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## Biomechanical Perspectives on Locomotion in Null Gravity

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### 1. Introduction

The current interest in locomotor activities in space is motivated, in part, by similar imperatives that fueled the terrestrial "running boom" of the 1970's. Running and walking are multidimensional forms of exercise that provide muscular, cardiovascular, and psychological benefits - all of which are essential to the well-being and optimum performance of astronauts. But these benefits could be obtained by a number of exercise modalities. What makes locomotion in space so attractive is the possibility that applying one-g locomotorlike forces to the lower extremity during space flight will reverse the losses of bone mineral that are of such great concern in the planning of long-duration missions in the Space Station or on interplanetary voyages (ref. 1). If such a preventative role for locomotion in disuse osteoporosis is indeed confirmed, then simulated one-g locomotion could indeed be the complete exercise for the astronaut.

In this paper, some of the biomechanical factors that must be considered in the study of locomotion in a zero-g or reduced-gravity environment are examined. The overall purposes are to achieve a description of those aspects of one-g locomotion that may be relevant to an understanding of the problem and to suggest experimental models. Comments will also be made on certain biomechanical aspects of cycling since it is also a candidate for use as an in-flight countermeasure against the various deconditioning effects that occur.

### 2. Biomechanical Hypotheses for Bone Demineralization

Although the discussion in this article is focused on biomechanical factors, this emphasis is not intended to imply that these are the primary etiological determinants of bone demineralization. It is likely that bone demineralization during

weightlessness is a multifactorial problem with endocrine (ref. 2), nutritional (ref. 3), neuromuscular (ref. 4), biomechanical, and other factors interacting to produce the changes that have been observed (refs. 5 to 7). It is accurate to say, however, that of these various possible causes, least attention has been focused on those of a biomechanical nature.

Although we may not realize it, life in a one-g environment is characterized by a series of collisions (ref. 8). Each time the foot hits the ground in walking or running, shock transients are experienced by the lower extremity. As we shall discuss later, these shocks can be measured by accelerometers attached to the lower extremity either by Steinman pins (refs. 9 and 10), by surface mounting (ref. 11), or by attaching accelerometers to the shoe (ref. 12). Various experiments have shown that transients as great as 40g may be experienced at the shoe in running (refs. 12 and 13) and as great as 8g at the tibia in walking (ref. 13).

Although no measurements of lower extremity accelerations have yet been made in space, it is reasonable to suppose that the orbital transients will be much less than those on Earth. We know from observation of in-flight films and from anecdotal reports from astronauts that the upper extremities become the main locomotor organs in space. The body is set into motion by arm forces, and when the destination in the spacecraft is reached, deceleration is again performed by the arms. The legs are simply used to "perch." Thus, the absence of skeletal transients is one "functional" theory to explain bone demineralization.

Skeletal transients are not, however, the only mechanical consequences of locomotor exercise on Earth which are absent in space. It is quite probable that the muscles of the lower extremity are only called upon to generate a fraction of the forces that they routinely exert on Earth. The skeletal implications of this state are that the tensile stresses at muscle attachments and the compressive stresses and bending moments that develop as result of muscular

forces are also absent. Thus, a second biomechanical hypothesis for the loss in bone mineral is that the absence of normal internal stresses in the bone due to reduced muscle forces is responsible. As a perspective on the testing of these hypotheses under a variety of conditions, methods for the study of locomotor activities and results from terrestrial experiments are now discussed.

### 3. Biomechanical Studies of Terrestrial Locomotion

If a force platform is interposed between the foot and the floor during running and walking, the typical results shown in figures 1(a) and 1(c), respectively, can be obtained (refs. 14 and 15). These diagrams show the vertical component of force in the two activities, both of which are characterized by sharply rising initial peaks resulting in the skeletal transients mentioned earlier. The more slowly rising peaks later during the contact phase are larger in both activities, reaching approximately 1.2 times body weight (BW) in walking at 1.5 m/s and 2.5 to 3.0 times body weight in running at 3.8 m/s. Nigg (ref. 16) has described the two distinct peaks in the running curve as the "impact" peak and the "active" peak, respectively.

