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Effect of Load Levels of Subject Loading Device on Gait, Ground Reaction Force, and Kinematics during Human Treadmill Locomotion in a Weightless Environment

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1 OPERATIONAL RELEVANCE OF FINDINGS (SUMMARY)

The findings of this study are directly relevant to exercise prescriptions and the configuration of subject loading hardware for exercise sessions with the Treadmill with Vibration Isolation and Stabilization (TVIS) on the International Space Station. In particular, the following recommendations can be made:

- Loading prescriptions or crew procedures should compensate for the decrease in external load between static (standing) and dynamic (walking or running) conditions.
    - When extender straps are used, the Subject Loading Device (SLD) load setting that is input to the TVIS control panel should be increased by approximately 7% to compensate for the average drop in load.
    - When bungees are used, the SLD load setting that is input to the TVIS control panel should be increased by approximately 9% to compensate for the average drop in load.
- 1G-like peak ground reaction forces are obtained during walking (3 mph) using an SLD setting of 100% of body weight. During running (7 mph), however, 1G-like peak ground reaction forces may only be achieved with SLD settings of about 120% of body weight. Consideration should be given to prescribing SLD load settings of greater than 100% of body weight, up to a maximum of 120% of body weight, to achieve 1G-like peak ground reaction forces.
- The recommendations above are additive. For instance, for running at 7 mph, consideration should be given to prescribing an SLD load setting as high as 127% of body weight when the subject is using extender straps, or 129% of body weight when the subject is using bungees, to achieve 1G-like peak ground reaction forces.
- Whenever possible, bungees should be used in line with the SLDs because they reduce load stiffness by a factor of 5 and reduce the asymmetry of the external load.
- These recommendations should be considered within the context of crew comfort and safety. Because the crewmember loading harness does not distribute load throughout the entire body as it would at 1G, crewmembers might find that loading in excess of 100% of body weight creates uncomfortable pressure on their shoulders and hips. This effect may be minimized by training crewmembers to optimally configure their harness. Furthermore, a gradual ramp up to the higher load settings during the course of a mission will allow the crewmember’s body to become accustomed to the localized pressure applied by the harness. If comfort and safety concerns prevent the use of load levels greater than 100% of body weight, consideration should be given to making up for the reduced bone and muscle loading through the use of tailored resistance exercise regimens.

2 GOAL

The purpose of this experiment was to determine the influence, on the associated gait, ground reaction force, and kinematics, of applying different levels of harness loading to subjects while they are utilizing the Treadmill with Vibration Isolation and Stabilization (TVIS) Subject Load Device (SLD) during locomotion in weightlessness.

3 INTRODUCTION

Prolonged exposure to weightlessness associated with space flight provokes profound physiological changes in humans. Two areas of significant concern are bone loss and neuromuscular deconditioning. Bone mineral density losses of 1-2% per month in critical weight-bearing areas, such as the proximal femur, during long-term space flight (Grigoriev et al., 1998) create an increased risk of bone fracture. Changes in lower-limb neuromuscular activation patterns (Layne et al., 1997) have raised concern about the ability of astronauts to escape from their spacecraft in an emergency landing situation. Countermeasures that stimulate bone maintenance (Turner, 1998) and locomotor control (Reschke, 1998) are vital to the success of human space flight.

It has been hypothesized that, to help maintain the bone mineral density that an astronaut has in normal gravity (1G), the ground reaction force (GRF) achieved during locomotion in weightlessness (0G) should mimic the GRF that occurs on Earth (Davis, 1996). Specifically, the peak ground reaction force (pGRF) and loading rate obtained at 0G should be comparable to the values achieved at 1G. Peak GRF and loading rate during locomotion affect the strain magnitude and strain rate applied to the bones, which are thought to affect osteogenesis (Rubin et al., 1984). McCrory et al. (2002) found that pGRF during horizontal suspension running was directly related to the load applied to the subject.

The TVIS with integrated SLD, which is currently in use on board the International Space Station (ISS), is intended as a countermeasure to bone loss and neuromuscular deconditioning. A crewmember exercising on the treadmill receives an external load by means of an upper-body harness that attaches to the SLD via extender straps or bungees. GRF is imparted as the SLD pulls the crewmember toward the treadmill surface during locomotion. The SLD utilizes a mechanical spring-loaded cable and pulley system to transfer specific loads to the crewmember via an upper-body harness. It is not known how this type of loading affects the gait, GRF, and kinematics during running on board the ISS.

