the altered gait patterns result from subject loads which are locally applied (at the hips and shoulders) compared to the more global action of gravitation force.

REFERENCES

Borg, G.V. MSSE, 14, 377-387,1982.

Cavanagh, P.R. et al. A Final Report to Krug International, 1-126, 1987.

Cavanagh, P.R. Workshop on Exercise Prescription for Long-Duration Space Flight, NASA Conference Publication 3051, 61-67, 1986.

Cavanagh, P.R. and Lafortune, M.A. J. Biom., 13, 397-406, 1980.

Convertino, V.A., and Sandler, H. Acta Astro., 35, 253-270, 1995.

Davis, B.L. et al. Av. Space Env. Med. 67, 235-242, 1996.

Greenisen, M.C. and Edgerton, V.R. In: Space Physiology and Medicine, (pp. 194- 210), Lea & Febiger, 1994.

McMahon, T.A. et al. J. App. Phys., 62, 2326-2337, 1987.

ACKNOWLEDGMENTS

This research was supported by NASA grant NAGW-4421 and by a STIR award under the Pennsylvania Space Grant Consortium.

GROUND REACTION FORCES IN 1G AND SIMULATED ZERO-GRAVITY RUNNING

Jean L. McCrory, Janice Derr*, and Peter R. Cavanagh
Center for Locomotion Studies , Department of Kinesiology and *Statistical Consulting Center
Penn State University, University Park, PA 16802

INTRODUCTION

This research compared ground reaction forces during overground (1G) running and zero-gravity (0G) simulated treadmill running at a full body weight load in two restraint harness designs.

REVIEW AND THEORY

Exercise will almost certainly play an integral part in minimizing the bone mineral loss and muscular atrophy that occur during spaceflight. It is hypothesized that an effective exercise regimen would elicit loads on the lower extremities that resemble those on Earth (Convertino and Sandler, 1995). No on-orbit testing has yet quantified the forces to which the lower extremity has been exposed, but it is believed that, to date, these forces have been much less than the forces experienced in 1-G (Cavanagh, 1987).

The Penn State Zero-Gravity Simulator (PSZS Davis et al. 1996) is a device which suspends subjects horizontally from multiple latex cords, with each cord negating the weight of a limb segment. A treadmill mounted on the wall under the PSZS enables subjects to run in simulated 0G. Subjects wear a harness to which a number of springs, which provide a gravity replacement load, are connected. The opposite end of each spring is connected to the side of the treadmill. During exercise, astronauts currently wear a similar harness in which the spring tethering load pulls at both the waist and shoulders (Greenisen and Edgerton, 1994).

The purpose of this study was to compare ground reaction forces from subjects wearing one of two harness designs under a 100% BW load in the PSZS with data from the same subject running across the laboratory floor. The objective was to gain insight into the effectiveness of the present exercise countermeasures for bone mineral loss and muscular atrophy in space.

PROCEDURES

Sixteen subjects (age 22.9±6.9 yrs, height 178.1±6.68 cm, and mass 72.8±5.8 kg) participated in this study. Subjects ran at 2.68 m/s. One Kistler force plate recorded normal force data as subjects ran across the laboratory floor and another, mounted within the treadmill belt, measured normal ground reaction forces of subjects in the PSZS. Two PSZS subject load configurations were assessed: a "shoulder only" design (SSO), in which 4 springs were attached to shoulder pads worn by the subject, and "waist and shoulder " design (WSS), in which 4 springs were attached to the shoulder pads and 4 to a waist harness. Load cells measured tension in the springs. Data were collected at 500 Hz.

RESULTS

All subjects could tolerate a 100% body weight load applied through the harness. The maximum active force was significantly greater in the 1G condition, although the timing of this event was the same in all conditions (Figure 1, Table 1). The magnitude of the passive peak was similar in all conditions, but this peak occurred earlier in the PSZS conditions, resulting in a significantly greater loading rate. The impulse was greater in the 1G condition.

Table 1: Ground Reaction Force results

	1G	SSO	WSS
Max. Active GRF (%BW)	*240.61 ±7.04	180.04 ±3.77	159.75 ±3.97
% stance Max Active GRF	43.57 ±2.89	43.80 ±1.60	43.99 ±1.64
Max. Passive GRF (%BW)	159.29 ±7.34	161.84 ±3.90	150.08 ±4.01
% stance Max Passive GRF	*15.01 ±1.11	10.64 ±0.59	10.07 ±0.60
Load Rate (BW/sec)	40.60 ±2.85	*51.97 ±1.52	*51.81 ±1.57
Impulse (BW sec)	*0.41 ±0.14	0.33 ±0.01	0.30 ±0.01

* indicates that $p<0.01$

The tension in the tethering springs fluctuated by 17.89±1.25%BW in the SSO condition and 36.83±1.30%BW in the WSS condition as the subject's COM oscillated toward and away from the treadmill surface. The average subject load was 96.36±1.59 %BW in SSO and 88.95±1.66%BW in the WSS condition. The flight phase impulse in the 1G was only approximately 87% of the flight phase impulse in the PSZS conditions.

DISCUSSION

The maximum force occurred at approximately the time of the minimum subject load (Figures 1 and 2), which was less that body weight at this time. The subject was pushing off less because a smaller force was needed to overcome the "gravitational" load. If the force curves are normalized to the gravity in 1G or the instantaneous subject load in the PSZS instead of body weight, the curves look similar in all conditions, as shown in Figure 3.

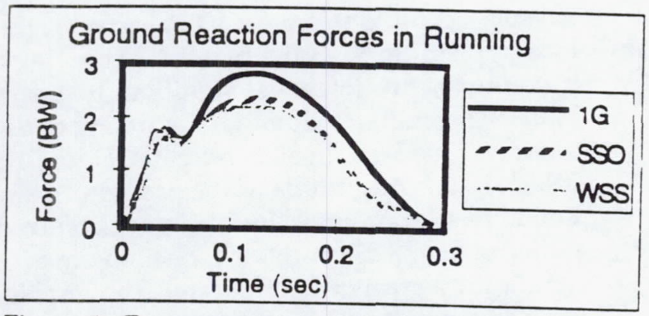


Figure 1: Representative ground reaction force curves from the running conditions.

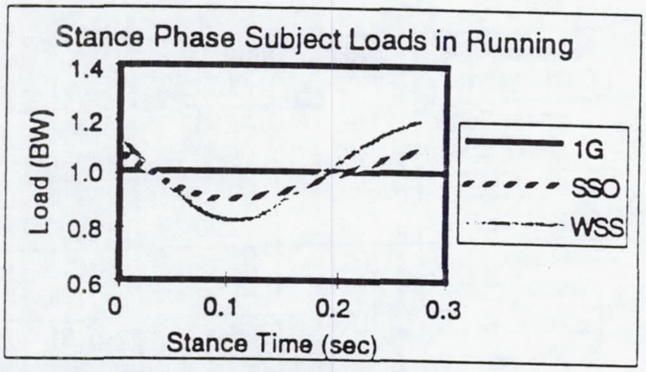


Figure 2: Representative subject load curves during the stance phase of running.

Because the flight impulse was greater in the PSZS conditions, the subject had a higher impact velocity - resulting in a greater loading rate.

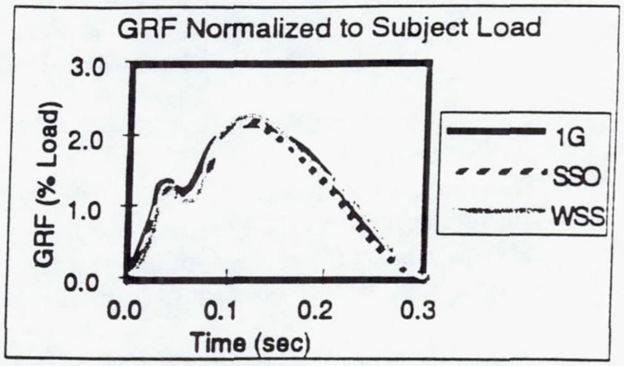


Figure 3: Maximum ground reaction force, for one subject, normalized to subject load instead of body weight.

The conclusions of this study are entirely dependent upon what aspects of the 1G forces are important for maintaining bone and muscle. If the aim is to equal 1G peak forces (Whalen et al., 1988), the fluctuation of subject load should be minimized since this appears to be responsible for requced normal GRFs during mid-stance. If it is greater loading rates that result in increased bone density (Lanyon and Rubin, 1984), then fully loaded 0G treadmill exercise will be effective as long as it is of the necessary duration.

ACKNOWLEDGMENTS

This research was supported by NASA grant NAGW-4421.

REFERENCES

Cavanagh, P.R. et al. A Final Report to Krug International, 1-126, 1987.

Convertino, V.A., and Sandler, H. Acta Astro., 35, 253-270, 1995.

Davis, B.L. et al. Av. Space Env. Med. 67, 235-242, 1996.

Greenisen, M.C. and Edgerton, V.R. In: Space Physiology and Medicine, (pp. 194- 210), Lea & Febiger, 1994.

Lanyon, L.E. and Rubin, C.T. J Biom. 17, 897-905, 1984.

Whalen, R.T. et al. J Biom. 21, 825-837, 1988.

First Biennial Space Biomedical Investigators' Workshop

January 11-13, 1999 - League City, Texas

National Aeronautics and Space Administration (NASA)
Universities Space Research Association (USRA)

EXERCISE COUNTERMEASURES FOR BONE LOSS DURING SPACE FLIGHT:
A Method for the study of Ground Reaction Forces and their Implications for Bone Strain

M. Peterman, J. L. McCrory, N. A. Sharkey, S. Piazza, and P. R. Cavanagh
The Center for Locomotion Studies, Penn State University, University Park, PA 16802

INTRODUCTION

Effective countermeasures to prevent loss of bone mineral during long duration space flight remain elusive. Despite an exercise program on MIR flights, the data from LeBlanc et al. (1996) indicated that there was still a mean rate of loss of bone mineral density in the proximal femur of 1.58% per month (n=18, flight duration 4-14.4 months). The specific mechanisms regulating bone mass are not known, but most investigators agree that bone maintenance is largely dependent upon mechanical demand and the resultant local bone strains (Frost, 1986; Jaworski and Uhthoff, 1986; Rubin and Lanyon, 1985; Jee and Li, 1990). A plausible hypothesis is that bone loss during space flight, such as that reported by LeBlanc et al. (1996), may result from failure to effectively load the skeleton in order to generate localized bone strains of sufficient magnitude to prevent disuse osteoporosis.

A variety of methods have been proposed to simulate locomotor exercise in reduced gravity (Davis et al. 1993). In such simulations, and in an actual microgravity environment, a gravity replacement load (GRL) must always be added to return the exercising subject to the support surface and the resulting skeletal load is critically dependent upon the magnitude of the GRL. To our knowledge, GRLs during orbital flight have only been measured once (on STS 81) and it is likely that most or all prior treadmill exercise in space has used GRLs that were less than one body weight. McCrory (1997) has shown that subjects walking and running in simulated zero-G can tolerate GRLs of 1 if an appropriate harness is used.

Several investigators have attempted to measure *in vivo* strains and forces in the bones of humans, but their efforts have been limited by technical difficulties and ethical considerations. Lanyon et al. (1975) applied a strain gauge to the anteromedial aspect of the tibial midshaft to measure strain during walking, both on a treadmill and overground. Strain gauges were mounted at the same location to study the effects of varied footwear (Milgrom et al., 1996) and vigorous activity (Burr et al., 1996) upon tibial strains. One reason to measure strains in the anterior tibia is that this region is more surgically accessible than other sites on the human lower extremity (Lanyon et al., 1975). Aamodt et al. (1997) were able to measure strains on the lateral surface of the proximal femur only because their experimental subjects were already scheduled for hip surgery. Lu et al. (1997) used an instrumented massive proximal femoral prosthesis along with electromyographic measurements to demonstrate that the axial forces carried by the femur strongly depend on muscular activity. These analyses of *in vivo* bone mechanics are valuable. However, their results may be confounded by the study of subjects with pathological conditions and the invasive nature of the procedure prevents the use of large numbers of subjects and multiple strain gauge locations. Gross et al. (1992) measured strain at three locations on the equine third metacarpal and used those data to construct a computer model of the internal strain environment of the bone. An analogous placement of multiple gauges in living humans would be difficult and potentially hazardous because of the depth of soft tissue overlying the tibia and femur.

The purpose of the present study was to present a method to measure external ground reaction forces in human subjects during 1G and simulated zero-G exercise and to measure internal bone strains under similar conditions using a unique cadaver model.

METHODS

The Penn State zero-G locomotion simulator used in this study (Figure 1) has been previously described by Davis et al. (1996). Briefly, the limbs of the subject are suspended from long elastic cords that act as quasi-constant force springs to offset the weight of the suspended segments. GRLs are applied through waist and shoulder harnesses, and ground reaction forces are measured by a force platform located under the belt of a vertically mounted treadmill. Subjects also performed the same exercises during level 1G locomotion over a Kistler force platform.

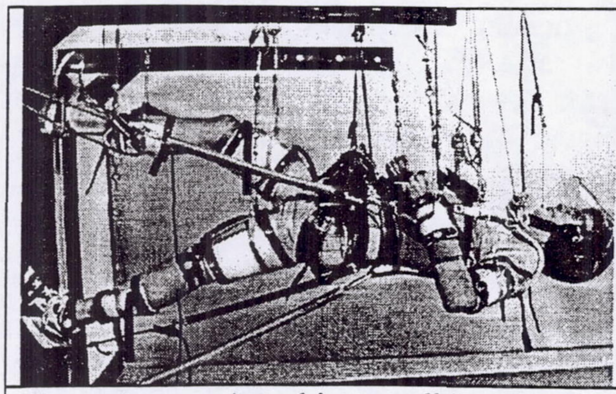


Figure 1: A subject walking in the zero-Gravity Locomotion simulator which has a vertically mounted force measuring treadmill.

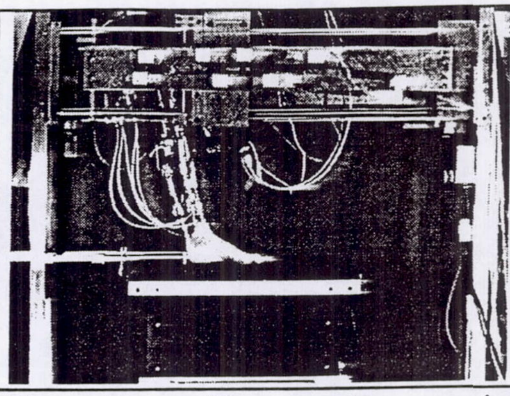


Figure 2: The cadaver simulator showing the muscle forces being applied as the limb moves through the appropriate kinematic trajectory.

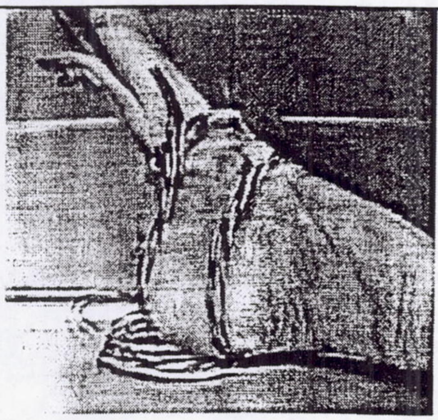


Figure 3: Strain gauges bonded to the distal tibia of a cadaver limb. The cut ends of the Achilles tendon are clamped to a force actuator.

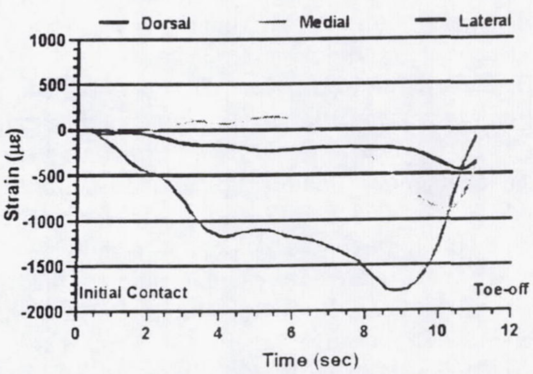


Figure 4: Metatarsal strains measured during simulated 1-G walking in 1/20th real time speed.

Each cadaver foot was mounted into an apparatus described by Sharkey and Hamel (1998) that dynamically loads the limb preparations and mimics the kinetics and kinematics of the tibia, foot, and ankle during the stance phase of gait from heel-strike to toe-off. Physiologic actions of the extrinsic foot muscles are simulated using force feedback controlled linear actuators interfaced with the tendons of the specimen. Previous studies have shown that the ground reaction forces and plantar pressures under the specimens are well within physiologic ranges and demonstrate inter-subject variation consistent with that seen in live subjects. The timing and force output of six separate muscle groups (constituting the major dorsi- and plantar flexors of the foot) were independently controlled with stepper motor powered linear actuators. Each muscle force profile was shape-matched to its corresponding mean EMG activity profile. Stacked epoxy-phenolic

encapsulated rosette strain gauges (Micro-Measurements Group, Inc. WA-06-060WR-120) were attached at four transversely co-planar locations on the distal tibia of the cadaver limbs using M-Bond 200 cyanoacrylate adhesive and Catalyst-C. Specimen surfaces were prepared in the following manner: a small region of the periosteum was removed by lightly sanding the region which was then cleaned and degreased with an acetone swab. The gauges, leadwires, and strain relief terminals were sealed with silicon rubber.

RESULTS

Ground reaction forces in the human subjects ambulating under zero-G conditions were significantly lower than their 1G equivalents (active force peaks 87.8 %BW vs. 124.4 BW % during walking, 180% BW vs 240% BW during running). This was apparently due to a bent-knee or "Groucho" running style which subjects used to minimize the applied load. Metatarsal strains measured in cadaver feet during simulated 1G walking were within the physiological ranges that have been previously reported (Figure 4). Data on tibial strains will also be presented.

CONCLUSION

The human zero-gravity locomotion simulator and the cadaver simulator offer a powerful combination for the study of the implications of exercise for maintaining bone quality during space flight. Such studies, when compared with controlled in-flight exercise programs, could help in the identification of a strain threshold for the prevention of bone loss during space flight.

This work was supported in part by NASA grant NAGW-4421

REFERENCES:

Aamodt, A., Lund-Larsen, J., Eine, J., Andersen, E., Benum, P., and Husby, O. S., 1997, *In vivo* measurements show tensile axial strain in the proximal lateral aspect of the human femur. Journal of Orthopaedic Research, 15: 927-31.