Our own studies have shown that running technique can drastically affect the nature of ground reaction forces (ref. 14). Individuals who strike the ground with the rearfoot display patterns similar to those shown in figure 1(a), but in a runner who makes first contact with the midfoot or the forefoot, the "impact" peak tends to be diminished or absent and the "active" peak tends to be larger (fig. 1(b)). This result has obvious relevance to the design of in-flight exercise since, if the transients are necessary, care should be taken to design the exercise system such that the astronaut cannot avoid heel contact.

It is instructive at this point to make a comparison of the reaction forces experienced by the foot during cycling (ref. 17) with those just described for running and walking. A force-measuring pedal has been designed and built in our laboratory, and typical results from a recreational cyclist pedaling at 90 rpm with a power output of 130 watts (about 50 percent of his maximum) are shown in figure 1(d) (from ref. 18). It is clear from figure 1 that the forces during cycling are different from those during walking and running in two important ways. First, during all phases of the pedaling cycle, the rate of change of force in cycling, the  $dF/dt$ , is considerably smaller than that in walking or running. There is no rapidly rising force analogous to the initial transients

seen in figures 1(a) and 1(c) during foot-ground contact. Second, the absolute magnitude of the forces in cycling are small - approximately 3 and 6 times smaller than those in walking and running, respectively.