The primary objective of this investigation was to determine if gait, GRF, and kinematics during locomotion at 0G, using a simulated ISS treadmill and SLD, are significantly different from those found during locomotion at 1G. The secondary objective of this investigation was to quantify the load magnitude, variability, and stiffness of the external load (EL) applied to the crewmember by the SLD during locomotion at 0G.

4 METHODS AND MATERIALS

Four subjects (3 men, 1 woman, 28.0 ± 5.6 years, 170.8 ± 9.1 cm, 75.1 ± 11.2 kg) ran at 3, 5, and 7 mph (4.8, 8.0, and 11.2 km/h) during weightlessness (0G) on board the KC-135 aircraft (Figure 1) and on the ground (1G). Before the study began, all subjects passed a modified Air Force Class III physical exam and signed an informed consent document. This study was approved by the Johnson Space Center Committee for the Protection of Human Subjects.

A treadmill instrumented with force plates (Kistler Gaitway, Amherst, NY) was used in place of the TVIS so that vertical ground reaction force could be measured accurately. The restricted size of the ISS treadmill running surface was simulated by placing over the treadmill running surface a ¼-inch-thick aluminum plate with a cutout equal in size to the TVIS running surface (Figure 1 and Figure 2). On the actual ISS treadmill, the SLD units are mounted so that their top surface is flush with the running surface and the load cable extends vertically. It was not possible to do this with the treadmill used for this experiment because the running surface is wider than on the ISS treadmill. Instead, the SLD units were mounted on either side of the cover plate by means of ½-inch–thick plates that extended out to the sides (Figure 1 and Figure 2). The load cable emerged horizontally from each SLD unit and was guided through pulleys mounted on each side of the cover plate cutout so that the cable would then emerge vertically at the point where it would normally exit the SLD on the ISS treadmill. Two harnesses were used: a Russian-designed harness and the U.S.-designed harness currently in use on the TVIS. Both harnesses were tested because it is anticipated that both harnesses will be used with the ISS treadmill. Each harness was attached to the SLD by two methods: relatively non-compliant extender straps and compliant bungees.

Figure 1. Treadmill Subject Load Device experiment being performed on NASA KC-135 Research Aircraft.

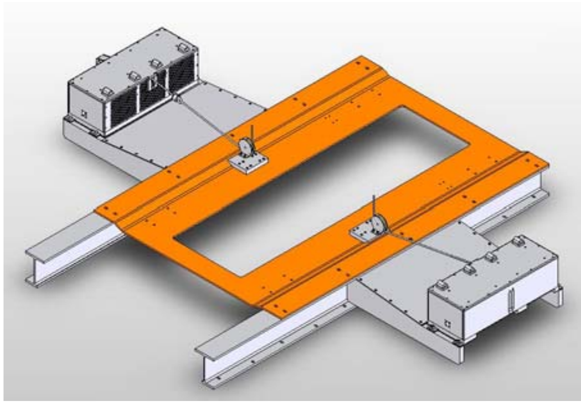

Figure 2. Treadmill cover plate and SLD mounting scheme.

2.1 Load and Displacement

SLD load settings were configured with the goal of loading each subject to 70%, 85%, or 100% of their body weight (BW) while standing, referred to as “Low”, “Medium”, and “Full” load levels. The load settings associated with these levels for a given subject were calculated before they were tested, and the SLD units were configured to apply the appropriate load before each trial (each flight parabola) started. No SLD loading was used during 1G locomotion.

Each SLD cable was attached to a load cell (Entran, Inc., Fairfield, NJ), which was used to measure the applied load at 120 Hz. The load cell was in turn attached to a cable that was fed through a pulley and connected to the subject harness via an extender strap or bungee. The pulley arrangement was used so that the amount of space taken up by the load cell would not significantly reduce the extension of a bungee placed in line with the SLD and thereby lower the amount of load that could be provided to the subject when bungees were used (Figure 3). A linear encoder (Ergotest Technology, Langesund, Norway) was mounted on the right side of the subject and used to measure the linear displacement of the load attachment point at the subject’s harness. Load and displacement data were captured and recorded by a data acquisition system (National Instruments, Austin, TX). An approximate loading stiffness for each trial was calculated by dividing the standard deviation of total load by the standard deviation of displacement. Mean stiffness was then obtained by averaging multiple trials for each speed, external load, and harness attachment condition.