Burr, D. B., Milgrom, C., Fyhrie, D., Forwood, M., Nyska, M., Finestone, A., Hoshaw, S., Saiag, E., and Simkin, A., 1996, *In vivo* measurement of human tibial strains during vigorous activity, Bone, 18: 405-10.

Davis, B. L., Cavanagh, P. R., 1993, Simulating reduced gravity: a review of biomechanical issues pertaining to human locomotion. Aviation, Space, and Environmental Medicine. 64(6): 557-566.

Davis, B. L., Cavanagh, P. R., Sommer, H. J., III, Wu, G., 1996, Ground reaction forces during locomotion in simulated microgravity. Aviation, Space and Environmental Medicine. 67(3): 235-242.

Frost H.M., 1986, Intermediary Organization of the Skeleton. CRC Press, Boca Raton.

Gross, T. S., McLeod, K. J., and Rubin, C. T., 1992, Characterizing bone strain distributions *in vivo* using three triple rosette strain gauges, Journal of Biomechanics, 25: 1081-7.

Jaworski Z.F.G.,and Uhthoff H.K., 1986, Reversibility of nontraumatic disuse osteoporosis during its active phase. Bone. 7: 431-439.

Jee W.S.S., and Li X.J., 1990, Adaptation of cancellous bone to overloading in the adult rat: a single photon absorptiometry and histomorphometric study. Anatomical Record. 227: 418-426.

Lanyon, L. E., Hampson, W. G., Goodship, A. E., and Shah, J. S., 1975, Bone deformation recorded *in vivo* from strain gauges attached to the human tibial shaft. Acta Orthopaedica Scandinavica, 46: 256-68.

LeBlanc A; Schneider V; Shackelford LL; West S; Oganov V; Bakulin A; Voronin L., 1996, Bone mineral and lean tissue loss after long duration spaceflight. J. Bone Mineral Research, 11: S323.

Lu, T. W., Taylor, S. J., O'Connor, J. J., and Walker, P. S., 1997, Influence of muscle activity on the forces in the femur: an *in vivo* study, Journal of Biomechanics, 30: 1101-6.

Milgrom, C., Burr, D., Fyhrie, D., Forwood, M., Finestone, A., Nyska, M., Giladi, M., Liebergall, M., and Simkin, A., 1996, The effect of shoe gear on human tibial strains recorded during dynamic loading: a pilot study, Foot & Ankle International, 17: 667-71.

McCrory J.L., 1997, A biomechanical comparison of 1-G and fully loaded simulated Zero-gravity locomotion. Ph.D. Thesis. Penn State University.

Rubin C.T., and Lanyon LE., 1985, Regulation of bone mass by mechanical strain magnitude. Calcified Tissue International. 37: 411-417

Sharkey N.A., and Hamel, A.J., 1998, A dynamic cadaver model of the stance phase of gait: performance characteristics and kinetic validation. Clin. Biomech. 13: 420-433.

ISB International Society of Biomechanics

This is the second volume in the International Society Biomechanics Series, published in association with The International Society of Biomechanics, University of Calgary, Canada.

Three-dimensional Analysis of Human Locomotion

Edited by

PAUL ALLARD
Laboratoire d'Étude du Mouvement, Research Center, Sainte-Justine Hospital, Montreal, and University of Montreal, Canada

AURELIO CAPPOZZO
Cattedra di Tecnologie Biomediche, Università degli Studi, Sassari, Italy

ARNE LUNDBERG
Department of Orthopaedic Surgery, Huddinge University Hospital, Sweden

and

CHRISTOPHER L. VAUGHAN
Department of Biomedical Engineering, University of Cape Town, South Africa

JOHN WILEY & SONS
Chichester · New York · Weinheim · Brisbane · Singapore · Toronto

Baffins Lane, Chichester,
West Sussex PO19 1UD, England

National 01243 779777
International (+44) 1243 779777
e-mail (for orders and customer service enquiries): cs-books@wiley.co.uk
Visit our Home Page on http://www.wiley.co.uk
or http://www.wiley.com

Published in association with The International Society of Biomechanics (ISB), University of Calgary, Canada.

Other Wiley Editorial Offices

John Wiley & Sons, Inc., 605 Third Avenue,
New York, NY 10158-0012, USA

WILEY-VCH Verlag GmbH, Pappelallee 3,
D-69469 Weinheim, Germany

Jacaranda Wiley Ltd, 33 Park Road, Milton,
Queensland 4064, Australia

John Wiley & Sons (Asia) Pte Ltd, 2 Clementi Loop #02-01,
Jin Xing Distripark, Singapore 129809

John Wiley & Sons (Canada) Ltd, 22 Worcester Road,
Rexdale, Ontario M9W 1L1, Canada

Library of Congress Cataloging-in-Publication Data

Three-dimensional analysis of human locomotion / edited by Paul Allard [et al.]
p. cm.
Includes bibliographical references and index.
ISBN 0-471-96949-4 (alk. paper)
1. Human locomotion. 2. Three-dimensional imaging in biology.
I. Allard, Paul, 1952-
[DNLM: 1. Locomotion—physiology. 2. Gait—physiology.
3. Computer Simulation. 4. Biomechanics. WE 103 T531 1997]
QP303.T585 1997
612.7'6—dc21
DNLM/DLC
for Library of Congress
97-11505
CIP

British Library Cataloguing in Publication Data

A catalogue record for this book is available from the British Library

ISBN 0 471 96949 4

Typeset in 10/12pt Times by Keytec Typesetting Ltd, Bridport, Dorset.
Printed and bound in Great Britain by Bookcraft (Bath) Ltd
This book is printed on acid-free paper responsibly manufactured from sustainable forestation, for which at least two trees are planted for each one used for paper production.

Contents

18

The Kinematics of Treadmill Locomotion in Space

W. E. THORNTON[1], P. R. CAVANAGH[2], F. L. BUCZEK[3], M. J. BURGESS-MILLIRON[4] AND B. L. DAVIS[5]

[1]701 Coward's Creek Road, Friendswood, Texas, USA

[2]The Center for Locomotion Studies, Pennsylvania State University, University Park, PA, USA

[3]Motion Analysis Laboratory, Shriner's Hospital for Children, 1645 West 8th Street, Erie, PA, USA

[4]Converse Inc., North Reading, MA, USA

and

[5]Department of Biomedical Engineering, Lerner Research Institute, The Cleveland Clinic Foundation, Cleveland, OH, USA

INTRODUCTION

Human adaptation to microgravity includes decreased cardiorespiratory capacity, neuromuscular changes, muscle atrophy, and bone demineralization. Such adaptations may interfere with or even preclude many normal functions on return to the earth or other planets with significant gravity. Effects on landing range from transient discomfort and disability following missions of a few days, to inability to walk and significant losses of bone mineral (particularly in the lower limbs) following flights that last months (Figure 1).

It is now well accepted that exercise countermeasures must play a major role in preventing a number of adverse sequelae of prolonged exposure to microgravity (Thornton, 1990). Different forms of exercise have been employed,

Three-dimensional Analysis of Human Locomotion. Edited by P. Allard, A. Cappozzo, A. Lundberg and C. Vaughan

ISBN 0 471 96949 4

Figure 1. On returning from an extended mission, a traveller may experience both temporary and chronic symptoms. (© King Features Syndicate. Reprinted with special permission of King Features Syndicate.)

ranging from walking on a slippery sheet of teflon, to tethered locomotion on a passive treadmill (Moore, 1990; Thornton and Rummel, 1977). US astronauts have not yet exercised on a motorized treadmill in 0 G, although Russian cosmonauts use a treadmill that has both passive and active modes, and the tension in the bungee tethers is nominally set at about 500 N. This device remains the centrepiece of their countermeasures for long-duration flights, including those that exceed 1 year.

In this study, we have used film from two shuttle missions to answer some initial questions concerning the kinematics of lower extremity joints and body segments during tethered treadmill running in space. The primary aim was to contrast passive treadmill running in 0 G with uphill running on a passive treadmill in 1 G. A secondary aim was to compare the kinematics of tethered running on a passive treadmill in 0 G and published data on uphill running on a motorized treadmill in 1 G (Milliron and Cavanagh, 1990).

METHODS

EXERCISE TREADMILL

The treadmill used for many of the shuttle flights, including STS-7 and STS-8 (the missions on which data that will be described in this chapter were collected), was designed with a 0.7-m running surface (Figure 2). The track on which the astronauts walked or ran consisted of a series of transverse aluminium plates linked together. Treadmill resistance was minimized by mounting the plates on precision ballbearings that travelled in a machined guide. Traction between an astronaut's feet and the track was possible because of the tethered arrangement in which four bungee cords were attached to a harness worn by the astronaut. These cords were adjustable and were connected to padded straps in the hip and

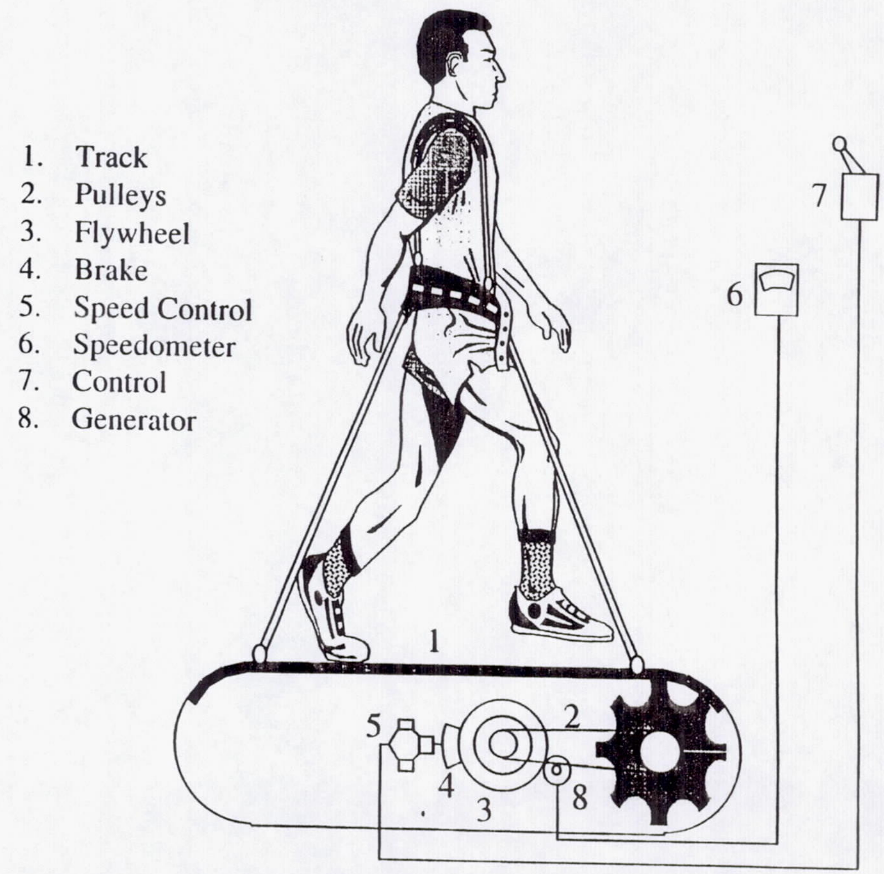


Figure 2. Passive treadmill used on NASA mission STS-7 and STS-8

shoulder regions in such a manner that the shoulder loads were approximately 35–40% of the total axial load. (Although the total tension was not measured, it was expected that loads equivalent to body weight would be generated by the tethering system.) To minimize fluctuations in the bungee tensions (ΔF) during the periods of exercise, the bungee cords were designed such that their length (L) was large compared to the oscillations (ΔL) of the upper body (i.e. since $\Delta F = K\Delta L/L$ (where K is a stiffness constant), ΔF can be minimized by increasing L).

TRIAL CONDITIONS

The primary material used in this study was 16-mm film of tethered treadmill running taken during the STS-7 and STS-8 missions (Figure 3). The camera had a nominal rate of 24 Hz (in 1 G) and a 5-mm focal length lens. Operational conditions (in flight) required that the camera be hand-held at a location approximately 1 m from the side of the treadmill. In an attempt to more fully understand the relationship between (1) 0 G and 1 G gait and (2) passive and motorized treadmill running, two other trial conditions were investigated. In the first of these, a single astronaut who was one of the four subjects for whom film in 0 G was available walked and ran in 1 G on a treadmill similar to one used on the shuttle. For these data, the same camera was used and was placed on a tripod 1.12 m from the treadmill. The orientation of this treadmill in 1 G was such that

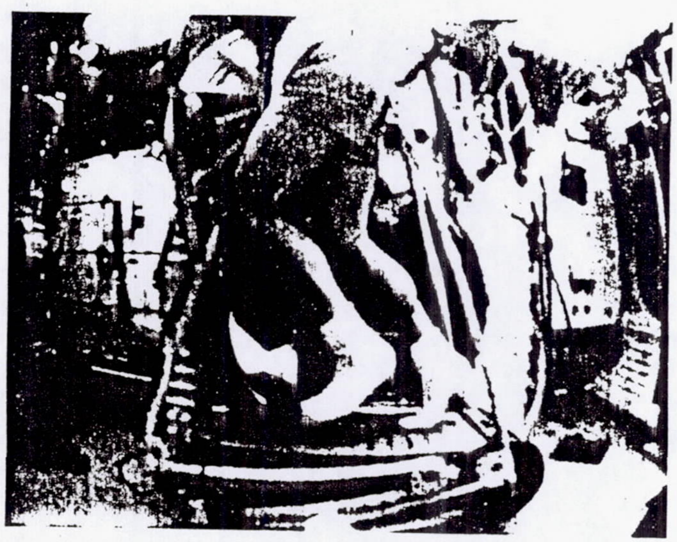


Figure 3. A single frame taken from the video recordings of astronauts exercising on the shuttle treadmill. Note the distortion caused by the camera lens. This distortion was removed by using methods described in Woltring (1980)

the subject exercised at a positive (uphill) inclination of 16.3% at speeds that were governor-controlled and set at either 1.4, 1.9 or 2.3 m/s. The second data set used for comparison was from a group of distance runners running at a fixed speed (3.4 m/s) on a motorized treadmill (Quinton) at both a level and uphill inclination of +20% (Milliron and Cavanagh, 1990).

COORDINATE DATA

All film from the shuttle treadmill (both 0 G and 1 G) was collected using a 5 mm lens, due to the restricted dimensions of the cabin. This resulted in considerable distortion in the final image. A solution to this problem was obtained by using the method described by Woltring (1980) to obtain a transformation between image space and object space.

Before the 1 G shuttle treadmill filming was conducted, a grid of known dimensions was exposed using the same camera and lens and subsequently analysed. This allowed the distortion parameters to be derived as well as the scaling factor for converting image-space coordinates into object-space coordinates. No such direct measures were available for the 0 G films, but since the camera lens was the same as for the 1 G situation, the same distortion parameters were used for both the 1 G and 0 G cases. A separate scaling factor was used for the 0 G films and was obtained using the known dimensions of the treadmill's track length, and since this was visible in all the film trials, separate scaling factors could be applied to each film sequence (since the camera position was not fixed).

Six body landmarks were used to identify segments in the films: (1) superior border of the greater trochanter, (2) lateral femoral epicondyle, (3) lateral malleolus, (4) posterior aspect of the heel, (5) head of the fifth metatarsal, and (6) most distal point of the second toe. Since no surface markers were placed on the subjects, positions of the landmarks were estimated. This was most difficult for the fourth marker in the 1 G trials, since the shoes that were worn were not as clearly demarcated as the socks or slippers that were worn in the 0 G trials. Marker 1 was also sometimes difficult to identify, due to the hip region being covered by relatively 'baggy' shorts.

All of the data obtained from the shuttle treadmill were digitally filtered with a cut-off frequency of 4.5 Hz. This frequency was the highest allowable, given the fact that the film data were collected at a rate of only 24 Hz (Winter, Sidwall and Hobson, 1974).

COORDINATE AXES

Since the primary purpose of this study was to relate 0 G treadmill locomotion to that in 1 G, a consistent axis system was required. In orbit, 'horizontal' and 'vertical' have little meaning, and for this reason a reference system relative to

the treadmill bed was employed. In this treadmill axis system (TAS), the *X*-axis was parallel to the running surface, with the *Y*-axis being orthogonal to it. To allow comparisons with 0 G kinematics, the 1 G data were also transformed to this reference frame. It is emphasized that this transformation means that the 1 G data no longer reflect a gravitational reference system. In the case of the trials conducted on the shuttle treadmill in a 1 G environment at an incline of 16.3%, the gravitational force vector actually acted at an angle of about 9° to the *Y*-axis.

LIMB SEGMENT AND JOINT ANGLES

Due to the physical set-up of the treadmill in the space shuttle, only the left side of the astronauts was filmed. The lower extremity segment angles were considered raw data, used primarily in combination to provide desired joint angles. A segment angle was defined as the counter-clockwise angle it made with the positive *X*-axis, when the segment was rotated about its distal endpoint. Joint angles (hip, knee and ankle) were defined according to the convention shown in Figure 4.

The customary method for performing kinematic analyses is to make use of an initial 'standing calibration'. This involves measuring the locations of markers on a subject's lower extremity when a limb is in a neutral (upright) position. The advantage of this approach is that by relating subsequent marker positions to the initial location, it is a simple matter to determine whether the joint angle has

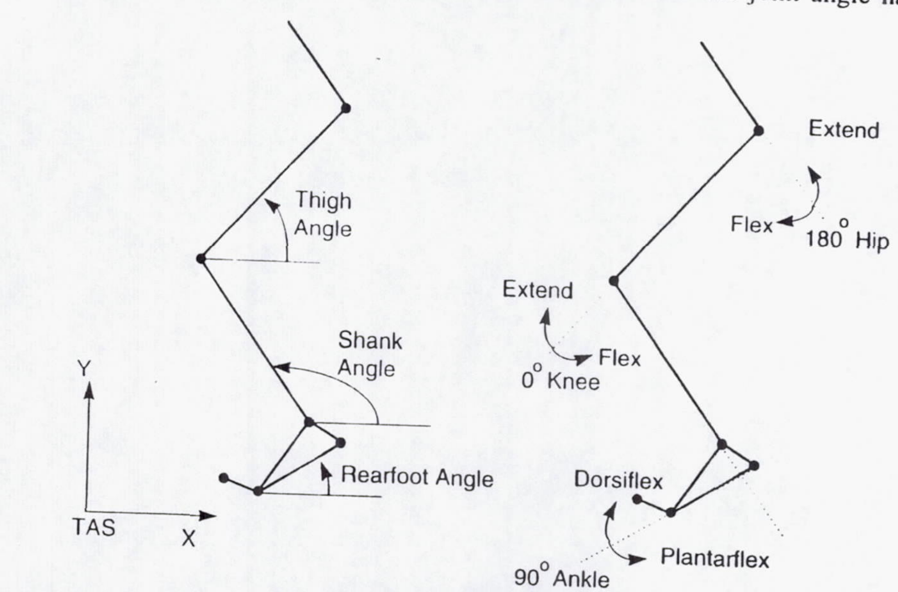


Figure 4. Conventions for determining segment and joint angles. Joint angles of 100° correspond to a flexed hip, a flexed knee and a dorsiflexed ankle

remained constant or whether flexion–extension has occurred. Without the 'standing calibration', it might appear as though the joint is flexed, whereas this may just be a result of marker location rather than anatomical factors. In the present case, since no film was available for any of the astronauts in a standing position, a point in midstance of walking was chosen that came closest to the orientation of the foot and ankle in standing. Segment lines were drawn between the 5th metatarsal head and the heel, and an angle was determined that adjusted the calculated foot segment to zero during the time that it was flat on the treadmill bed.