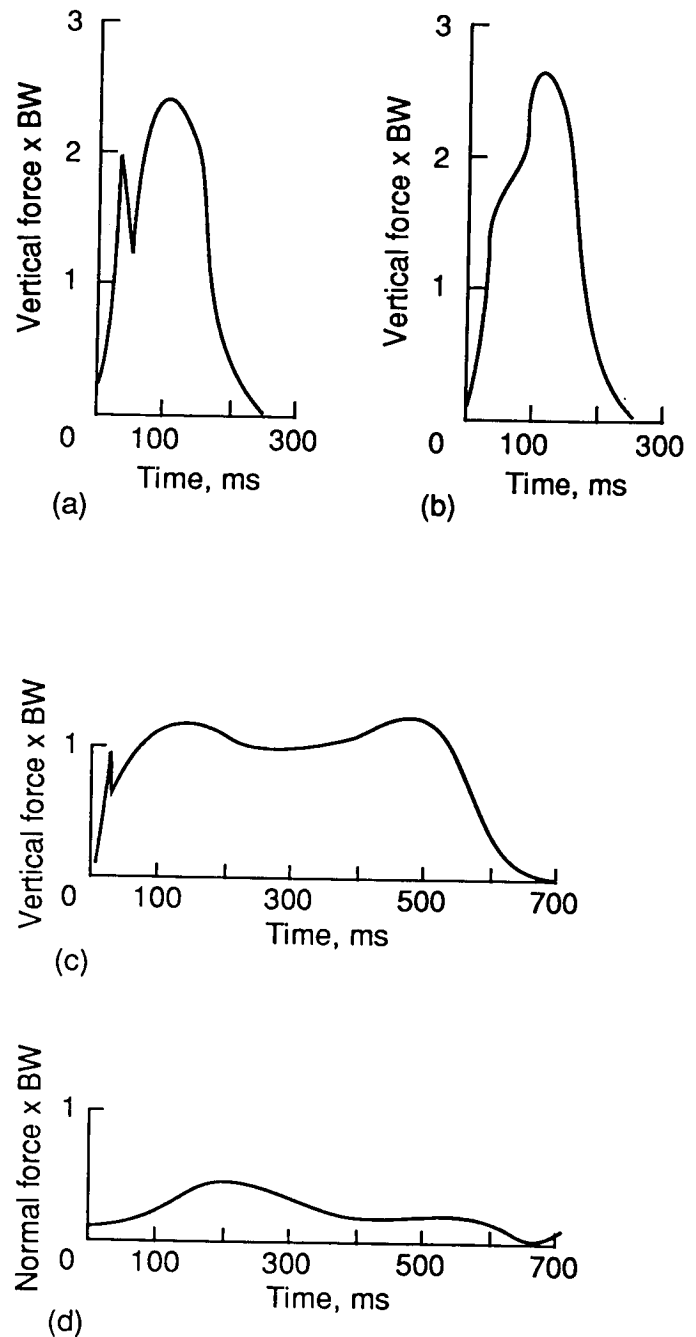


Fig. 1 - Force-time curves for various modes of exercise. Note the brief early transients in figures 1(a) and 1(c) and the increasing peak forces and rates of change of force as the progression from cycling to walking to running occurs. (a) Rearfoot running at 3.8 m/s. (b) Forefoot running at 3.8 m/s. (c) Walking at moderate speed. (d) Cycling at 90 rpm and 130 watts.

The absence of transients is, of course, due to the fact that the body weight is supported during cycling and not used as an inertial mass which collides with the supporting surface. The smaller "active" forces in cycling are also related to this fact, but, at a muscular level, there are important differences which must be considered in the design of exercise protocols in space. The period of weight acceptance or cushioning in both walking and running is characterized by eccentric muscle action that is immediately followed by concentric action as the body is propelled upward and forward. Such a "stretch-shorten" cycle (ref. 19) is entirely absent in cycling. It is well known (ref. 19) that the largest muscular forces can be generated during eccentric action. Thus, even at high power outputs and low pedal rates, lower extremity muscle forces in cycling will never approach those of walking or running. Future studies may well demonstrate the absence of, or reduction in, the number of eccentric muscle actions to be a primary difference between terrestrial and reduced-gravity locomotion.

Once ground reaction forces have been determined, the addition of kinematic data and body segment parameter information enables some first-order estimates of the bone-on-bone articular forces to be made (refs. 20 and 21). Although such estimates have large error bounds, it has been estimated that forces at the talocrural joint during slow running may exceed 12 times body weight (ref. 22). Thus, forces of this magnitude could exist between the tibia and the head of the talus of a 180-pound individual. This result emphasizes the importance of large muscular forces which are principally responsible for the bone-on-bone forces in the joint being approximately 4 times greater than are the ground reaction forces at the foot.

#### 4.1 Studies of Locomotion in Zero g and Reduced Gravity

##### Modeling

The tremendous advantage of modeling in the present context is that gravity can be removed by the stroke of a pen. One does not need orbital experiments or brief moments of weightlessness during aircraft flight to test the hypotheses. All that is needed is to set a single variable to zero in the model. Unfortunately, the complexity of most biomechanical models of locomotion cannot approach that of the intact human locomotor system (ref. 8); this tends to limit their "ecological validity." There have been

some successes, however. Kane and Scher (ref. 23) predicted arm and leg movements that would generate self-rotation in a weightless environment using linked rigid-body models and Lagrangian mechanics. In 1964, Margaria and his coworker (refs. 24 and 25) correctly predicted that a "bounding" gait would be appropriate for the lunar environment and pointed out on the basis of ground reaction forces and frictional considerations that the maximum speeds for lunar running would be 1.7 m/s and 3.4 m/s on loose and firm terrain, respectively. Margaria's predictions were largely confirmed by the subsequent locomotor experience of astronauts on the lunar surface.

Despite the successes mentioned previously, there are two areas critical to the current topic that have not been well served by biomechanical models. These are the consequences of impacts to the skeleton (ref. 26) and the solution of individual forces in lower extremity muscles (ref. 22). The implication of both of these shortcomings is that, in the near future, direct experimentation is more likely than modeling to lead to operationally significant results.

#### 4.2 Direct Experimentation

A passive tethered treadmill has already been flown on most Space Shuttle missions since STS-3. The device, shown schematically in figure 2(a), incorporated elastic bungee cords attached both to a harness and to the treadmill to accelerate the astronaut back to the treadmill bed after pushoff (ref. 27). The only biomechanical information available from treadmill running during these missions is lower extremity kinetic data that are currently being obtained from the analysis of short clips of 24-frame/s 16-millimeter film taken with a wide-angle lens (ref. 28). Other possible in-flight data that could be collected from Space Shuttle or Space Station missions in the future are shown in figure 2(b).