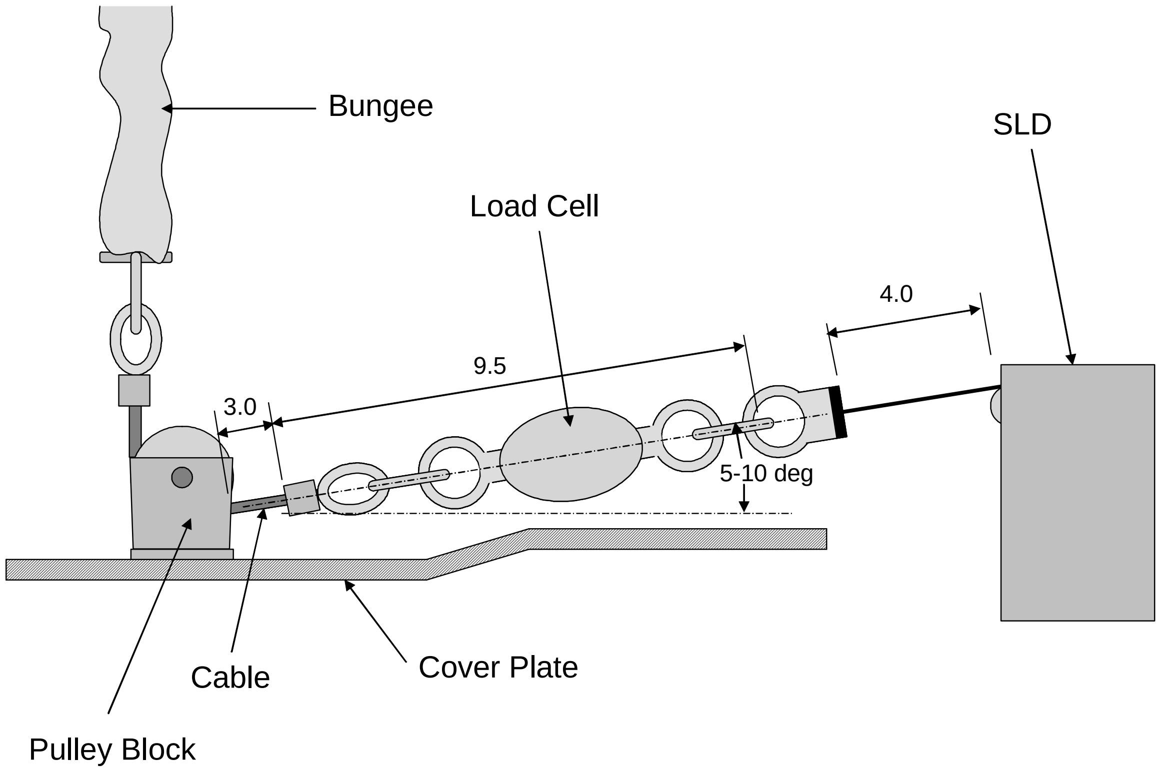

Figure 3. Pulley arrangement used to accommodate load cells and side mounting of SLD units so that a loading arrangement similar to that on the space station treadmill could be achieved.

2.2 Gait Parameters

Gait parameter outcome variables of interest were contact time, stride length, and cadence (stride frequency). Contact time was defined as the duration, in seconds, that the foot was in contact with the treadmill during a footfall. Initial contact was defined as the instant that the GRF rose above 10 N; toe-off, or the instant the foot left the ground, was defined as the instant that the GRF fell below 10 N. Stride length was the actual distance on the treadmill surface that the foot traveled while in contact with the treadmill belt plus the distance that the treadmill belt traveled while the foot was airborne. Cadence was the number of steps that the subject took per minute.