In a similar way, an offset angle was calculated for the trunk. If a line is drawn upwards from the hip marker, through the centre of the partially visible upper body, the angular deviation of this line from the *Y*-axis constitutes the trunk offset angle. This procedure was performed for each frame and it allowed the calculation of the hip angle, rather than just the orientation of the thigh segment relative to the *Y*-axis. For the 1 G standard treadmill data, standing calibrations were available. Correction angles were specified for the thigh, shank and rearfoot segments. The trunk offset angle was taken from the literature (Yoneda et al., 1979). The conventions used to determine joint motion (after applying the correction angles) are shown in Figure 4. All of these assumptions are typical of the compromises needed to analyse data collected under operational conditions where the rigour that can be imposed in a laboratory setting is not possible.

RESULTS

Before presenting the quantitative kinematic data, it is appropriate to present some observations based on a subjective view of the film and known biomechanical conditions. A passive treadmill must be 'driven' by the subject generating a force along the tread that must overcome the treadmill track's frictional resistance. On earth this is most easily done by elevating the treadmill, thereby producing a backward horizontal force that is a function of both the elevation angle and the subject's weight. In weightlessness, this force is a function of the subject's angle with respect to the tread and the bungee load. Also, as speed increases, so do the frictional losses and there is a concomitant increase in the degree of forward lean. Although the precise inclination of the upper body could not be determined in the film analysed in this study (due to most of the trunk being out of the view of the camera), it was apparent that the astronauts did exhibit marked forward lean which increased with speed. This style of locomotion was associated with a footstrike pattern that was characterized by an extreme forefoot strike—as opposed to the rearfoot strike seen in typical heel–toe running. It was evident that the handle on the treadmill was important for stability during gait, especially for the faster runs. The relationship between forward lean, support of the handle and footstrike pattern will be discussed later.

An initial, unexpected, result was that there was a discrepancy between the treadmill track speed as measured using the video camera and the speed (in 1 G) as set with a governor-controller. An explanation for this discrepancy could be that the camera's frame rate was higher than the nominal rate (24 frames/s). The camera's specifications stipulate that the frame rate can have an uncertainty of ±10%. If the upper limit of this range is used, then the three settings for the treadmill in 1 G (1.4, 1.9 and 2.3 m/s) are in better agreement with the speeds of 1.38, 1.86 and 2.07 m/s listed in the right-hand column in Table 1. For the remainder of this chapter it will thus be assumed that the camera had a frame rate of 26.4 frames/s rather than the nominal 24 frames/s. However, for completeness, both the nominal and 'fast camera' data are presented in Table 1.

The results describing the stride parameters for 0 G and 1 G conditions are given in Table 1. The most notable factor in the 0 G shuttle treadmill data was the low speed for both walking (0.75–1.04 m/s) and running (1.05–1.61 m/s). Even the trials that were designated as fast runs were only at speeds between 1.93 and 2.09 m/s. During locomotion on a similar treadmill in 1 G, the walking speeds were 1.38 and 1.86 m/s and the running speeds were 1.86 and 2.07 m/s. For comparison, typical distance running speeds (Cavanagh and Kram, 1990) are

Table 1. Stride parameters for locomotion on a passive treadmill in 0 G and 1 G

Activity	Stride length (m)	Stride length (stat)	Nominal speed		Speed based on a faster camera (m/s)	Stride frequency based on a faster camera (Hz)
			m/s	stat/s		
Walk (0 G)	0.83	0.47	0.69	0.39	0.75	0.90
	1.22	0.67	0.93	0.51	1.02	0.84
	1.02	0.56	0.88	0.48	0.97	0.95
	1.14	0.64	0.95	0.53	1.04	0.91
Run (0 G)	0.96	0.55	0.96	0.55	1.05	1.09
	0.98	0.52	1.06	0.57	1.16	1.18
	1.14	0.64	1.3	0.73	1.43	1.25
	1.15	0.63	1.46	0.8	1.16	1.40
	1.16	0.66	1.38	0.79	1.52	1.31
Fast run (0 G)	1.14	0.64	1.76	0.99	1.93	1.69
	1.31	0.72	1.8	0.98	1.98	1.51
	1.27	0.69	1.9	1.03	2.09	1.65
Walk (1 G)	1.15	0.63	1.25	0.68	1.38	1.20
	1.37	0.74	1.69	0.91	1.86	1.36
Run (1 G)	1.25	0.67	1.69	0.91	1.86	1.49
	1.35	0.73	1.88	1.02	2.07	1.53

The 12 rows in 0 G correspond to four astronauts walking or running at three different speeds. The trials in 1 G correspond to a single subject walking and running at two speeds

between 3 and 4 m/s, while speeds of sprinting extend from 6 to 10 m/s. Thus the speeds of locomotion on the shuttle treadmill in 0 G were about 60% of those found using the same device in 1 G, while the latter were, in turn, about 60% of those typically found during overground distance running. Two factors that relate to velocity of running are stride frequency and stride length. Of these, it was found that stride length was less on the shuttle treadmill (almost certainly due to the small length of the running surface), whereas stride frequencies during running on the shuttle treadmill in 0 G (mean = 1.25 Hz) and in 1 G (mean = 1.51 Hz) were comparable to those reported for overground running in 1 G (range 1.38–1.44 Hz) reported by Cavanagh and Kram (1990).

The relationships between hip, knee and ankle motions during fast running in 0 G (mean velocity = 2 m/s) and running in 1 G (velocity = 1.97 m/s) are shown in Figures 5 to 8. The use of these kinds of angle–angle diagrams has been well described in the literature (Lamoreux, 1971). The advantage of these plots is that they are presented as a shape which is extremely sensitive to changes in an individual's style of locomotion (Milliron and Cavanagh, 1990), and they are not sensitive to errors in a camera's frame rate.

DISCUSSION

It has previously been shown that the kinematics of both running and walking gait are dependent on speed (Andriacchi, Ogle and Galante, 1972; Lamoreux, 1971; Sinning and Forsyth, 1970). In addition, it has been shown that running in simulated microgravity on a passive treadmill differs from running on a

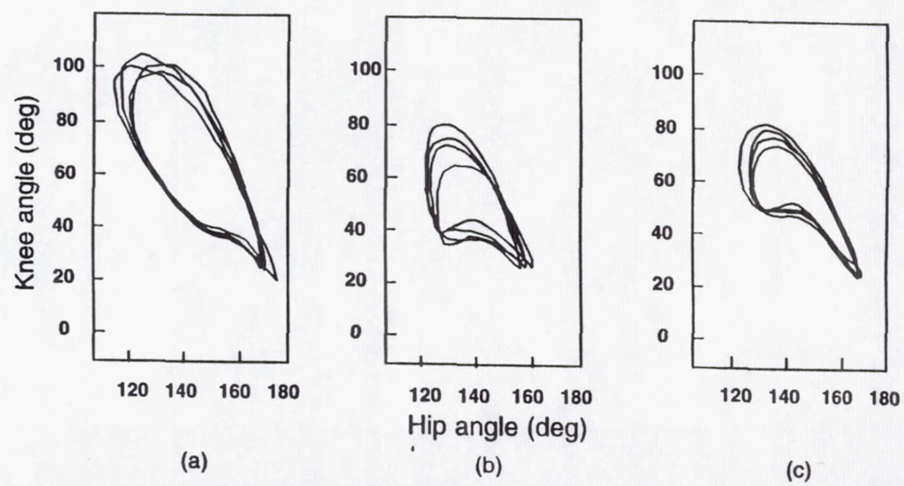


Figure 5. Hip–knee angles during fast running on shuttle treadmill in 0 G. Each panel refers to a different astronaut (a = 1.93 m/s, b = 1.98 m/s, c = 2.09 m/s)

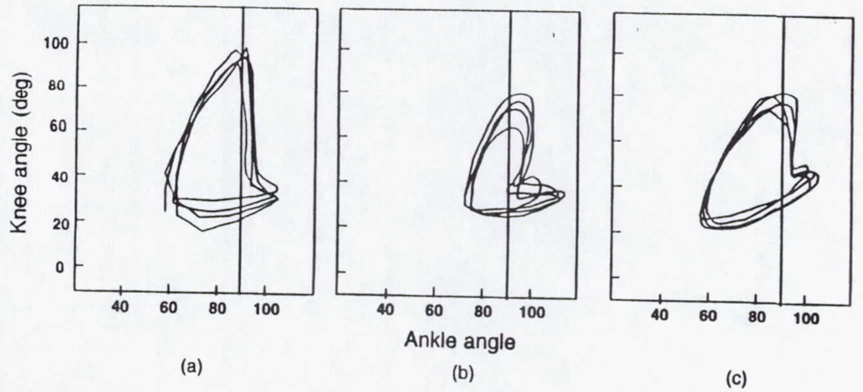


Figure 6. Ankle-knee angles during fast running on shuttle treadmill in 0 G. Each panel refers to a different astronaut (a = 1.93 m/s, b = 1.98 m/s, c = 2.09 m/s)

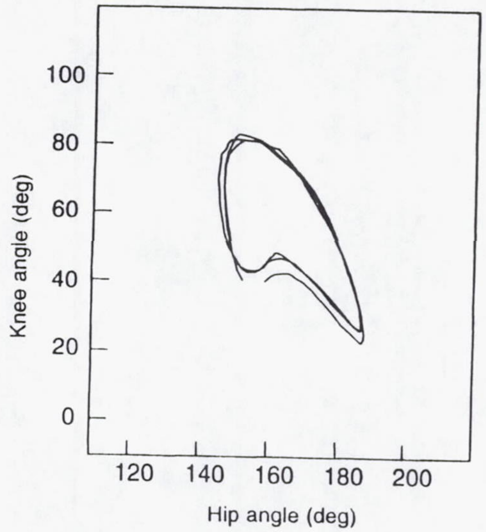


Figure 7. Hip–knee angles during running on shuttle treadmill in 1 G. (Single subject: grade = +16.3%, velocity = 1.86 m/s). Note the difference in hip angles compared to Figure 5c

motorized treadmill (Davis et al., 1996). These issues need to be taken into account when comparisons are made between gait on the shuttle treadmill in 0 G and 1 G and when trials in 0 G are compared to running on a standard motorized treadmill at various inclinations in 1 G (Milliron and Cavanagh, 1990). It should be emphasized that running in 0 G resulted in velocities that were about 60% of those achieved in 1 G. For this reason, emphasis will be placed on comparisons

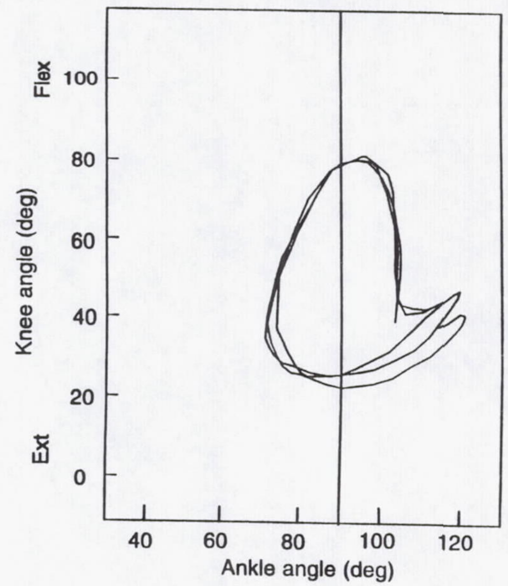


Figure 8. Ankle–knee angles during running on shuttle treadmill in 1 G. (Single subject: grade = +16.3%, velocity = 1.86 m/s)

between the fastest trials in 0 G and the slowest running trials in 1 G. The discrepancy between 0 G and typical motorized treadmill locomotion was even greater—the highest speed of 2.09 m/s in the former is markedly less than speeds reported in the literature (Milliron and Cavanagh, 1990). However, it would not have been practical to expect subjects to run on a motorized treadmill in 1 G at a velocity of about 2 m/s, since this speed would result in a very unnatural running gait. Likewise, it seems that the environmental and physical constraints that affect locomotion in 0 G preclude astronauts from running at speeds between 3 and 4 m/s.

Figure 5 shows the data of 0 G fast running. Apart from the increased knee flexion during the swing phase for one subject, there were general similarities between all the trials—for instance, it is apparent that neither the knee nor the hip joints ever became fully extended. This style of locomotion was probably a result of a variety of factors, e.g. the tension in the bungee cords that restrained the astronauts to the treadmill, the limited length of the running surface (Davis and Cavanagh, 1993), the passive nature of the treadmill, and the fact that there was a handle that offered support to the upper body. In leaning forwards, the astronauts were not only better able to grasp the handle, but, in addition, there was greater stretch in the posterior cords, which undoubtedly produced forces that made it easier to drive the treadmill track.

While horizontally directed forces may be what an astronaut focuses on as he or she exercises on a passive treadmill, what is probably of most importance to the musculoskeletal system is the degree of axial loading. While this has never been quantified during an actual NASA mission, indirect measurements using accelerometers mounted in the spacecraft structure (Dunbar, Giesecke and Thomas, 1991) suggest that treadmill exercise results in loads being applied to the lower limbs that far exceed the loads encountered during any of the other activities of daily living in microgravity. Since the accelerations that result from treadmill exercise are of concern to scientists who utilize the unique environment of 0 G for conducting their experiments (e.g. researchers who study crystal growth in weightlessness), considerable attention has recently been given to isolating treadmill vibrations from the shuttle superstructure. The emphasis in these efforts has been to design a device that transmits impact loads to the shuttle below NASA specifications and yet allows 1 G equivalent loading on the human body. The latter factor is considered important in light of work showing that impulsive loading to the lower extremity can stimulate bone formation as high as the cervical vertebra (Burr, Martin and Martin, 1983).

By selecting one trial for a single astronaut, it is possible to compare 0 and 1 G conditions at similar speeds (2.09 and 1.86 m/s respectively). Figures 5c and 7 give these results for hip–knee coordination, while Figures 6c and 8 relate to the motion at the knee and ankle. The hip–knee diagram shows that the knee joint has very similar patterns of motion, whereas the hip angle differs considerably with regard to the region where motion occurred. In 0 G the hip had a minimum flexion of 160°, while the 1 G data show a value close to 190°. The maximum flexion in the 1 G condition was about 120°, while in 0 G it was only 145°. Thus although the ranges of motion were comparable, the movement occurred in different parts of the available range. This was not surprising, since subjective analysis showed the subject to be leaning far more during the 0 G trials. With regard to the ankle–knee diagrams, it is apparent that the ankle was positioned in plantarflexion for most of the cycle (Figure 6). This was typical for most of the trials in 0 G, with only the data for one subject (Figure 6b) displaying comparable amounts of dorsiflexion and plantarflexion. The reason for this pattern was once again due to the marked forward lean which resulted in subjects running on their toes. Running on the passive treadmill in 1 G (Figure 8) produced similar kinematics to those shown in Figure 6, with the main difference being that there was more dorsiflexion of the ankle (i.e. maximum of 120°).

In the case of running on a motorized treadmill in 1 G, hip–knee and ankle–knee diagrams (Milliron and Cavanagh, 1990) allow one to ascertain that a +20% grade produces kinematics that most closely resemble the fast-run 0 G trials. The similarity in the support phase is considered most important, where, in both cases, the knee was somewhat flexed at footstrike and there was only a small amount of knee flexion after footstrike before the 'extensor thrust' phase of coordinated knee extension and hip extension began. In the case of ankle–

knee diagrams (Figures 5 and 8), the 20% uphill condition again provided the closest match.

CONCLUSIONS

Locomotion on a treadmill in 0 G will probably remain a centrepiece of NASA's exercise countermeasures programme. This form of physical activity has the potential to cause large bone and muscle forces as well as substantial metabolic loading during a period of continuous treadmill exercise. A critical concern is the provision of a treadmill which can approximate 1 G performance in space. At this point, no adequate objective measurements of in-flight treadmill kinetics or of the human response to this activity have been made.

Interpretation of the results obtained in the present study is limited by the following: (1) bungee tensions were not measured; (2) ground reaction forces were not measured in parallel with the kinematic measurements; and (3) the instrumentation used to film the astronauts could itself have been affected by microgravity. Despite these shortcomings, what is apparent is that exercise during NASA missions STS 7 and STS 8 resulted in leg motions that were similar to those found during 1 G locomotion on an inclined passive treadmill and on an active treadmill at an even steeper grade. In addition, it was apparent that the majority of the loads were transmitted through the forefoot, and one can surmise that this style of running would result in physiologically significant tensions in the calf musculature and resultant ankle compressive loading. Further speculation regarding limb loading is complicated by the fact that varying amounts of force are transmitted through (1) the treadmill handle and (2) bungee cords that act as a tether.

New generations of treadmills are being manufactured that could provide important information for planners of long-duration space missions. If these types of treadmill are flown on future missions, it will be possible to control bungee tensions more precisely, control for grade and speed, and, most importantly, provide data on the rates and magnitudes of limb loading. These data could then be incoporated into biomechanical models of the lower limb to more fully understand mechanisms of load transmission from distal to proximal structures and to optimize in-flight exercise protocols in such a way that muscle and bone loss could be reduced.

ACKNOWLEDGEMENT

Supported in part by a grant to the Penn State Center for Locomotion Studies from Krug International.

REFERENCES

Andriacchi, T. P., Ogle, J. A. and Galante, J. O. (1972) Walking speed as a basis for normal and abnormal gait measurements. *J. Biomech.*, **10**, 261–268.