Tibial transients could be monitored by surface-mounted accelerometers, whereas accelerometers mounted on a bite bar or a helmet could detect cranial accelerations. The mounting of the treadmill to the deck via force transducers should be explored, although this method may not be practical because of storage requirements. A more satisfactory solution may be to instrument the footwear of astronauts with pressure-sensitive insoles or with inertia switches. Figure 3 shows plantar pressure distribution obtained on Earth between the bare foot of a running subject and the ground (ref. 29). If a similar technique could be developed for in-shoe

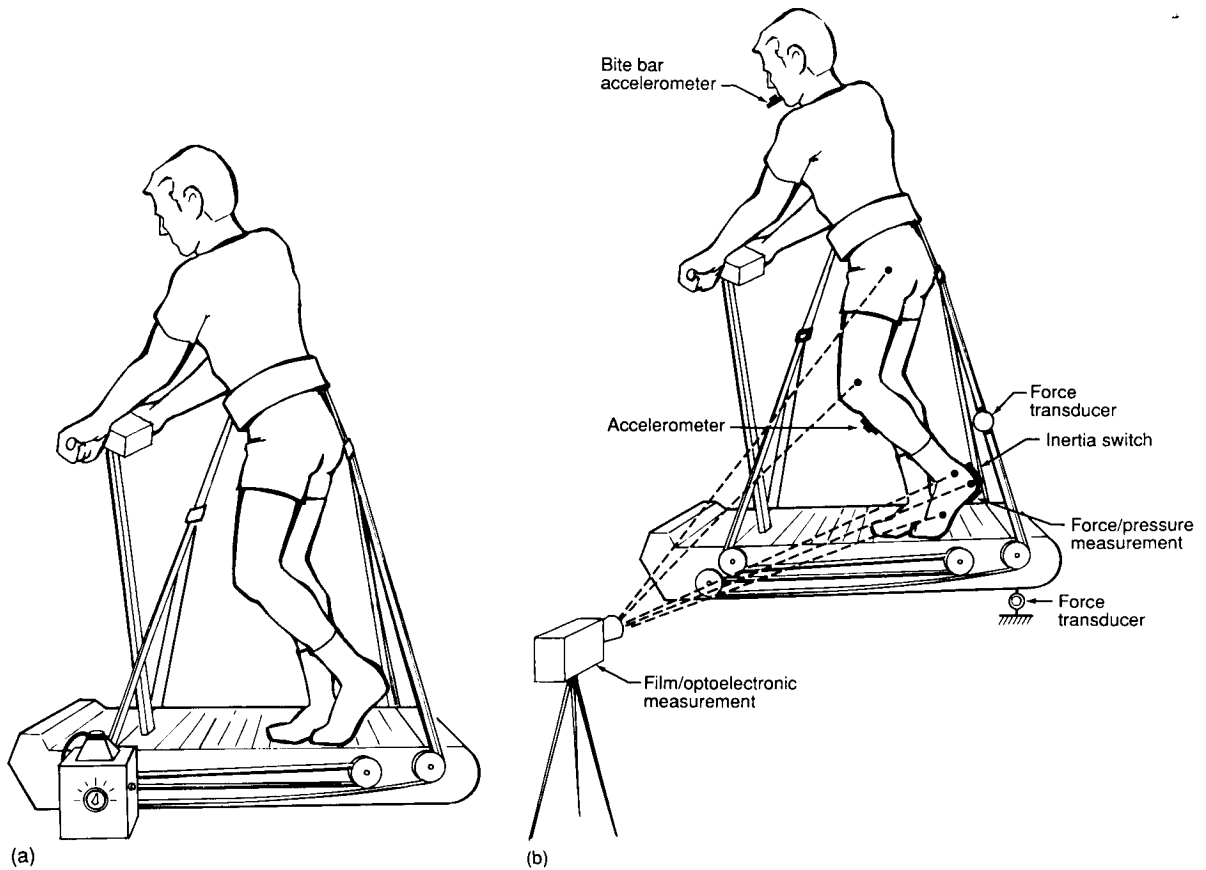


Fig. 2.- Use of treadmill under weightless conditions. (a) Schematic diagram of passive tethered treadmill used on Space Shuttle missions. (b) Possible methods for monitoring the biomechanics of locomotion in space. See text for further details.

monitoring (ref. 30), it would offer the possibility of collecting a complete history of lower extremity loading during typical activities in space.

A simpler, though less complete, technique might involve the development of a battery of inertia switches. These switches, typically used in emergency locator transmitters for aircraft and in weapons applications, can be designed to close at a predetermined acceleration. Thus, an array of shoe-mounted switches with thresholds of, for example, 5g to 20g and associated accumulating registers could collect information on the number of transients above certain levels experienced by the lower extremity during flight. Kinematic data from film, video, or other optoelectronic devices (ref. 31) could also be collected using instrumentation capable of fulfilling several other purposes during the mission. Any of these techniques could also be used during aircraft flights that offer brief periods of weightlessness.