2.3 Ground Reaction Force

Vertical GRF data were collected during 2 trials at each configuration with a force treadmill (Kistler Gaitway, Amherst, NY) at 250 Hz for the 5- and 7-mph trials and 100 Hz for the 3-mph trials for 25 sec. GRF variables (load acceptance rate, peak impact force, peak active force, and impulse) were determined for 8 successive footfalls. Peak impact and peak active force were identified manually for each footfall. Loading rate was found as the peak impact force magnitude divided by the time from initial contact to peak impact force. Impulse was found as the area under the curve of the GRF trajectory. Before any analysis took place, each footfall was low-pass filtered at 40 Hz to remove any motion artifact. For this report, peak active force is considered to be equivalent to peak ground reaction force (pGRF). For each trial, the mean of each variable was computed using all recorded footfalls. Any partial footfall measurements were eliminated from the analysis.

Two harnesses were tested: a Russian harness and the U.S. harness in use on the ISS at the time. An analysis was performed to determine what effect, if any, the harness type had on ground reaction force (GRF) outcome variables during running in weightlessness. The results of this analysis were used to determine if GRF data trials for the two harness types could be pooled

2.4 Kinematics

Motion data were collected with a six-camera infra-red video motion capture system (eMotion, Inc., Milan, Italy). Reflective markers were placed on the left side of the subject with the assumption that leg motion was symmetrical. Markers were placed to approximate the joint centers of the hip, knee, ankle, and fifth metatarsal of each subject, and 3-D position data were collected at 120 Hz during 1G trials and at 60 Hz or 120 Hz during 0G trials. Before any data were collected, the activity volume was calibrated to within 1 mm of re-prediction accuracy.

Position data were filtered using a 4^{th}-order Butterworth low-pass filter with a cutoff frequency of 4 Hz, determined after a power analysis of the raw signal data. Knee angle was quantified as the angle separating the vector directed from the hip to the knee from the vector directed from the knee to the ankle. Ankle angle was defined as the angle between the vector directed from the knee to the ankle and the vector directed from the ankle to the fifth metatarsal. The investigators conducted a static trial before conducting any motion trials, to define the marker orientation that was associated with each joint in the neutral position; this angle was subtracted from the angle computed during motion trials, which resulted in 0° as the anatomical position. Positive knee angle values reflect flexion and positive ankle angles reflect plantar flexion. Angular velocity was calculated using finite central differences. Positive angular velocities indicate flexor movement whereas negative angular velocities indicate extension movement.

2.5 Data Analysis

Alpha was set at $p < 0.05$ for all comparisons.

1) <u>Load and Displacement</u>: Descriptive statistics were computed for mean total load, total standard deviation, load asymmetry, and stiffness. Statistical analysis was performed using a two-factor (load and speed), one-way analysis of variance (ANOVA) to determine if there were differences in load levels or speeds, or if load and speed interact. Hardware malfunctions and early flight termination on one of the flights caused the number of successful trials for each speed/external load level/external load configuration condition to be variable. Therefore, all trials were pooled within each condition for further analyses.

2) <u>Gait Parameters</u>: Each gait descriptor outcome variable was computed for 8 successive footfalls for each trial. Mean scores were then found as the average of each outcome variable for the entire trial. A single-factor ANOVA was used within each locomotion speed to identify the effects of external load on outcome variables. Any significant

interactions were analyzed with Tukey-Kramer multiple comparison tests to identify significant differences between loads.

3) Ground Reaction Force

   a. Ground Reaction Force Parameters: Analysis was performed for 3-mph and 7-mph locomotion trials at 1G and 0G. Multiple analyses were completed for each variable to determine if data from the multiple conditions (load, harness type, use of extenders vs. bungees) could be pooled. Means for each outcome variable for each locomotion trial were computed from the analysis of 8 footfalls. Differences between speed, loading level, harness, and SLD use with extender straps or bungees were examined using ANOVA or t tests, depending on the number of conditions being compared. The specific test used for each comparison is listed for each result. When an ANOVA was used, any significant interactions were further examined using Tukey-Kramer post hoc t tests.

   b. Prediction of Peak GRF from External Load: During exercise on ISS it is common to prescribe increased EL as the mission continues, in an attempt to place greater loads on the musculoskeletal system. It was of interest to determine if a relationship existed between EL magnitude and peak GRF, and if EL could predict the peak GRF during locomotion. Mean dynamic EL and mean peak GRF for each trial were used in a regression analysis for each speed, to determine if an equation could be developed relating EL to peak GRF. Separate analyses were performed for walking at 3 mph and running at 7 mph. A secondary analysis, pooling the walking and running data, was performed. Slopes and r^2 values were analyzed to determine significance.

