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

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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.

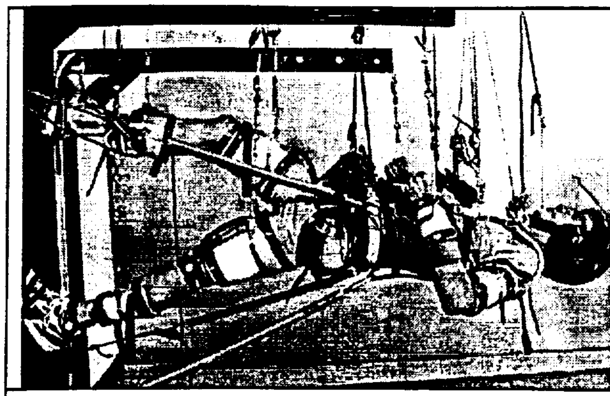

Figure 1: A subject walking in the zero-Gravity Locomotion simulator which has a vertically mounted force measuring treadmill.

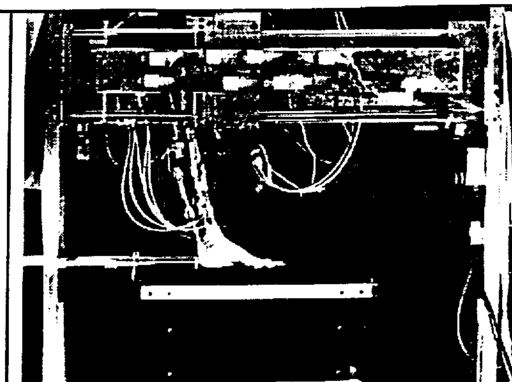

Figure 2: The cadaver simulator showing the muscle forces being applied as the limb moves through the appropriate kinematic trajectory.

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.

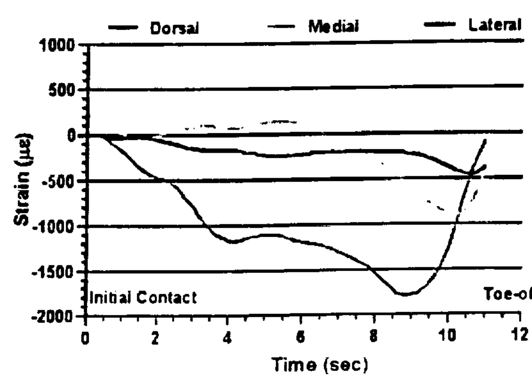

Figure 4: Metatarsal strains measured during simulated 1-G walking in 1/20th real time speed.

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

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