### 4.3 Ground-Based Experiments

There are two major reasons why ground-based experiments should be pursued in the near future. First, flight experiments are extremely expensive to conduct and involve considerable lagtime between planning and the availability of data. A second and more urgent consideration is that ground-based experiments are needed to provide design information for exercise devices to be built in the Space Station. Although occurrence of the first Space Station mission is not anticipated until the late 1990's, the basic design requirements for in-flight exercise equipment must be finalized soon.

The principles of ground-based zero-g locomotion simulation devices have already been elucidated (ref. 32), and it appears that such a device has been used in the U.S.S.R. for ground-based experiments. A typical system, shown in figure 4,

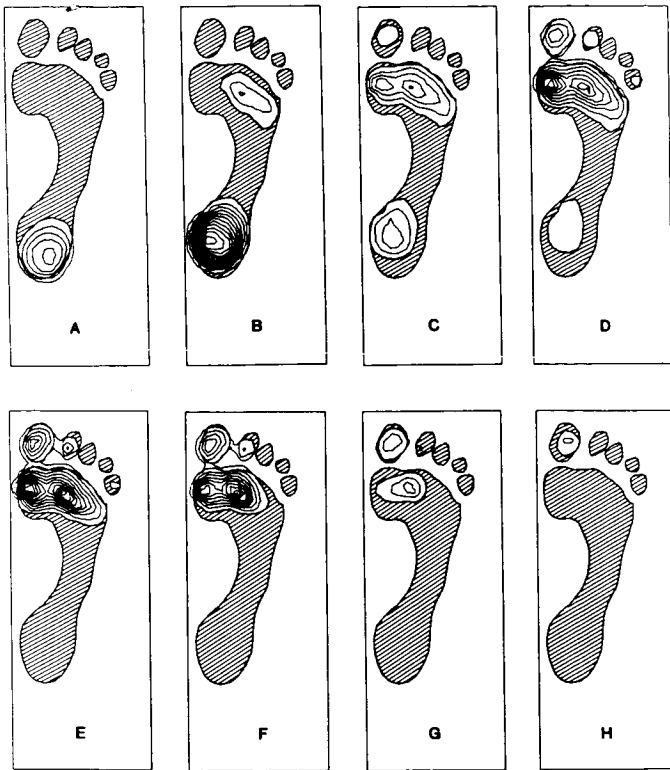


Fig. 3.- Plantar pressure distribution at eight instants of time during foot contact in barefoot running. If such measurements could be made under weightless conditions, a complete loading history for the lower extremity could be developed.

would involve supporting the trunk in either a prone or a side-lying position with the weight of the head, the trunk, and the arms totally supported by stiff suspension cables. Part of the weight of the lower extremities could be supported by compliant cables. A treadmill would be mounted vertically on the wall via force transducers, and bungee cords attached to a harness would provide the major means of applying axial loads to the lower limbs. The subject could be instrumented using any of the methods described previously.