Burr, D. B., Martin, R. B. and Martin, P. A. (1983) Lower extremity loads stimulate bone formation in the vertebral column: implications for osteoporosis. *Spine*, **8**(7), 6781–6786.

Cavanagh, P. R. and Kram, R. (1990) Stride length in distance running: velocity, body dimensions, and added mass effects. In: *Biomechanics of Distance Running* (ed. P. R. Cavanagh), Human Kinetics, Champaign, IL, pp. 35–63.

Davis, B. L. and Cavanagh, P. R. (1993) Simulating reduced gravity: a review of biomechanical issues pertaining to human locomotion. *Aviat. Space Environ. Med.*, **64**, 557–566.

Davis, B. L., Cavanagh, P. R., Sommer, H. J. III and Wu, G. (1996) Ground reaction forces during locomotion in simulated microgravity. *Aviat. Space Environ. Med.*, **67**(3), 235–242.

Dunbar, B. J., Giesecke, R. L. and Thomas, D. A. (1991) The microgravity environment of the space shuttle Columbia payload bay during STS 32. *NASA Technical Paper* 3141, NASA, Washington DC.

Lamoreux, L. W. (1971) Kinematic measurements in the study of human walking. *Bull. Prosthet. Res.* **10**(15), 3–84.

Milliron, M. J. and Cavanagh, P. R. (1990) Sagittal plane kinematics of the lower extremity during distance running. In: *Biomechanics of Distance Running* (ed. P. R. Cavanagh), Human Kinetics, Champaign, IL, pp. 65–105.

Moore, T. P. (1990) The history of in-flight exercise in the US manned space program. In: *Workshop on Exercise Prescription for Long-duration Space Flight* (ed. B. A. Harris Jr. and D. F. Stewart), NASA Conference Proceedings 3051, NASA, Washington DC, pp. 19–21.

Sinning, W. E. and Forsyth, H. L. (1970) Lower-limb actions while running at different velocities. *Med. Sci. Sport Exer.*, **2**(1), 28–34.

Thornton, W. E. (1990) Work exercise and spaceflight: II Modification of adaptation by exercise (Exercise prescription). In: *Workshop on Exercise Prescription for Long-duration Space Flight* (ed. B. A. Harris Jr. and D. F. Stewart), NASA Conference Proceedings 3051, NASA, Washington DC, pp. 107–115.

Thornton, W. E. and Rummel, J. (1977) Muscular deconditioning and its prevention in space flight. In: *Biomedical Results from Skylab* (ed. R. S. Johnston and L. F. Dietlein), NASA SP377, NASA, Washington DC, pp. 191–197.

Winter, D. A., Sidwall, H. and Hobson, D. (1974) Measurement and reduction of noise in kinematics of locomotion. *J. Biomech.*, **7**, 157–159.

Woltring, H. (1980) Planar control in multi-camera calibration for 3-D gait studies. *J. Biomech.*, **13**, 39–48.

Yoneda, Y., Adrian, M., Walker, F. and Dobie, D. (1979) Kinematic and kinetic analysis of sprinting and jogging. In: *Science in Athletics* (ed. J. Terauds and G. Dales), Academic Publishers, Del Mar, CA pp. 85–91.

19

Gait Data: Reporting, Archiving and Sharing

ROY B. DAVIS[1], SYLVIA ÕUNPUU[2] AND PETER A. DELUCA[2]

[1]Newington Children's Hospital, Newington, Connecticut, USA

[2]Connecticut Children's Medical Center, Hartford, Connecticut, USA

INTRODUCTION

There are no universally accepted standards or conventions for the reporting, archiving and sharing of gait analysis data. 'Gait analysis' refers to a variety of approaches that seek to qualify and quantify human locomotion. These methods range from the qualitative observations made as the subject walks down the clinic hallway or laboratory pathway to the full three-dimensional (3D) quantitative analysis of the instrumented subject based on a collection of measured values. 'Gait data' refers to a myriad of parameters that results from these evaluations that quantify and/or qualify certain aspects of locomotion. The term 'data' as used here refers not only to quantities that are measured directly, but also to the values that are computed based on these measurements. For example, distance and time measurements (data) are often combined to indicate velocity (technically 'information', but considered as 'data' in the discussion that follows). Gait reporting approaches vary depending upon the type of data collected and the motivation of the report, e.g. prospective research versus clinical interpretation. Strategies for archiving gait data are also influenced by the reasons for retaining the data, e.g. the recording of a few variables for a particular project versus the storage of all available data for comparison of gait characteristics pre- and post-treatment. The motivations for gait data sharing also dictate to some degree the approach taken. For example, it is relatively simple for a pair of individual investigators to implement a strategy for combining data sets for research

Three-dimensional Analysis of Human Locomotion. Edited by P. Allard, A. Cappozzo, A. Lundberg and C. Vaughan
 ISBN 0 471 96949 4

TECHNICAL NOTES

JOURNAL OF APPLIED BIOMECHANICS, 1999, 15, 292-302

Evaluation of a Treadmill With Vibration Isolation and Stabilization (TVIS) for Use on the International Space Station

The TVIS Study Group

Jean L. McCrory, David R. Lemmon, H. Joseph Sommer, Brian Prout, Damon Smith, Deborah W. Korth, Javier Lucero, Michael Greenisen, Jim Moore, Inessa Kozlovskaya, Igor Pestov, Victor Stepansov, Yevgeny Miyakinchenko, and Peter R. Cavanagh

A treadmill with vibration isolation and stabilization designed for the International Space Station (ISS) was evaluated during Shuttle mission STS-81. Three crew members ran and walked on the device, which floats freely in zero gravity. For the majority of the more than 2 hours of locomotion studied, the treadmill showed peak to peak linear and angular displacements of less than 2.5 cm and 2.5°, respectively. Vibration transmitted to the vehicle was within the microgravity allocation limits that are defined for the ISS. Refinements to the treadmill and harness system are discussed. This approach to treadmill design offers the possibility of generating 1G-like loads on the lower extremities while preserving the microgravity environment of the ISS for structural safety and vibration free experimental conditions.

Key Words: microgravity, locomotion, kinematics, running, space station

The zero-gravity environment can affect an astronaut's body in a variety of adverse ways. Possible changes include bone demineralization, muscular atrophy, decreased blood volume, decreased cardiovascular fitness, and neurological decrements (Convertino & Sandler, 1995; Nicogossian, Sawin, & Huntoon, 1994). The bone loss and muscle atrophy of the lower extremity that occur in weightlessness have been widely documented (Convertino, 1990; Convertino & Sandler, 1995; Grigoriev & Kozlovskaya, 1988; Grigoryeva & Kozlovskaya, 1987; LeBlanc et al., 1998; Whedon, 1984). The bones of the lower extremity appear to be at greater risk of demineralization

J.L. McCrory, D.R. Lemmon, H.J. Sommer, B. Prout, and P.R. Cavanagh are with the Center for Locomotion Studies at The Pennsylvania State University, University Park, PA 16802. P.R. Cavanagh is also based with the Departments of Kinesiology and Biobehavioral Health, Orthopaedics and Rehabilitation, and Medicine at Penn State University, University Park and Hershey, PA. D. Smith is with Lockheed-Martin Engineering Services, Houston, TX 77058. D.W. Korth and J. Lucero are with the Wyle Laboritories, Houston, TX 77058. M. Greenisen and J. Moore are with the NASA Johnson Space Center, Houston, TX 77058. I. Kozlovskaya, I. Pestov, V. Stepansov, and Y. Miyakinchenko are with the Institute of Biomedical Problems, Moscow, Russia.

than the bones of the upper body during space flight. If 5% of total body calcium were lost after 1 year in space, there would be a 25% loss of calcium in the lower extremity (Whedon, 1984). The loss in muscular strength during space flight can be just as profound. It is also usually greater in the legs than in the upper body and greater in extensor muscles than flexors. Data from both the U.S. and Russian space programs have shown losses in force development of up to 50% in the knee extensor muscles (Convertino, 1990; Grigoriev & Kozlovskaya, 1988; Grigoryeva & Kozlovskaya, 1987). Appropriate countermeasures to these changes are important to allow astronauts to perform mission duties, to allow safe egress from the spacecraft in case of emergencies, and to minimize time needed for post-flight rehabilitation.

To date, no effective countermeasure has been identified for either bone loss or muscle atrophy. There are strong indications, however, that exercise will form a crucial part of any protocol to minimize the adverse effects of space travel. This is likely to require both concentric and eccentric muscle actions together with ground reaction forces that resemble those encountered on Earth (Cavanagh, 1986; Convertino & Sandler, 1995).

In a 1G study, Cavanagh, Davis, and Miller (1992) examined the ergometer, rower, and treadmill in terms of the maximal perpendicular load and loading rates elicited by each form of exercise. The average maximal loads for the treadmill, rower, and ergometer were 1628 N, 307.3 N, and 371.2 N, respectively (Cavanagh et al., 1992), and the average loading rates were 5.9×10^4 N/s, 2.91×10^3 N/s, and 1.55×10^3 N/s, respectively (Cavanagh et al., 1992). These results demonstrate that the treadmill is clearly the most effective of the three modes of exercise in producing high loads and high loading rates on the lower extremity.

A passive treadmill bolted to the floor of the mid-deck has been flown on board the space shuttle since the third shuttle flight (Thornton, 1989). The ground reaction forces between the astronaut's foot and the treadmill were transmitted to the vehicle, causing considerable vibration. This is of concern not only because of the initiation of possibly unacceptable structural vibration in the vehicle, but also because many on-board experiments are required to be conducted in the absence of vibration.

NASA has specified the requirements for maximum allowable vibration as a function of frequency at various sites on the ISS. Vibrations at certain important locations where experiments are conducted may not exceed 0.8 μg at a frequency of 0.4 Hz, while vibrations at 10 Hz may not exceed 50 μg. Although locomotor exercise is the most promising exercise countermeasure to space-induced osteoporosis and muscular atrophy, the current treadmill, when hard-mounted to the spacecraft structure, produces vibrations that significantly violate microgravity disturbance limits for research on-board the Station.

Recently, a motorized treadmill was developed to permit exercise while not exceeding the vibration limits (1999). The Treadmill Vibration Isolation and Stabilization System (TVIS) is, for all practical purposes, a free-floating treadmill, held within its operational volume by highly flexible wire rope isolators. The treadmill requires a stabilization system to provide a usable running platform in the absence of significant mechanical coupling to the spacecraft. This stabilization system consists of two separate subsystems: a linear hybrid active/passive inertial throw-mass system and a gyroscope designed to precess in the pitch plane to stabilize against roll. The TVIS is illustrated in Figure 1.

There are four linear throw-mass stabilizers, one at each corner of the treadmill, oriented in the *Y* axis (headward-footward direction). These stabilizers hold the treadmill

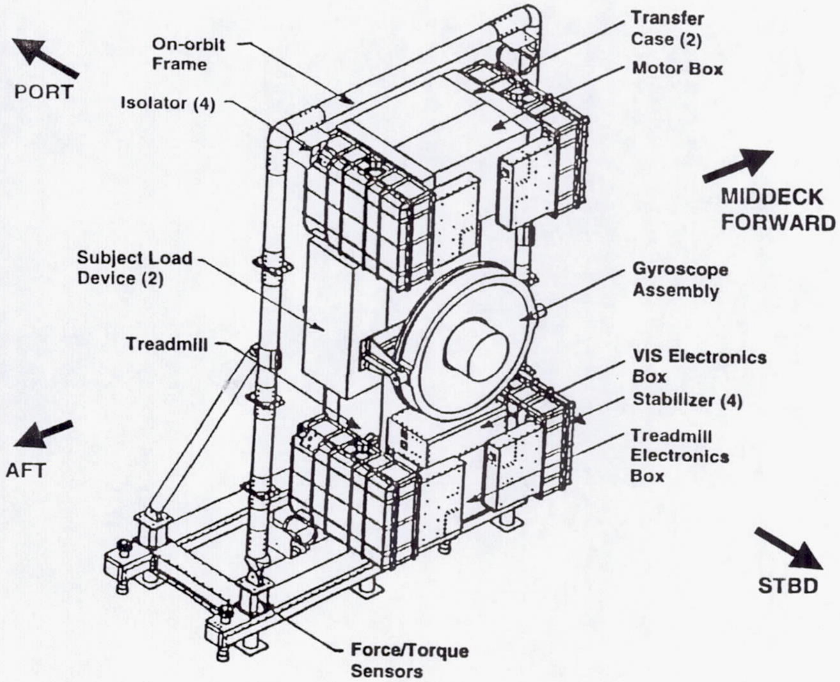


Figure 1 — The vibration isolated treadmill and its components. Gyroscope assembly is on the bottom (opposite side as the running surface). The treadmill is shown as it was mounted during use on the STS-81 flight, including the on-orbit frame and force-torque sensors.

against bounce and pitch. Each stabilizer consists of a 60-lb. steel throw-mass guided by a linear bearing and connected by springs to the stabilizer housing. The spring stiffness is set to produce a passive stabilization resonance oscillation of the throw-mass at 2.7 cycles per second.

Also connected to the throw-mass, in mechanical parallel with the springs, is a linear motor. This motor is the actuator of a closed loop control system. The control system uses accelerometers to sense heave and pitch, and moves the four throw-masses as required to reduce these disturbances. These four accelerometer signals are processed in the vibration isolation system electronics box located on the under-side of the treadmill to derive a control input signal of Y-axis translational acceleration and another control input signal of angular acceleration in the pitch plane. The two control input signals are further processed by analog control electronics in the implementation of two separate and independent closed loop feedback control systems. These control systems apply the input signals for the linear motors with the goal of holding the TVIS stationary against Y-axis translation and pitch rotation disturbances.

The linear stabilizers are actually hybrid active/passive inertial stabilizers, because they incorporate passive springs in mechanical parallel with the linear motor to assist in moving the stabilizing inertial throw-mass. In essence, the springs assist the

motor in the task of accelerating the throw-mass in the implementation of the feedback control system. The result is that the average power requirement for the linear stabilizers is less than 10% of what it would be if the springs were not present.

The total mass of the TVIS is 320 kg. Its rotational inertia for pitch, roll, and yaw are 43.4 kg · m^2, 21.0 kg · m^2, and 40.8 kg · m^2, respectively. The center of mass of the TVIS is located at the geometric center of the running platform, 14.2 cm below the level of the running surface. The gyroscope has a rotational moment of inertia of 0.76 kg · m^2 and operates at a nominal speed of 1600 rpm.

The objective of this research was to use three-dimensional (3D) photogrammetry to measure the translational and rotational stability of the tread surface of the TVIS system during walking and running on an actual shuttle mission. The effectiveness with which the TVIS system attenuated the vibration transmitted from the treadmill to the Shuttle mid-deck during exercise was also determined.

Experimental Design

Subjects

Three astronauts on-board STS-81 served as subjects for this experiment. Their personnel designations were as follows: Pilot (Subject 1, 76.8 kg), Mission Specialist-3 (Subject 2, 86.8 kg), and Commander (Subject 3, 83.6 kg).

Treadmill Protocol

Prior to the exercise session, each astronaut was required to don a harness which was then connected to two Subject Load Devices (SLDs), located on either side of the treadmill. The SLDs consisted of a set of linear springs (housed within a metal box for safety purposes) that provide a gravity replacement to keep the astronaut tethered to the treadmill and provide loading to the lower extremity during exercise. The spring constants of the devices were approximately 1057 N/m with a variable pre-load setting. The astronauts were instructed to adjust the amount of load in the SLDs so that they were being pulled to the treadmill at approximately two thirds of their body weight. A digital readout of the tethering load was visible on the treadmill control panel.

Each astronaut was requested to perform a 30-min exercise session on TVIS. This session included a minimum of 2 min at each of the following speed ranges: slow walk (1–2.5 mph), moderate walk (2.5–3.5 mph), jog (4–6 mph), and run (6–7 mph), while the remainder of the 30-min exercise session was at whatever speed they preferred. Data were collected on Flight Day 3 and Flight Day 8.

Photogrammetry

Markers were placed on the SLD on the left-hand side of the treadmill and on the TVIS frame. The O-shaped markers, made of black electrical tape, had an outside diameter of 0.75 in. and an inside diameter of 0.38 in. The physical locations of 12 black ring targets affixed to the SLD were determined prior to the flight from close-range still photographs.

Two Canon Hi8 L1A video cameras (60 Hz) recorded the motion of the SLD box on the left side of the treadmill relative to the vibration isolation frame (see Figure 2). One of these camcorders was attached via a Bogen arm to the camera mount on the ceil-

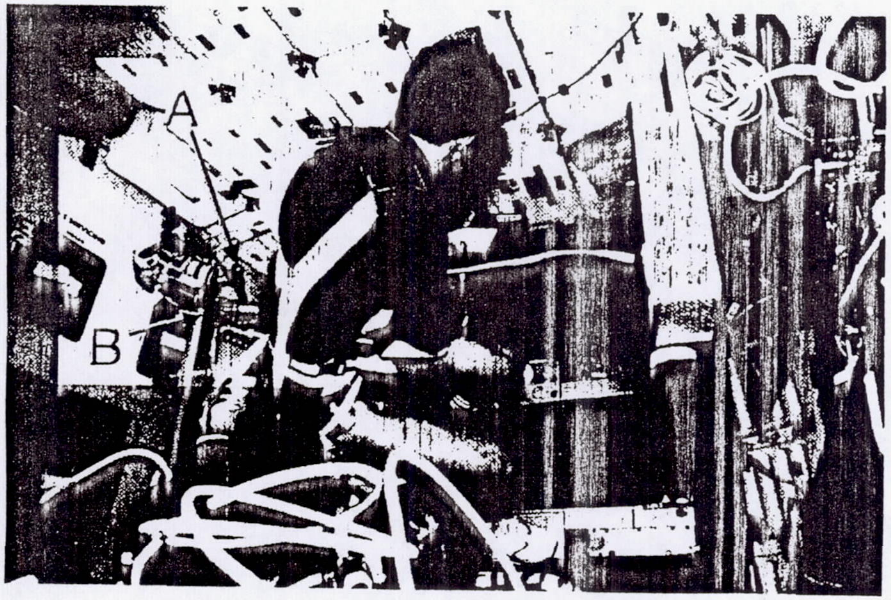


Figure 2 — The experimental arrangement in the mid-deck of STS-81. (a) One of the Hi-8 video cameras mounted on a Bogen arm to the floor of the mid-deck. (b) One of the force cubes designed to measure vibration transmitted to the vehicle. Note that three of the circular ring targets used for video analysis are visible just in front of the subject's left foot, and two others are visible immediately behind his left heel. Note also that the photograph is *rotated 90° clockwise* from a vehicle reference frame. Vehicle fore is "up," and vehicle aft is "down," as viewed in the photograph.

ing of the Shuttle's mid-deck, while the second camcorder was attached via a Bogen arm to the TVIS frame near the floor of the mid-deck. A pulse from the flash of a still camera was used to synchronize the data on the two video tapes. A third camcorder was used to qualitatively document the motion of the astronaut on the treadmill. A diagram of the set up is shown in Figure 2. Two-dimensional (2D) image coordinates of the targets were obtained by manual digitization using a Peak5 Motion Analysis System and a 386/20 Everex personal computer interfaced with a Sanyo GVR-S955 Super VHS VCR and a Sony video monitor.