This kind of experimental arrangement could provide answers to important questions regarding locomotion in reduced-gravity situations. For example, can passive elastic restraints generate sufficient forces to apply one-g locomotor forces to the lower extremities? What influence does the technique of the subject have on the forces and accelerations experienced by the lower limb? What effect does equipment modification have on the locomotor pattern? Once these and other questions are answered, the device could then be incorporated into bed-rest studies so that the effectiveness of quantifiable locomotor exercise as a countermeasure to bone demineralization could be investigated.

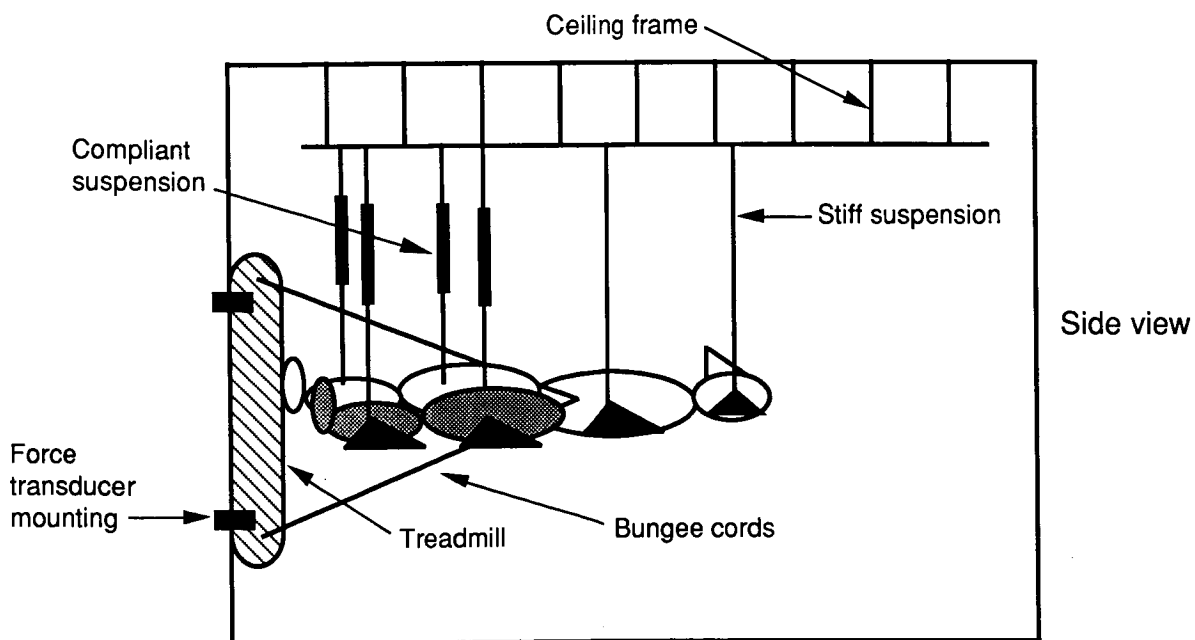


Fig. 4.- Schematic diagram of a "vertical" tethered treadmill that would enable study of the mechanics of zero-g locomotion. See text for further details.

## 5. Summary and Concluding Remarks

A number of important features of various locomotor activities have been discussed in this paper, and approaches to the study of these activities in the context of space flight have been suggested. In particular, the magnitude of peak forces and the rates of change of force during terrestrial cycling, walking, and running were compared. It was shown that subtle changes in the conditions and techniques of locomotion can have a major influence on the biomechanical consequences to the skeleton.

The various hypotheses that identify locomotor exercise as a countermeasure to bone demineralization during weightlessness deserve to be tested with some degree of biomechanical rigor. Various approaches for achieving such scrutiny have been discussed.

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