Monocular nonlinear photogrammetry calibration procedures using an iterative Marquardt-Levenberg algorithm (Marquardt, 1963) were employed to estimate external and internal camera parameters, respectively, for each camera. Image data from the first image in each data sequence coupled with the physical locations of the ring targets relative to the SLD were used as respective photogrammetric calibration standards. Six external parameters describing camera pose (position and attitude) and two internal camera parameters describing focal length and radial lens distortion were estimated for each camera, respectively. External camera parameters and focal length were comparable across data sets. Lens distortion was typically less than 10% and well within compensation ranges. The root-mean-square (RMS) of image coordinate calibration residuals was typically less than one pixel.

Two-dimensional image coordinates for the ring targets from both cameras in synchronized images were combined to determine 3D global locations for the ring targets using 3D ray tracing triangulation. The global locations of the ring targets were then available to determine TVIS pose for each pair of images using traditional 3D rigid body kinematics. Each sequence of TVIS pose measurements was subsequently low-pass filtered with a cut-off frequency of 10 Hz using a fourth-order zero-phase recursive Butterworth filter.

Selection of Samples for Analysis

In order to describe both the typical and atypical motions of the vibration isolated treadmill, a variety of samples were chosen. Prior to digitization, the video tapes were viewed to qualitatively assess the motion of the treadmill as each subject ran and walked. In general, there was only a small amount of treadmill movement (see results below), and typical segments were termed *small motion*. Segments chosen because of their excessive treadmill movement were termed *large motion*.

Force Transmission

On-orbit forces and torques were measured by the TVIS data acquisition system, which is comprised of four force cubes serving as the mechanical attachment means between the isolated TVIS frame and the mid-deck floor (see Figure 2). Forces transmitted across the TVIS wire rope isolators were exerted on the TVIS frame and sensed by the force cubes. The TVIS frame was made as rigid as possible, but had a noticeable flexibility at the top (front of the treadmill) due to the cantilever arrangement. The frame exhibited flexibility manifest as a natural vibration in the pitch plane of approximately 10 Hz (frame alone, without TVIS attached). No natural vibration of the frame was noted in any other degree of freedom. However, the frame was much stiffer in all degrees of freedom than the wire rope isolators, which were used to attach the TVIS. Therefore, the frame flexibility played no significant role in treadmill dynamics during normal operation. Frame stiffness became an issue only when TVIS motion was excessive and the wire rope isolators reached the end of the normal functional envelope, which imposed a sudden jerk on the treadmill and on the frame.

Each force cube measured its three principal axis forces at a sample rate of 60 Hz. The analog signals were low pass filtered with an eighth order elliptic filter having a corner frequency of 30 Hz prior to digitization using a 12-bit ADC. The digitized signals were processed by a coordinate transformation matrix to produce the three treadmill principal axis forces and torques. These data were supplied to the ISS Microgravity Team for analysis, which consisted of using the measured forces as input to a large scale finite element model of the ISS. The outputs from this model were the predicted transients at important experimental sites in the station, which were then compared with the allowable microgravity allocation for the treadmill described above.

A TVIS coordinate reference frame at the center of the top surface of the treadmill belt was defined using the International Society of Biomechanics standards. The axes of the motion analysis were such that the X axis was in the treadmill fore-aft direction (fore positive), the Y axis was in the treadmill normal direction (up positive), and the Z axis was in the treadmill right-left direction (right positive).

Results

Subjective Observations

On the 2 days of testing, a total of 136 min of walking and running activity was captured on videotape from the three subjects. Subjectively, it appeared that the vibration isolated treadmill provided a remarkably stable platform for locomotion throughout the vast majority of the exercise periods. However, on only six occasions, a limited cycle pitch oscillation—in which the treadmill oscillated from maximum nose-down limit to maximum nose-up limit—was induced in the system and the subjects stopped the treadmill to regain stability.

Photogrammetry

Data from 12 trials were digitized. Eight trials were identified as small motion trials and 4 as big motion trials. These trials included (a) from Subject 1: 2 walks, 1 run, 2 big motion runs, 1 GYRO OFF walk, and 1 GYRO OFF run; (b) from Subject 2: 1 walk, 1 run, and 1 big motion run; and (c) from Subject 3: 1 walk and 1 big motion run. Data were collected at 60 Hz. Each trial consisted of a minimum of 240 frames of data. Data collected synchronously from both cameras were combined to determine 3D global locations for the ring targets using 3D ray tracing triangulation.

The motion of the TVIS treadmill is reported as three translational motions (antero-posterior thrust, inferior-superior bounce, and right-left shift) and three rotational motions (roll about the X axis, yaw about the Y axis, pitch about the Z axis). All motions are described as occurring about the center of the tread surface of the TVIS. The data were normalized such that all motion was referenced to the movement of the treadmill at the first data point in the trial. Table 1 shows the average values of the range of motion for each of the six motions during typical walking and running trials. All these trials were designated as small motion trials. It is apparent from Table 1 that the treadmill was generally more stable during the running trials than during the walking trials. However, in either mode of locomotion, typical values for peak to peak displacement of the treadmill were only 1–2.5 cm in translation and 0.5–1.5° in rotation (except for the 2.4° pitch in walking). When an astronaut was walking on the treadmill, the motions of bounce and pitch were almost twice as great as motion in any other directions.

During the "large motion" trials, the amplitude of treadmill motion was approximately two to three times greater with mean values of 4.9 ± 1.38, 3.9 ± 1.25, and 3.0 ± 1.01 cm for thrust, bounce, and shift, respectively, and rotations of 4.1 ± 1.39°, 1.7 ± 0.38°, and 7.3 ± 2.00° in yaw, roll, and pitch, respectively. The data from typical small and large motion running are plotted in Figure 3.

Table 1 Average Walking and Running Range of Motion Values for Each of the Six Motions During Typical "Small Motion" Trials (Average ± RMS)

Activity	Thrust (cm)	Bounce (cm)	Shift (cm)	Yaw (°)	Roll (°)	Pitch (°)
Walk	1.4 ± 0.38	2.5 ± 0.53	1.0 ± 0.29	0.6 ± 0.16	1.0 ± 0.19	2.4 ± 1.67
Run	1.2 ± 0.27	1.7 ± 0.43	1.0 ± 0.27	0.5 ± 0.10	1.2 ± 0.20	1.5 ± 0.33

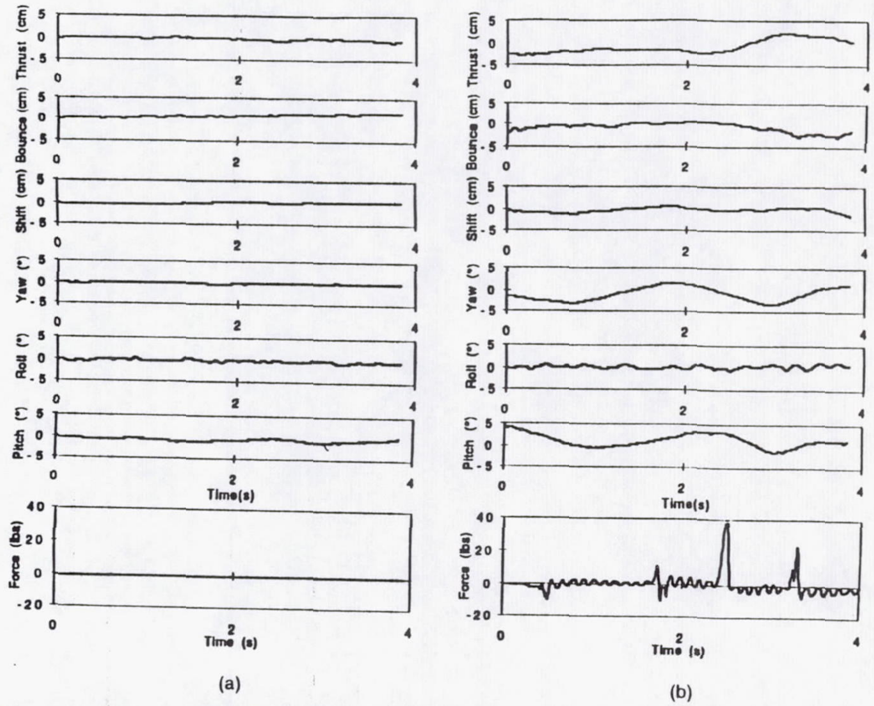


Figure 3 — Linear and angular displacements of the vibration isolated treadmill, together with the force transmitted to the vehicle. (a) Record from a typical small motion run. (b) Record from one of the six large motion events.

The gyroscope on the underside of the TVIS was designed to reduce roll. In order to determine the effectiveness of the gyroscope, single walking and running trials with the gyroscope turned off were collected from Subject 1. During walking, the roll was 3.2° greater in the GYRO OFF trial than in the GYRO ON trial. The gyroscope was less necessary in the running conditions, in which the roll was only 1.0° greater when the gyroscope was not being used.

Force Transmissions

Six sequences of forces transmitted to the TVIS frame (and therefore to the mid-deck of the Shuttle) were supplied to the NASA Microgravity Team and used as input to the finite element model of the ISS. It was determined that the integrated acceleration responses over any 10-s interval due to all six treadmill disturbances met the microgravity requirements as long as the Active Rack Isolation System (ARIS) was used at the experimental site to provide additional vibration isolation. In fact, all trials met the requirements without ARIS except for two large motion trials from Subject 2. Examination of the force transients that occurred during the limit cycle pitch oscillations indicated that there would be a significant possibility of exceeding the microgravity allocation once the TVIS was mounted in the ISS rather than on the somewhat flexible frame that was used in this shuttle experiment.

Discussion

This research has demonstrated that the TVIS provides a relatively stable surface for locomotion, particularly during running. The relatively small linear and angular displacements of the treadmill (<2.5 cm translation and <1.5° rotation) that were typical of the majority of the activity periods appear to be well within the limits of safe and comfortable locomotion. Treadmill stability in running was better than during walking, and this can presumably be related to the different foot placement in the two activities and to the operation of active stabilization at higher speeds (whereas slower speeds relied on the passive stabilization). However, it is clear that the problem of limit cycle pitch oscillation needs to be solved by design changes to allow for safe running and to allow exercise bouts to be completed without the need to stop and stabilize the treadmill.

It is interesting to note that the subject who ran on both experimental days was subjectively much more stable on Day 2 than on Day 1. In 9.5 min of running on Day 1, he needed to stop three times because of limit cycle pitch oscillation. On Day 2, he ran for 35 min with only one stop. This suggests that training on a high fidelity ground simulator prior to flight may have a significant effect on the subject's ability to run successfully on the vibration isolated treadmill. Running on a free-floating surface is a novel task, and it is reasonable to assume that a period of training may be required to successfully learn the new skills required.

This experiment demonstrated stability of the TVIS in reaction to the impact and propulsive forces arising from walking and running but also exhibited a pitch oscillation resonance modality driven by a slow fore-aft wandering of the astronaut on the running surface, as occurred during the big motion running trials. This pitch oscillation was shown to develop into a limit cycle interaction with the wire rope isolators. This post flight evaluation has resulted in three planned modifications to the TVIS design to eliminate the pitch oscillation resonance phenomenon. First, a device has been designed to limit the fore-aft excursion of the runner to prevent or reduce the development of the oscillation while still allowing the movement necessary during normal walking and running. Also, the wire rope isolators are being replaced with spring/dashpot isolators, which will impose a low level of damping on TVIS motion even within the nominal operational envelope. And lastly, the control loop gain and phase margins are being adjusted to provide more stabilization against the development of the resonance phenomenon.

Because astronauts are weightless in a microgravity environment, they require the use of a harness and SLD system to supply them with the force necessary to return themselves to the treadmill belt. As an astronaut impacts or pushes off from the treadmill, it creates a torque on the TVIS. This torque is determined by the placement of the foot and the amount of load in the SLDs. The results of this study have shown that, in spite of this torque, the TVIS provided a stable surface on which to exercise.

However, astronauts in this study did not load the SLDs to full body weight, mainly because of the discomfort caused by the harness. As the tethering load increased, the harness tended to slip over the subject's hips, placing a greater percentage of load on the shoulders. Also, the total amount of load decreased as the harness slipped, because the springs had to be stretched less to reach from the SLDs to the harness. Previous 1G studies in zero-gravity simulation have shown that as the level of load decreases, the magnitude of the ground reaction forces decrease (McCrory et al., 1996).

It is clear that the success of any exercise intervention involving locomotion will depend critically on the comfort of the harness, which is transmitting the gravity replacement load to the body. Since the inception of this experiment, a new rigid harness

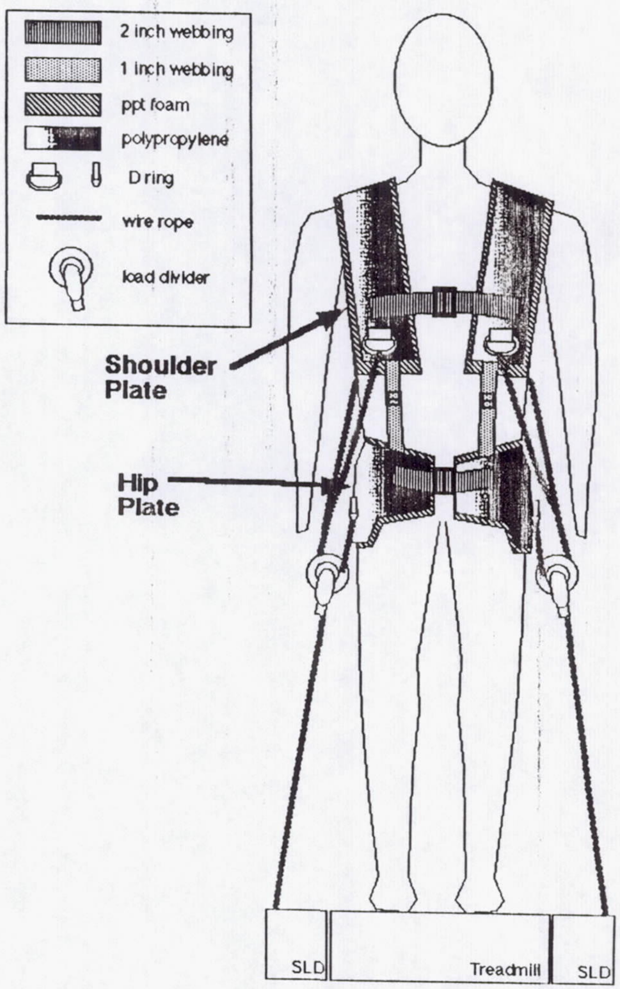


Figure 4 — The new harness designed to be more comfortable and provide a more evenly distributed load between the hip and shoulders of an exercising astronaut.

(shown in Figure 4) has been developed at Penn State. The new harness has two advantages over the previous one. First, loads are applied to the shoulders and hips through molded polypropylene plates padded with soft foam. This reduces pressure and prevents the harness from slipping. Second, pulleys provide a consistent load distribution between the shoulders and hip. This prevents overloading of the shoulders and spine. The harness has now been used during treadmill exercise on another space shuttle mission, STS-84. Details of the new harness design are left to subsequent publications.

These design changes to both the TVIS and the harness will soon be evaluated in a new ground-based horizontal suspension test bed. Future tests will show whether use

of the new harness will allow increased loading in the SLD. Higher SLD loads would mean that the astronauts feet would impact and push off from the treadmill with a greater force, resulting in a greater torque being placed on the treadmill. This increased torque could affect the TVIS motion.

In summary, this study has shown that it is possible to walk and run on-orbit on a completely free floating but vibration isolated and stabilized treadmill. The hybrid control system of the TVIS keeps both linear and angular displacement within acceptable limits for the vast majority of the time. The occasional limit cycle pitch oscillations are presently problematic and need to be the subject of future design modifications. Further effort needs to be devoted towards improvements in harness design to allow full body weight forces to be applied to the body through the hips and shoulders with maximum comfort to the astronaut.

References

Cavanagh, P.R., Davis, B.L., & Miller, T.A. (1992). A biomechanical perspective on exercise countermeasures for long term spaceflight. *Aviation, Space, and Environmental Medicine*, **63**, 482-485.

Cavanagh, P.R. (1986). Biomechanical perspectives on locomotion in null gravity. *Workshop on Exercise Prescription for Long-Duration Space Flight*. NASA Conference Publication 3051, 61-67.

Convertino, V.A. (1990). Physiological adaptations to weightlessness: effect on exercise and work performance. *Exercise and Sport Science Reviews*, **18**, 119-166.

Convertino, V.A., & Sandler, H. (1995). Exercise countermeasures for spaceflight. *Acta Astronautica*, **35**, 253-270.

Grigoriev, A.I., & Kozlovskaya, I.B. (1988). Physiological responses of skeletomuscular system to muscle exercises under long-term hypokinetic conditions. *The Physiologist*, **31**, S83-S87.

Grigorieva, L.S., & Kozlovskaya, I.B. (1987). Effect of weightlessness and hypokinesia on velocity and strength properties of human muscles. *Kosmicheskaia Biologiia I Aviakosmicheskaia Meditsina*, **21**, 27-30.

LeBlanc, A., Shackelford, L., & Schneider, V. (1998). Future human bone research in space. *Bone*, **22**(Suppl. 5), 113S-116S.

Marquardt, D.W. (1963). An algorithm for least-squares estimation of non-linear parameters. *Journal of the Society of Industrial and Applied Mathematics*, **11**, 431-441.

McCrory, J.L., Baron, H.A, Derr, J.A., Davis, B.L., & Cavanagh, P.R. (1996). Subject load-harness interaction during zero-gravity treadmill exercise. *Proceedings of the American Society of Biomechanics XX Annual Meeting*, Atlanta, GA.

Nicogossian, A.E., Sawin, C.F., & Huntoon, C.L. (1994). Overall physiologic response to space flight. In A.E. Nicogossian, C.L. Huntoon, & S.L. Pool (Eds.), *Space physiology and medicine* (3rd ed.). Philadelphia: Lea & Febiger.

Smith, D.C. (in press). TVIS: A vibration isolated treadmill for the International Space Station. *Lockheed-Martin Technical Report*.

Thornton, W. (1989). Work, exercise and space flight. III. Exercise devices and protocols. In B.A. Harris, Jr. and D.F. Stewart (Eds.), *NASA CP 3051 Workshop on Exercise Prescription for Long-Duration Space Flight* (pp. 31-42). Houston, TX: NASA Office of Management, Scientific and Technical Information Division.

Whedon, G.D. (1984). Disuse osteoporosis: Physiological aspects. *Calcified Tissue International*, **36**, 146-150.

Acknowledgment

This study was supported by a grant from Wyle Laboratories (formerly Krug International) to the Center for Locomotion Studies at The Pennsylvania State University.

JOURNAL OF APPLIED BIOMECHANICS, 1999, **15**, 303-317

A Procedure for the Automatic Determination of Filter Cutoff Frequency for the Processing of Biomechanical Data

John H. Challis

This article presents and evaluates a new procedure that automatically determines the cutoff frequency for the low-pass filtering of biomechanical data. The cutoff frequency was estimated by exploiting the properties of the autocorrelation function of white noise. The new procedure systematically varies the cutoff frequency of a Butterworth filter until the signal representing the difference between the filtered and unfiltered data is the best approximation to white noise as assessed using the autocorrelation function. The procedure was evaluated using signals generated from mathematical functions. Noise was added to these signals so that they approximated signals arising from the analysis of human movement. The optimal cutoff frequency was computed by finding the cutoff frequency that gave the smallest difference between the estimated and true signal values. The new procedure produced similar cutoff frequencies and root mean square differences to the optimal values, for the zeroth, first, and second derivatives of the signals. On the data sets investigated, this new procedure performed very similarly to the generalized cross-validated quintic spline.

Key Words: filtering, data processing, noise removal

The analysis of human movement often requires the estimation of derivatives from sampled displacement data contaminated with noise. These noisy data can be obtained, for example, using cine-film or video based motion analysis systems or from the use of goniometers. There are a large number of potential sources of noise in the recording process, many of these sources can be minimized by careful experimental procedures, but some noise will still remain. This random noise is normally removed by using a procedure that low-pass filters the data—the assumption being that the noise is white and so has a flat power spectrum, and is not correlated between samples. By contrast, the movement signal is assumed to be band-limited, occupying predominantly the low frequencies. The low-pass filter therefore removes the high frequency components, leaving the movement signal unaffected, or at least the majority of it, and removing most of the noise. The problem is to select the degree of filtering that optimally achieves this. In this sense, optimal means obtaining the best approximation to the true signal underlying the noisy sampled signal. Subsequent to the determination of the low-pass filtered signal, the data are numerically

The author is with the Biomechanics Laboratory in the Department of Kinesiology at The Pennsylvania State University, University Park, PA 16802-3408.

THE IMPLICATIONS OF REDUCED GROUND REACTION FORCES DURING SPACE FLIGHT FOR BONE STRAINS

[1,3]Marc M. Peterman, [1,3]Andrew J. Hamel, [1,2]Neil A. Sharkey,
[1,2,3]Stephen J. Piazza, and [1,2,4]Peter R. Cavanagh
[1]The Center for Locomotion Studies and the Departments of [2]Kinesiology; [3]Mechanical Engineering; [4]Biobehavioral Health, Medicine, Orthopaedics and Rehabilitation;
The Pennsylvania State University, University Park and Hershey, PA
Email: celos@psu.edu Web: www.celos.psu.edu

INTRODUCTION

The specific mechanisms regulating bone mass are not known, but most investigators agree that bone maintenance is largely dependent upon mechanical demand and the resultant local bone strains. During space flight, bone loss such as that reported by LeBlanc et al. (1996) may result from failure to effectively load the skeleton and generate sufficient localized bone strains.

In microgravity, a gravity replacement system can be used to tether an exercising subject to a treadmill (Davis et al. 1993, McCrory 1997). It follows that the ability to prevent bone loss is critically dependent upon the external ground reaction forces (GRFs) and skeletal loads imparted by the tethering system. To our knowledge, the loads during orbital flight have been measured only once (on STS 81). Based on these data and data from ground based experiments, it appears likely that interventions designed to prevent bone loss in micro-gravity generate GRFs substantially less than body weight. It is unknown to what degree reductions in external GRFs will affect internal bone strain and thus the bone maintenance response.

To better predict the efficacy of treadmill exercise in micro-gravity we used a unique cadaver model to measure localized bone strains under conditions representative of those that might be produced by a gravity replacement system in space.

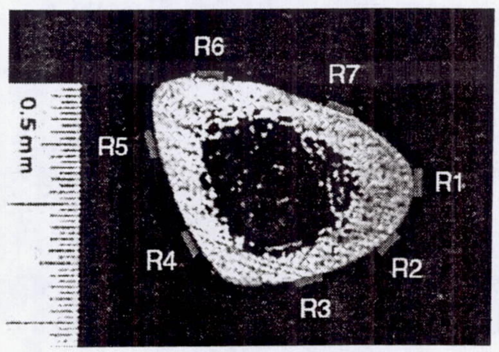
METHODS

Cadaver limbs were mounted into a dynamic loading apparatus that reproduces the kinetics and kinematics of the tibia, foot, and ankle during the stance phase of gait (Sharkey and Hamel, 1998). The device is able to produce GRFs equivalent to those produced in life. Physiologic muscle actions are simulated using force feedback controlled linear actuators interfaced with the tendons of the specimen using freeze clamps.

The distal third of the tibiae were each instrumented with seven miniature strain gauge rosettes (Micro-Measurements Group, Inc. EA-06-031RB-120) oriented on a transversely co-planar section (Figure 1). Dynamic gait simulations were conducted at GRFs corresponding to 25%, 50%, 75%, and 100% of body weight (BW). Strain data were collected over the entire stance phase

MEDIAL

POSTERIOR



ANTERIOR

LATERAL

Figure 1. Tibial cross-section showing strain gauge placement.

and the results were used as input for a computer model of bone strains over the stance phase.

RESULTS

Decreased GRFs produced proportional decrements in peak tibial strains (Figure 2). At all gauge locations, development of maximum strains within the specimen corresponded to the second peak of the GRF profile. The overall strain maxima for each condition was compressive and occurred along the posterio-medial border of the tibial cross section. Maximum tensile strains occurred anteriorly on the tibial crest. Animations of the strain profiles from heel-strike to toe-off were created using SIMM and MATLAB and will be presented.

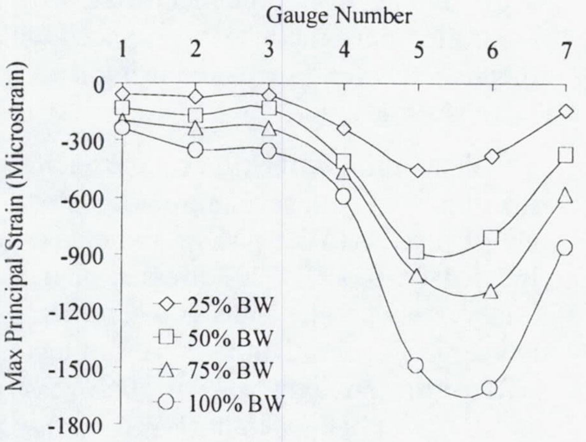


Figure 2. Peak maximum principal strains on the tibia for varying GRFs during simulated walking.

DISCUSSSION

Several investigators have attempted to measure *in vivo* strains in the human tibia, but these studies have been somewhat limited (Lanyon et al., 1975; Milgrom et al., 1996; Burr et al., 1996; Aamodt et al., 1997). *In vivo* human studies have focused on the antero-medial aspect of the tibia due to ethical concerns and technical constraints. Using a robust cadaver model we found that peak strains occur at sites other than those typically measured in live human subjects.

The strain distributions measured in the current study indicate bending as the primary mode of tibial loading and muscle force (i.e. the triceps surae) as the principal modulator of bone strain. Carter et al. (1987) and Turner (1998) have derived separate theoretical equations of bone adaptation based, among other things, on the magnitude of stress or strain experienced by the bone. Our data suggest that *in-situ* bone strains can be reliably related to the external loading environment, so that these equations can be used to calculate a theoretical stimulus using GRFs as input.

CONCLUSION

Internal tibial strains can be predicted using external GRFs. This inter-relationship is important for understanding the bone stimulating potential of various exercise interventions during space flight.

REFERENCES

Aamodt A., et al. (1997). *J Orthop Res,* 15: 927-31.

Burr D.B., et al. (1996). *Bone*, 18: 405-10.

Carter D.R., et al. (1987). *J Biomechanics,* 20: 785-94.

Davis B.L., et al. (1996). *Aviat Space Environ Med,* 67(3): 235-42.

Frost H.M. (1986). *Intermediary Organization of the Skeleton*. CRC, Boca Raton.

Lanyon L.E., et al. (1975). *Acta Orthop Scand*, 46: 256-68.

LeBlanc A., et al. (1996). *J Bone Mineral Res*, 11: S323.

McCrory J.L. (1997). *Ph.D. Thesis*. Penn State University, University Park.

Milgrom C., et al. (1996). *Foot & Ankle Int*, 17: 667-71.

Turner C.H. (1998). *Bone,* 23(5): 399-407.

Sharkey N.A., Hamel A.J. (1998). *Clin Biomech*, 13: 420-433.

ACKNOWLEDGEMENTS

This work was supported in part by NASA grant NAGW-4421.

PENNSTATE

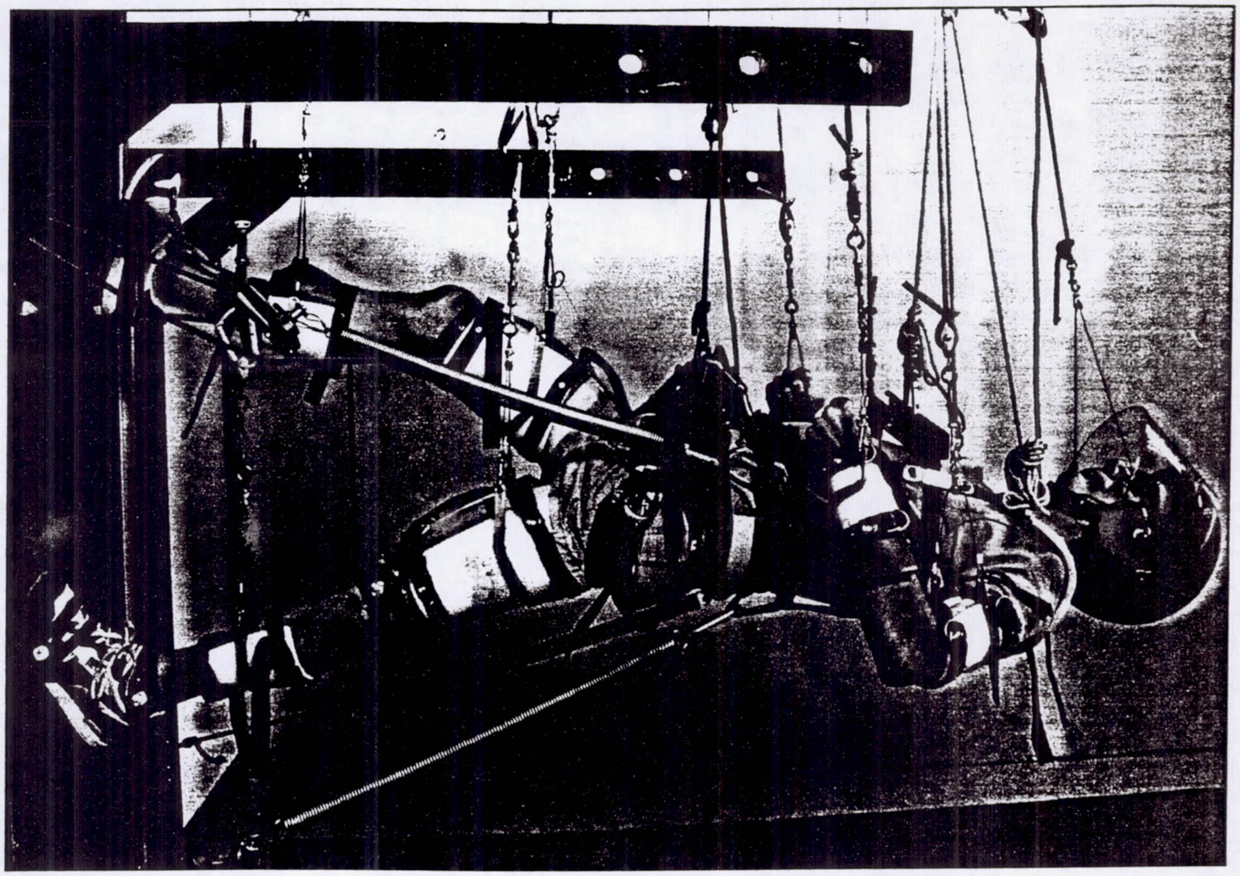
1855

Health & Human Development NEWS

Volume 6, Number 1

Fall 1992

In This Issue



Clues to better space exercise

Faking weightlessness

Of all the adverse effects that zero gravity has on the body, especially during long space flights, calcium loss may threaten astronauts' health the most. And Penn State studies suggest that the way shuttle astronauts exercise in space now may not be as helpful for slowing calcium loss as it could be.

"Studies of the physiological effects of space flight show that exercise is useful in slowing down the rate of bone loss that occurs in zero gravity," says **Brian Davis '91g ExSci.** "Unfortunately, there have been few, if any, scientific investigations into the most appropriate form for such exercise."

Davis (who recently took a position at the Cleveland Clinic Foundation) and three other researchers in the Center for Locomotion Studies conducted a three-year study of space exercise through a NASA grant to the College of Health and Human Development. Team members included H.J. Sommer III, professor of mechanical engineering, Andris Freivalds, associate professor of industrial engineering, and **Peter Cavanagh,** professor of locomotion studies and director of CELOS, who coordinated the project.

As part of the study, the researchers developed a weightlessness simulator. "Our challenge involved not only simulating weightlessness in Earth gravity, but also figuring out how to re-introduce

(continued on page 13)

Faking weightlessness

(continued from front page)

gravity on the simulator," Cavanagh says. "Without that gravity effect, treadmill exercise in space is impossible to perform."

In the Penn State Zero Gravity Locomotion Simulator, the subject is suspended horizontally by a series of bungee-like cords that are attached by cuffs at 11 points on the arms, legs, torso, chest, and head. Through a series of pulleys, the cords counteract the force of gravity on the subject's movements as he or she hangs horizontally nearly three feet off the floor.

The treadmill portion of the simulator is mounted vertically on the wall "below" the horizontal subject's feet. Comfortable running on the treadmill can be achieved only by using a set of springs that tether the subjects to the treadmill. "The springs re-introduce gravity," Davis explains. In contrast to the motorized treadmill used in the Penn State study, NASA uses a passive device that requires astronauts to lean forward and push the belt with their feet.

"If you look at NASA film of crew members exercising, you can see that they're just running on the balls of their feet without making heel contact," Cavanagh says. "There is serious reason to doubt that this kind of movement does much to prevent the bone loss problem."

Adds Davis: "Stress on the bones is required for calcium to be deposited, but astronauts probably aren't getting stress at the right rate. Given that they cannot run with Earth-normal force in zero gravity, they should run faster to achieve beneficial stress rates. This is best accomplished with a motorized treadmill." (Shuttle treadmills currently aren't motorized, because of the need to conserve precious electrical power during orbital flight.)

So far, 18 subjects have taken part in experiments with the Penn State simulator. In the meantime, NASA continues to experiment with space exercise using airplanes that fly parabolic trajectories to create weightlessness for astronauts on board. But through this method, the zero-gravity effect can be maintained for only about 30 seconds at a time. Next, the researchers hope to make direct comparisons of the same subjects running in the simulator and on aircraft.

Research at zero G

Health and Human Development researcher Peter Cavanagh gained a first-hand feel for weightlessness when he flew on a modified Boeing 707, called a KC-135, at Johnson Space Center in Houston. "The pilot flies a series of parabolic trajectories that make the participants weightless for 20 to 30 seconds during each pass," explains Cavanagh, who was there to observe a NASA experiment on treadmill running. In the in-flight experiment, NASA astronaut Jim Bagian ran on an instrumented treadmill while researchers monitored the forces exerted on his feet. Cavanagh is hoping to conduct similar experiments back at University Park, in order to compare the effectiveness of the newly constructed Penn State treadmill to the more costly aircraft simulation.

College cites three undergrads for honors research

Three HHD undergraduate students earned prizes this spring in an honors thesis competition sponsored by the college and its alumni society.

Students write an honors thesis as part of the University Scholars Program, an honors program for outstanding students. Each year the college awards cash prizes to several students whose theses show "quality of scholarship and a contribution to knowledge and the betterment of human well-being."

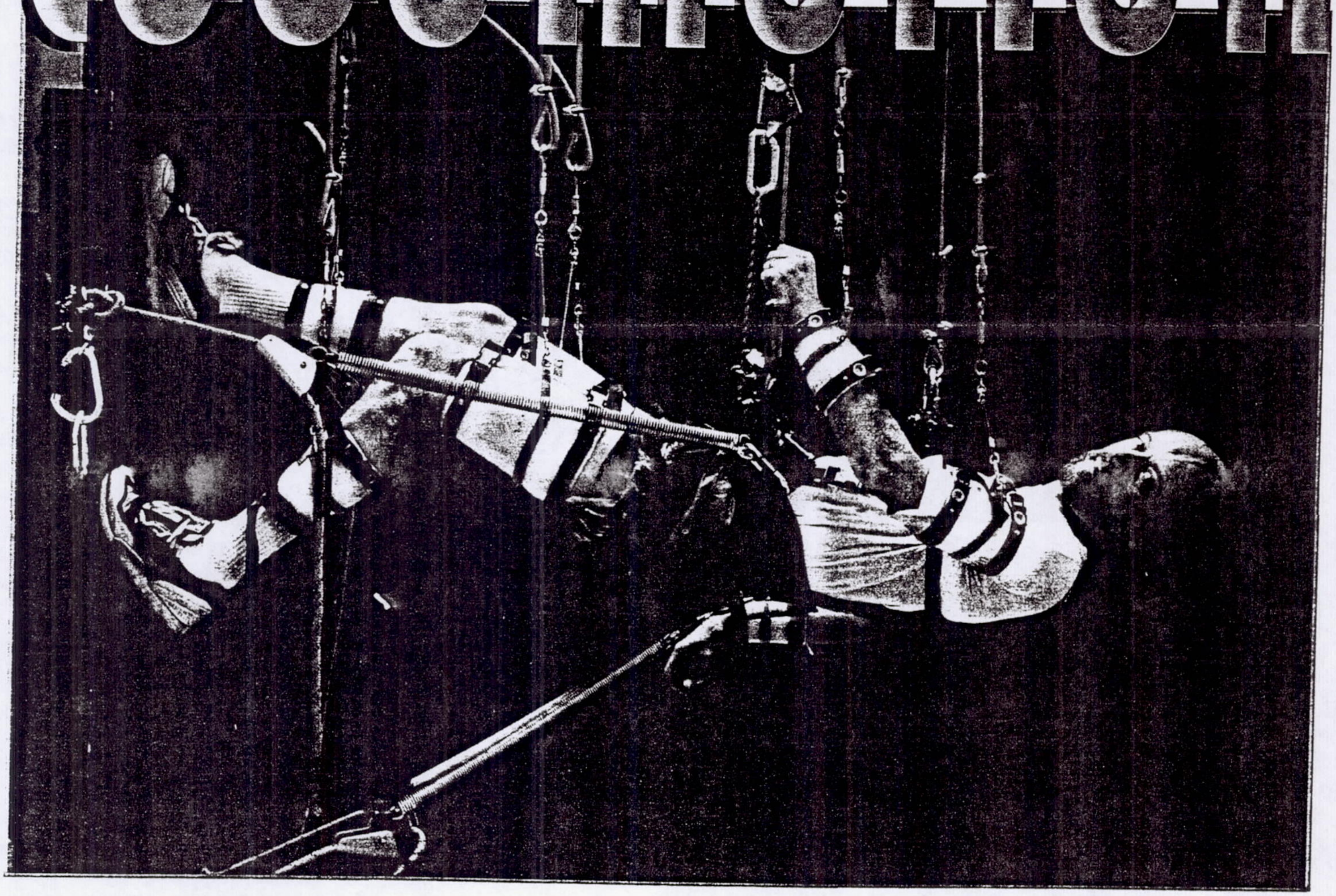
First prize: **Krista Giersch '92 HPA,** for "Alternative to Institutionalization of the Elderly: A Client Service and Reimbursement Comparison of Adult Day Care and Home Health Care in Rural Pennsylvania." Giersch worked with **Dennis Shea,** assistant professor of health policy and administration.

Second prize: **Stephanie Bourque '92 HDFS,** for "Police Officers' Attitudes Toward Domestic Violence." Bourque worked under the direction of **Jennifer Mastrofski,** assistant professor of human development.

Third prize: **Kristine Webber '92 HPA,** for "Patient Satisfaction in University Health Centers: The Development and Evaluation of a Women's Health Department Satisfacation Questionnaire." She worked under **Diane Brannon,** associate professor of health policy and administration.

Giersch is now pursuing a master's in health policy and administration at the University of North Carolina in Chapel Hill; Bourque now is attending the University of Maryland School of Law; and Webber is working for a physician group practice in Washington, D.C.

Doing the Locomotion



Up the wall: Peter Cavanagh takes a turn in the Penn State Zero Gravity Locomotion Simulator, designed to help astronauts exercise effectively in space.

BY DAVID PACCHIOLI

PHOTOGRAPHS BY JAMES COLLINS

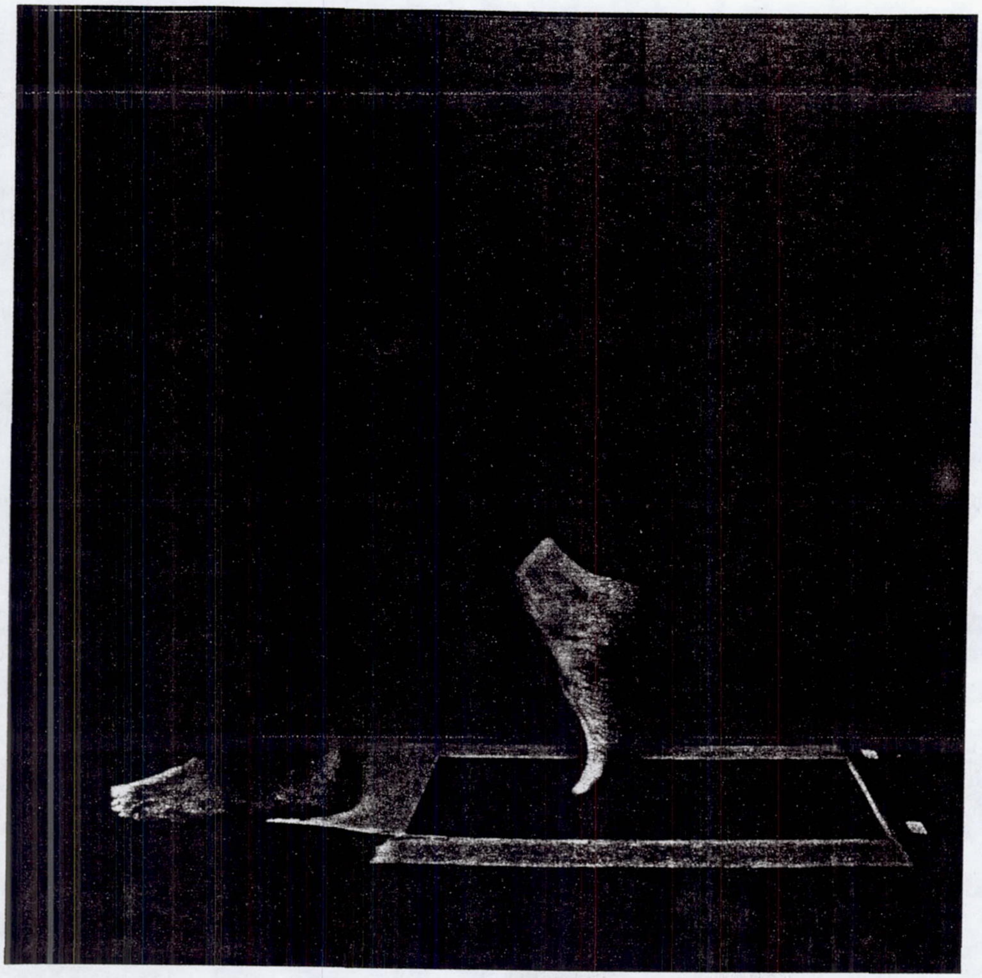
"The first thing to understand," says Peter Cavanagh, "is that there's no such thing as standing still." Cavanagh, a lean, nimble man with a neat reddish beard, pops up from his chair to demonstrate.

"Even when we think we're standing still, we're engaged in continuous sway," says the 47-year-old distinguished professor of locomotion studies, biobehavioral health, medicine, and orthopedics. "The body oscillates in an apparently random way— Call it micro-sway." He stands with his hands at his sides.

"We have looked at patterns of people who have tried to stand still."

He goes on swaying imperceptibly for a minute.

"To make it worse," he says at last, "all you have to do is close your eyes." He does so, and his movement grows noticeable.



Making tracks: *Pied de* Cavanagh completes a stride across an electronic pressure platform, which records a precise image of the sole.

"Next," says Cavanagh, "tip the head back."

Now we're swaying.

"The balance organs of the inner ear work best when the head is erect," Cavanagh says, nose pointed to the ceiling, in his faded British accent. "Tipping the head back puts them at a disadvantage."

If he wanted to take things farther, Cavanagh would now place a thickness of foam, or something equally unstable, under his feet.

"On the soles of our feet," he explains, "we have hundreds of receptors that respond continually to pressure. We also have joint-angle sensors in our ankles that note how the angle between foot and leg changes during sway. Standing on foam decreases the efficiency of this measuring system."

No foam handy, Cavanagh retakes his seat.

As director of Penn State's Center for Locomotion Studies (CELOS), Cavanagh has put numerous experimental subjects — safely tethered to an overhead track — through similarly disorienting paces, as part of a continuing effort to understand the biomechanics of balance. Keeping ourselves upright is, after all, the first step in locomotion, in getting ourselves, bipedally, from one place to the next: from the car to the mall, say, or from the couch to the refrigerator. Anybody who thinks it's a snap to get this system running in synch needs to have a look at a 10-month-old baby taking it for an early test spin.

Oh, sure, soon enough we get pretty smooth. We come to take it for granted, just as later we do the ability to drive a car. If all goes well, we cruise along, what once seemed impossible having become second nature.

But should a "pathology" develop in one of the control systems, our "postural stability" is going to drop — and chances are that we are going to drop, too, hitting the deck with a painful thud, or worse.

Such pathology develops naturally, unfortunately, with age. The body's control systems slowly degrade. When the effects of aging are combined with those of a degenerative disease like diabetes, t change can be dramatic.

People with diabetes suffer many side effects. One of the more common of these is neuropathy — loss of feeling — the feet. The anesthetic effect of neuropathy can be gruesomely complet Cavanagh tells of a carpenter who, havir trouble removing his boot, summoned l wife to help. What she noticed — and h couldn't feel — was a nail driven clear through the boot's sole, its point protru ing from the leather upper.

Even when *not* so advanced, neuropathy is a serious problem; and one of tl things it affects is balance.

"People with diabetic neuropathy swa as much with their eyes open and their heads forward as non-diabetics do with eyes closed and head back," Cavanagh reports. Their lack of sensitivity makes these people with diabetes, many of then elderly and fragile, more susceptible to falls, and thus to serious injury.

As Cavanagh's colleague, assistant professor Ge Wu, phrases it, "They don't know where their feet are."

"The foot," Cavanagh says, "is the key element in locomotion."

A skeletal foot sits on the table, reinforcing the point. The model is nicel flexible; its numerous small bones are carefully labeled, the joints connected with tiny springs.

"The feet — and the legs. We put these organs under the microscope. We want to understand them from the engineering perspective," Cavanagh says, of the work being done in CELOS.

He started into locomotion as a doctoral student at the Royal Free Medical School in London. There he scrutinized walking by measuring the electrical activity in muscle fibers. "I found that with sensitive equipment you can detect unequivocally and precisely when the muscle is working." Typically, this result got him to thinking practically: "Suppose you had somebody with one limb paralyzed and the other intact," he thought. "If you had a robotic device to move the paralyzed leg, conceivably the signal from the good leg could tell the other one when to move."

About that time, Cavanagh says, "I got into running. I ran marathons — Boston, Avenue of the Giants, Boston again. And I got interested in energetics, the mechanics of locomotion. It's like two cars that look the same but one uses twice as much gas to go the same distance. Why is it that two people can have the same body

weight, the same structure, yet one can run on 20 percent less oxygen?"

He also became interested in running injuries, which is what first swung his attention to feet. "With the running boom — this was the late '70s — had come this boom in injuries. Overuse injuries. And I realized that footwear was the key to understanding these injuries, and to preventing them: controlling excessive motion and cushioning."

At Penn State, Cavanagh became a recognized authority on running-shoe design. He consulted for major shoe manufacturers, designing a number of shoes, including one that was fitted out with a miniature computer that recorded lap times, calories burned, and other crucial data. ("An idea," he says, "whose time hasn't come.") In 1980, he authored *The Running Shoe Book.*

In 1981 Cavanagh and then graduate student Ewald Henning invented an electronic device for measuring the pressure applied to the soles of the feet during standing and walking. By standing on this pressure platform, a person could obtain a color-coded computerized image of the bottoms of the feet which pin-pointed areas of high pressure. The device would be useful for identifying potential problem areas before they resulted in injury.

Yet while developing this imaging device, Cavanagh had begun to tire of sports medicine. For one thing, he says, he grew leery of the use that was being made of his data by running-shoe manufacturers. As he grew older, he wanted to make more of a contribution to society. He envisioned another use for the pressure platform.

"I realized that it could be applied in a very different context," he says, "to help people who wanted not to run faster but to keep their feet."

Neuropathy, the loss of feeling, can strike the arms as well as the legs, hands as well as feet. It usually occurs symmetrically — if one limb is affected, so is the other — and it is worse in distal areas, those places farthest from the body's center. But it is especially a problem in the feet.

"The nerves just die," says Jan Ulbrecht, a diabetologist and member of the CELOS faculty. "It's a biochemical process that we can't explain yet."

What is known is that this loss of the protective sensation of pain in a heavy-stress area like the bottom of the foot can lead to real trouble. Unnoticed rubbing or chafing from something as minor as a tight shoe can easily result in an ulcer, which goes equally unfelt. Left untreated for weeks, months, even years, the ulcer becomes infected, and the infection spreads, eventually to bone. Too often, the final result is catastrophic: Each year over 60,000 lower-limb amputations are performed in the United States as a result of diabetes-related complications. In addition to the human toll — amputation's immense physical and emotional impact, and the poor prognosis for long-term survival — this drastic procedure is an economic nightmare. Each amputation costs $50-60,000 to perform, when total costs of surgery and rehabilitation are considered.

The thing is, Cavanagh says, many of these amputations are preventable.

Until recently, by the time a diabetic patient developed a serious foot ulcer, it was all over. Such ulcers were almost impossible to heal, and even if a clinician managed to heal one, it would soon return. Loss of blood circulation was blamed for most foot problems, and standard practice was to amputate sooner rather than later, in order to prevent even worse problems.

In 1991 Cavanagh and Ulbrecht opened up the Diabetes Foot Clinic as a joint program of CELOS and the Nittany Valley Rehabilitation Hospital. The idea was to learn more about the diabetic foot in order to improve treatment. One of their goals was a series of tests to predict who would get foot ulcers and where.

Today, having treated more than 750 patients at University Park and at a second clinic in the Hershey Medical Center, Cavanagh, Ulbrecht, and their colleague Gregory Caputo, at Hershey, argue that neuropathy, not poor circulation, is the major culprit in the diabetic foot. They have successfully healed ulcers using a special weight-bearing cast and a tough antibiotic regimen. Testimonials abound from people with diabetes who faced amputation. In September 1994, Cavanagh, Ulbrecht, and Caputo, along with colleagues at the Harvard Medical School, published a definitive guide to management of the diabetic foot in the *New England Journal of Medicine*, emphasizing careful screening and early detection.

The key, says Cavanagh, is vigilance. Especially after an initial ulcer has been healed, "The patient remains at lifetime risk," Cavanagh stresses. "You cannot relax your guard."

In addition to frequent, thorough examinations, the foundation for preventing recurrence is protective footwear.

Improving footwear design is one of the lab's current concerns. Even at the clinic, Cavanagh says, "prescribing footwear is not the science we would like it to be." It's largely a hit-or-miss process.

"What we do is, we give a patient a specially designed shoe and say, Tell us how it goes. But what if it doesn't go well? The stakes are too high. Every ulcer creates a very high risk for losing a limb."

A better way to prescribe, Cavanagh suggests, would be to do the guesswork on a computer model, to predict on a simulated foot where problems are likely to occur, instead of correcting for them after the fact. A pair of shoes could be designed in accordance with the prediction of where pressures are likely to be highest. "Then, once they're as good as they can be, we bring the patient into the lab and measure the interaction with the foot using this." He holds up a thin green insole, made of what looks like molded rubber. It's actually a portable version of the pressure platform: the insole contains 100 electrodes, which are wired to a computer on the patient's waist.

Meanwhile, CELOS researchers continue to investigate other alternatives to below-the-knee amputation. Ph.D. student John Garbalosa recently presented results of a study that looked at the efficacy of a partial amputation technique.

"Many surgeons," Cavanagh explains, "if they see an ulcer on the big toe, will take the whole leg below the knee, figuring that's eventually what's going to happen anyway. The recurrence of infection is high.

"But we think that's because of inadequate management. The partial foot has extra-special needs for protection. Given that protection, partial amputation can be quite successful."

Out on the lab floor, Ge Wu is conducting a test. A young woman in jeans and a red sweater stands on a small raised platform. The woman is fitted snugly with a black harness that loops around her shoulders, crosses her chest, and encircles her thighs like a parachute rig; the harness is connected to a strap hooked to the ceiling. At a silent signal, the platform suddenly jerks back a few inches. The woman lurches forward, her hands fly up, and she recovers — all in an instant.

Wu is trying to understand how people maintain their balance, given a sudden "sensory challenge": Or, put negatively, what makes them fall.

It's not as much fun as it looks. "Falls," says Cavanagh, "are the leading accidental cause of injury and death in the elderly."

Balance, as Cavanagh has already demonstrated, is the function of a complicated system. Three systems, actually: the visual, the vestibular (inner ear), and the proprioceptive (that's the sensory apparatus in feet and ankles). These control systems interact in a dense network of signal and feedback.

All in that instant after the bus jerks to a stop or the escalator pulls your feet out from under you, the brain gathers and integrates information from eyes, ears, and feet in order to make the proper response, to initiate commands to the appropriate muscles, to scream out the warning: *Mayday! Mayday! We're about to fall!*

"The thing that makes it tricky," says Wu, an engineer, "is that the control mechanism is over-redundant."

The human body, it seems, is designed with a certain amount of built-in overlap. To some degree, the systems cover for one another. While this is great for our survival, Wu acknowledges, it makes the system that much harder to comprehend.

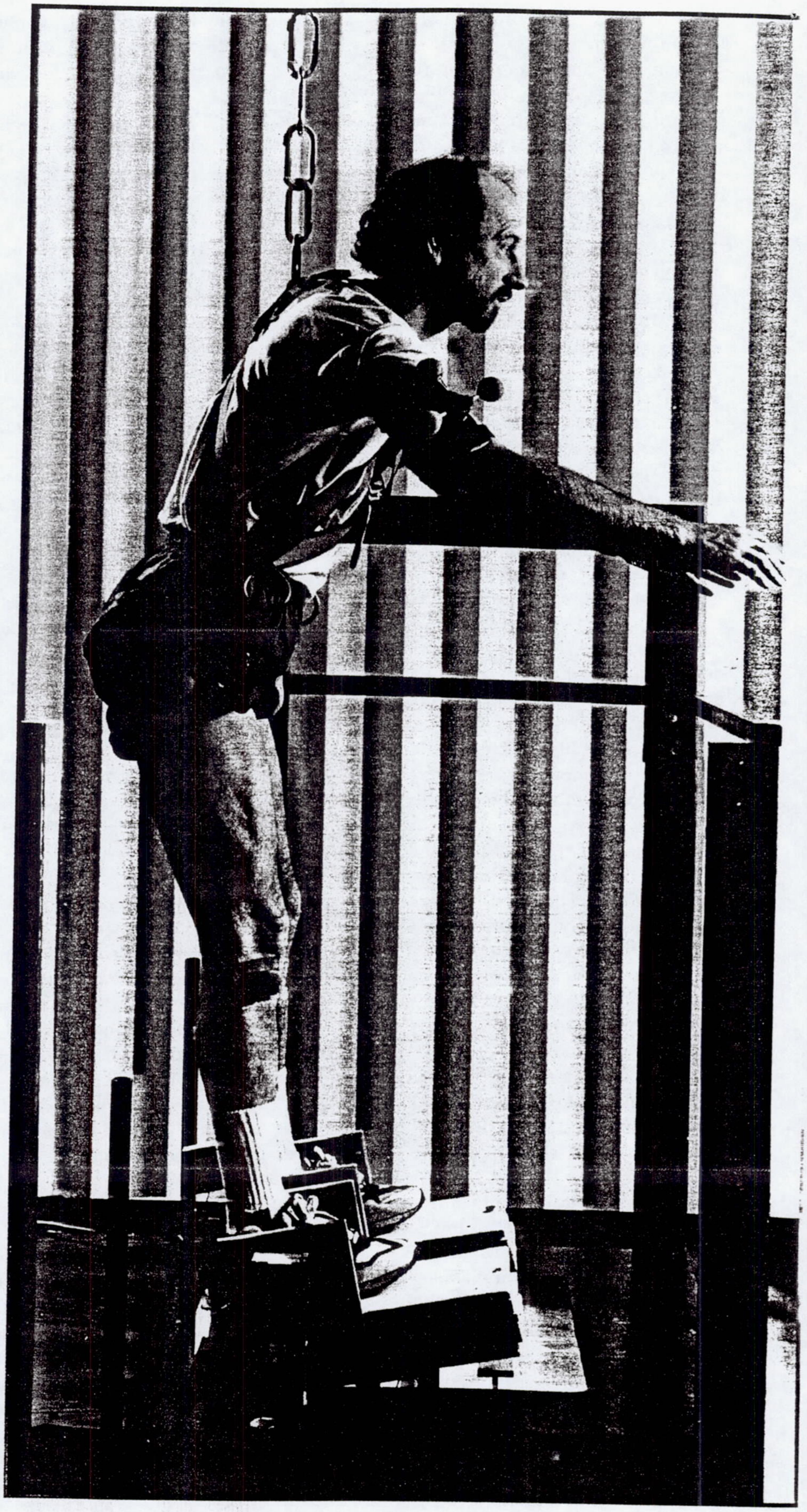
She combines two approaches. The first is experimental, and involves manipulating environmental conditions to separate out the role of each physiologic system. In this context, Cavanagh notes, patients with diabetic neuropathy are valuable test subjects: their lack of feeling can reveal the proprioceptive contribution.

The second approach involves modeling the particulars of balance and posture on the computer.

"This is especially challenging, because mechanically the human body has so many degrees of freedom," Wu says. Her model human has six simplified joints, at the hips, knees, and ankles. Unlike many such models, it simulates movement in three dimensions.

Already, Wu has been able to simulate the amount of torque or force that acts at each joint in the effort to maintain balance, given a perturbation like the platform jerk-back. The amount of torque increases with distance from the body's center: the ankle is the most important joint for balance control. "This is consistent with experimental findings," Wu says, "which confirms that our model is working." Her eventual hope is to use the model to predict falls, showing exactly what types of movements put a body at risk.

For now, though, understanding falls means strapping subjects into the test harness and letting fly. And to that approach, Wu, whose training is in the



Jerked around: Cavanagh models a system built for measuring the body's reaction to a sudden loss of balance.

design of precision instruments, has made a substantial contribution.

As a Ph.D. student at Boston University, she developed a device called the integrated kinematic sensor, or IKS, which combines three different sensors to provide direct readings of three important variables of body movement: orientation, speed, and acceleration. (Other sensors, she explains, rely on measuring one variable and calculating the rest, a procedure which increases the amount of noise and error.)

Its real-time operation makes the IKS especially valuable. Wu has used visual feedback to show neuropathic diabetic patients when, in response to perturbation, their center of gravity was outside the area of their support base — the block formed by the outsides of their feet. ("When this happens," says Wu, "you're going to fall.")

"We're trying to see if this kind of training is helpful in improving postural stability," she says.

Don Streit comes at falling from another angle. Streit, like Wu, is a mechanical engineer. But instead of preventing people from falling, he has focused on trying to lessen their injuries when they do — and on facilitating their recovery from injuries that can't be avoided.

One of Streit's ideas is "soft" flooring.

Streit's floor consists of two thin polyurethane sheets sandwiched around a layer of tiny columns made of the same material. Strong enough to withstand ordinary foot traffic, the floor is designed to "give" against the greater force of someone falling: the columns buckle slightly and then rebound. Computer simulations have shown that the new floor can lessen the force of an impact by 35 percent over conventional flooring. This forgiving quality should significantly reduce the number of debilitating injuries like hip fractures, a common problem among the elderly.

"If you landed on your hip," says Streit, "the flooring would actually tend to wrap around the hip. It conforms to the part of the body that impacts it."

Another of Streit's devices looms just across the lab. It's a large revolving arm, like a crane, whose "bucket" is a seat-like harness. The variable gravity rehabilitation system, he calls it, or, for short, the rehab device. It was designed with hip-fracture repairs and hip-replacement surgeries in mind, as a way to speed the rehabilitation process.

"With these procedures," Cavanagh explains, "rehab is very traditional — it is extremely limited until the patient can walk with the physical therapist's help. Even then, one false move, a slip, and you can do major damage."

With Streit's device, things will be much safer — and easier. Once a patient is secured in the harness, the revolving arm assumes most of the weight. Computer controls allow an attendant to dial in what percentage of body weight he or she wants the patient to bear, increasing the load as the patient gets stronger.

"With this device, a therapist will be able to take a patient the day after surgery — but weighing only 25 percent of their body weight," says Cavanagh. "This is rehab under very controlled conditions."

Hanging from the ceiling nearby is yet another mechanical attempt to defy gravity. This one looks both less substantial and more involved than Streit's crane — a bona fide contraption. It goes by the name of the Penn State Zero Gravity Locomotion Simulator.

Astronauts in zero gravity, Cavanagh explains, lose bone mass very quickly — a kind of osteoporosis that begins within 24 hours of being in space. "If there is no countermeasure, astronauts arriving on Mars after a two-year flight would face a very real possibility of spontaneous broken bones."

The problem, it seems, is lack of activity: specifically, the absence of enough stress on the musculoskeletal system to keep the bones from acquiring the necessary calcium to replace what is lost.

To help counteract this effect, the space shuttle is equipped with an exercise treadmill. But, says Cavanagh, treadmill running in zero gravity may not be giving astronaut bones the jolt they need. "If you look at NASA film of crew members exercising, you can see that they're just running on the balls of their feet without making heel contact."

Nobody really knows just how much stress is needed to stimulate bone production. In order to study the problem, NASA needs an environment of zero gravity — or its equivalent. Unfortunately, that's no easy thing to achieve.

Underwater simulation is one approach that has been tried.

"This is good for some things," Cavanagh says, "but not for exercise. The viscous drag is too much."

Another option is the KC-135, a modified 707 jet that flies Keplerian trajectories — parabolic arcs — over the Gulf of Mexico, producing, for its passengers, weightless interludes of 20 to 30 seconds.

Airplane time, however, is expensive, and the ride can be pretty harsh. Cavanagh, himself a certified pilot, calls his one KC-135 flight "my single worst experience." (The plane was scheduled to make 42 parabolas. "After nine, I lost all interest in science. I lost all interest in *life*.")

There had to be a better way. Sparked by an illustration in an old Russian cosmonaut-training manual, Cavanagh and graduate student Brian Davis built the zero gravity simulator.

The device works by suspending a person horizontally three feet off the floor, using a combination of elastic cords attached by cuffs to the arms legs, torso, chest, and head — "like a marionette," Cavanagh says. As the subject moves, the cords use pulleys to counteract the force of gravity, simulating weightlessness. The subject runs on a treadmill mounted on the wall "below" his or her feet.

Cavanagh and his team hope to use the simulator to measure the effects of weightless exertion, and eventually to design exercises that will provide the necessary stress.

In addition, he says, lessons learned in mock zero gravity may have down-to-earth importance. Exercise could well prove to be a useful adjunct to the standard endocrine supplementation for treatment of "ordinary" osteoporosis.

The idea brings a smile to the clinically-oriented Cavanagh. "Once again," he says, "this is the thing I love. Research reduced into practice that influences people's lives."

Peter R. Cavanagh, Ph.D., is distinguished professor of locomotion studies, biobehavioral health, medicine, and orthopedics in the department of exercise and sport science, and director of the Center for Locomotion Studies, Room 10 Intramural Building, University Park, PA 16802; 814-865-1972. Jan S. Ulbrecht, M.D., is associate professor of biobehavioral health and medicine, and medical director of the diabetes foot clinic at the Nittany Valley Rehabilitation Hospital. Ge Wu, Ph.D., is assistant professor of exercise and sport science, biobehavioral health, and mechanical engineering. Donald Streit, Ph.D., is associate professor of mechanical engineering in the College of Engineering. Gregory Caputo, M.D., is associate professor of medicine at the Hershey Medical Center.

CELOS projects reported above have received funding from the National Institutes of Health, the American Diabetes Association, the Diabetes Research and Education Foundation, the Veterans Administration, the Whitaker Foundation, NASA, and the Centers for Disease Control.

The Pennsylvania State University

The Graduate School

Department of Kinesiology

A BIOMECHANICAL COMPARISON OF 1-G AND FULLY-LOADED SIMULATED ZERO-GRAVITY LOCOMOTION

A Thesis in

Kinesiology

by

Jean L. McCrory

Submitted in Partial Fulfillment
of the Requirements
for the Degree of

Doctor of Philosophy

May 1997

Abstract

Exercise will almost certainly play an integral part in minimizing the bone mineral loss and muscular atrophy that occur during spaceflight. It has been hypothesized that an effective exercise regimen can be developed to elicit loads on the lower extremities and require muscle actions which resemble those encountered on Earth (Cavanagh, 1986; Convertino and Sandler, 1995).

The Penn State Zero-Gravity Simulator (PSZS Davis et al. 1996) is a device which suspends subjects horizontally from multiple latex cords, with each cord negating the weight of a limb segment. A treadmill mounted on the wall under the PSZS enables subjects to run in simulated 0G. Subjects wear a harness to which a number of springs, which provide a gravity replacement load, are connected. The opposite end of each spring is connected to the side of the treadmill. During exercise, astronauts currently wear a similar harness in which the spring tethering load pulls at both the waist and shoulders (Greenisen and Edgerton, 1994).

Ground reaction forces, muscular activations, and joint angles of the left leg during overground, treadmill, and fully-loaded zero-gravity simulated locomotion were assessed in order to gain insight into the effectiveness of the exercise regimen used by NASA to prevent the muscular atrophy and bone demineralization which occur in weightlessness.

There were three hypotheses to this research. It was hypothesized that there will be no differences in peak ground reaction forces and peak loading rates between overground gait and gait in the full bodyweight loaded conditions in the ZLS. A second hypothesis was that that there will be no differences in hip, knee, and ankle joint positions between walking or running overground, on a standard treadmill, and in full bodyweight loaded

conditions in the ZLS. The third hypothesis was that the muscular activations, as a percentage of maximal voluntary contraction, will be similar between walking or running overground, on a standard treadmill, and in full body-weight loaded conditions in the ZLS.

Methods

Sixteen individuals (age 22.9±6.9 yrs, height 178.1±6.68 cm, and mass 72.8±5.8 kg) were studied at two speeds (1.35 m/s walking and 2.68 m/s running). Data were collected during overground locomotion, standard treadmill locomotion, ZLS "shoulder springs only" locomotion, and ZLS "waist and shoulder springs" locomotion. Ground reaction forces were assessed using a force plate mounted within the ZLS treadmill belt and another mounted in the laboratory floor. Angles of each subject's left ankle, knee, and hip were measured with electrogoniometers. Electromyographic data were collected of each subject's left tibialis anterior, gastrocnemius, rectus femoris, vastus lateralis, biceps femoris, and gluteus maximus muscles. Spring tensions in the Subject Load Devices (SLDs) were measured in the ZLS conditions using load cells mounted at the spring attachment sites on the treadmill.

Results and Discussion

The motion of the ankles and hips of subjects in the ZLS were representative of those observed in 1G locomotion. However, the knee was significantly more flexed in the ZLS conditions than in 1G. Maximum knee extension in stance was -0.84±1.85° in overground walking (negative value indicates knee extension), -2.44±1.31° in treadmill walking, 3.23±1.32° in "shoulder springs only" walking, 1.76±1.14° in "waist and shoulder spring" walking, 6.95±1.75° in overground running, 11.44±0.98° in "shoulder springs only" running, and 12.01±1.00° in "waist and shoulder springs" running.

This greater degree of knee flexion was an attempt either to reduce the discomfort felt by the subjects at the shoulders and hips due to the tethering load, or to provide the subjects with a longer flight time, which was shorter in the ZLS.

The large magnitude of the fluctuation of subject load was the most notable finding in the study of the spring tension. The fluctuations were as follows: 13.52±1.70%BW in "shoulder springs only" walking, 21.97±1.49%BW in "waist and shoulder springs" walking, 17.89±1.25%BW in "shoulder springs only" running, and 36.83±1.30%BW in "waist and shoulder springs" running. The differences between the "shoulder springs only" condition and the "waist and shoulder springs" condition can be attributed to the fact that twice as many springs were used in the latter condition. The fluctuations during walking were predictable based upon normal 1G walking center of mass (COM) oscillations (Cavagna et al., 1963). However, the fluctuations during running were less than predicted based upon 1G COM oscillations measured by Morgan et al. (1990). The COM oscillations during running were less in the ZLS than in the study by Morgan et al. (1990) because the subject load, or the "apparent gravity" was greater than 1G during the flight phase of the ZLS trials because the springs were more stretched during flight that during standing. In the ZLS trials, the peak to peak oscillation of the COM was only 6.4cm, not the 10 cm measured in 1G by Morgan et al. Because of the decreased flight phase duration and the fact that subjects ran with a greater degree of knee flexion during stance in the ZLS than in 1G, the average subject load was 96.37±1.59%BW in "shoulder springs only" running and 88.95±1.66%BW in "waist and shoulder springs" running.

The maximum active ground reaction force peaks were significantly larger in the overground conditions than in their ZLS counterparts. The

active force peak values were as follows: 124.37±7.10%BW in overground walking, 87.84±5.23%BW in "shoulder springs only" walking, 81.14±4.51%BW in "waist and shoulder springs" walking, 240.61±7.04%BW in overground running, 180.04±3.77%BW in "shoulder springs only" running, and 159.75±3.97%BW in "waist and shoulder springs" running. Two distinct mechanisms may be responsible for these results: the increased amount of knee flexion during stance in the ZLS and the fluctuation of subject load due to the normal mechanics of running. McMahon et al. (1980) noted lower maximum active force peaks when the subjects in their study were told to "groucho run", or run with an increased amount of knee flexion. Also, when the ground reaction forces of this study are normalized to subject load instead of body weight, the forces in all conditions appear remarkably similar.

The loading rate was significantly different between conditions ($p<0.05$), with overground running resulting in a lower loading rate, at 40.60 ± 2.85 % BW/sec, than the ZLS conditions, at 51.95 ± 1.53 for the "shoulder springs only" condition and 51.81 ± 1.57 for "waist and shoulder springs" condition.

The magnitude of the passive force peak was not significantly different between conditions, but the passive force peak occurred earlier in the running cycle in the ZLS conditions than in the overground condition (15.01±1.11% of cycle overground, 10.64±0.59% of cycle "shoulder springs only", and 10.07±0.60% of cycle "waist and shoulder springs"). The flight phase impulse (flight time * gravity or the subject load) for the overground data was an average of 93.2% of the flight impulse obtained from the "shoulder springs only" condition and only 83.2% of the flight impulse from the "waist and shoulder springs" condition. The impulse-momentum relationship states that the product of force and time equals product of mass and change in

velocity. Therefore, if the impulse was greater in one condition for a subject with a given mass, the impact velocity would be proportionately greater.

The tibialis anterior, rectus femoris, and vastus lateralis were significantly more activated, in terms of their activation integrals, in the ZLS than in the 1G conditions. The gastrocnemius, biceps femoris, and gluteus maximus produced the same levels of activation during all four conditions.

Conclusion

The influence of the gravity replacement load on the perceived comfort, knee kinematics, and ground reaction force variables was the most important finding of this study. This influence was present in several different aspects: the mechanical aspects of load fluctuation and the psychological aspects of discomfort or heightened consciousness of the load.

The load fluctuation had a dramatic effect on the ground reaction forces. The effects of the load fluctuation were stated above, but the desirability of this large fluctuation will remain to be determined until after it is known whether large forces or large loading rates are more important in the maintenance of bone density.

The discomfort of the subject load affected subjects such that they bent their knees to reduce this load. In order to replicate the knee kinematics of overground locomotion, particular attention should be paid to harness design. Maximization of harness comfort is of utmost importance if subjects are to run for extended bouts of exercise over a period of several months.

While the ground reaction force results were shown to be relatively ambiguous in terms of maintenance of bone mass, the electromyographic data were much more encouraging, especially in light of the fact that the health of the muscles also will influence the density of the bone (Currey,

1984). The activation integrals of the tibialis anterior, rectus femoris, and vastus lateralis were greater in the zero-gravity simulated conditions, while there were no significant differences between any of the conditions with regard to the activations of the gastrocnemius, biceps femoris, and gluteus maximus. This was surprising considering that while the subject in the ZLS was under a full body weight gravitational load, the actual limb segments were unweighted. Subjects thus did not have to resist gravity in each of the body segments as they raised their legs off of the surface to run; rather, it was the stance leg had to resist the pull of the springs. Nevertheless, a resulting increase in eccentric contractions was not noted in the ZLS conditions. It therefore must be concluded that, within the limitations of extrapolating to force from EMG patterns, fully-loaded tethered treadmill locomotion for sufficient duration could be a satisfactory countermeasure to both bone loss and muscular atrophy during long duration space flight. Issues of the duration of exercise for effective countermeasures remain to be answered and are outside the scope of this thesis.