4) Kinematics: Only 1G trials at 7 mph and full-load 0G trials with the SLD and extender strap at 7 mph were analyzed. Four strides from two trials under each condition (1G and 0G) were analyzed for each subject. For each outcome variable, repeated measures ANOVAs were used to determine significant differences between trials and gravitational conditions. Among the variables tested were peak instantaneous knee and ankle angular displacement (minimum and maximum), and peak instantaneous knee and ankle angular velocity (flexor and extensor). If the ANOVA showed a significant difference for these variables, Tukey post hoc tests were used to delineate any interactions.

3 RESULTS

3.1 Load and Displacement

3.1.1 Mean Dynamic External Load

The mean dynamic EL was computed as the sum of the mean load for the right and left load cells for each trial (Figure 4). No significant difference was found between speeds, and the interaction of speed and load level was not significant. Tukey-Kramer post hoc tests revealed that all load levels were different from one another (Table 1). These data suggest that the dynamic EL increased with increased static EL, but the increase was not affected by speed.

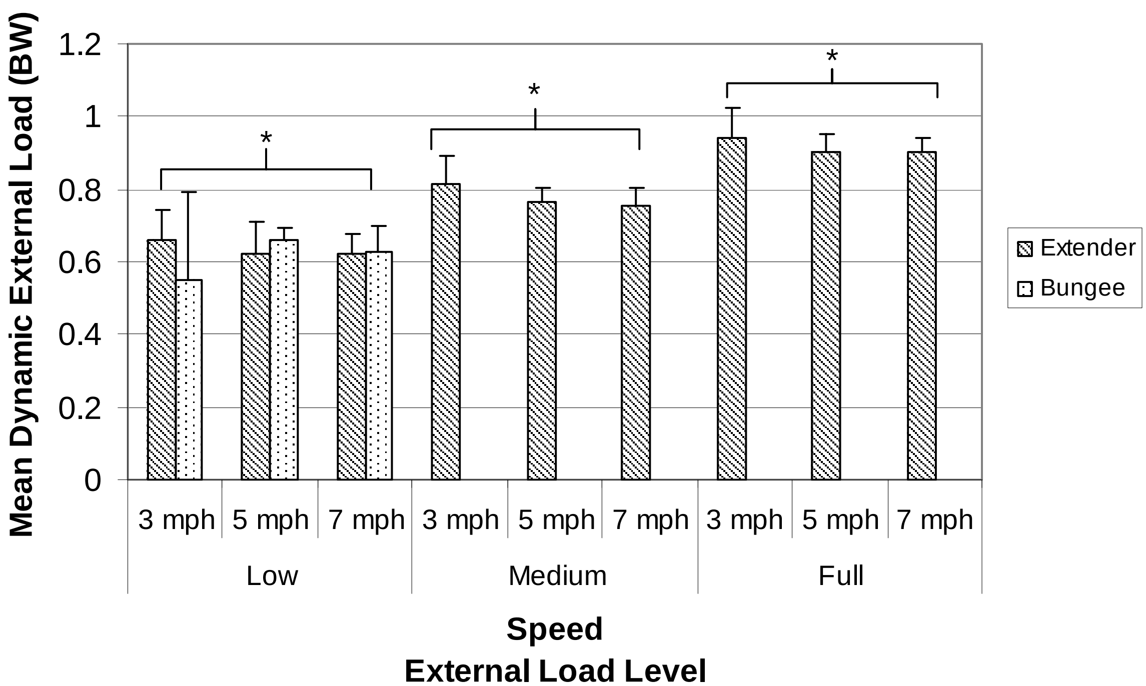

Figure 4. Mean dynamic SLD load for all loading configurations. * Mean dynamic external loads for all load levels were significantly different from each other, $p < 0.05$.

Table 1. Mean (SD) dynamic external load provided by SLD during locomotion.

		Mean (SD) Dynamic External Load (BW)	
Load Level	**Speed (mph)**	**SLD + Bungee Configuration**	**SLD + Extender Strap Configuration**
Low	3	0.55 (0.24)	0.66 (0.09)
	5	0.66 (0.03)	0.62 (0.09)
	7	0.63 (0.07)	0.62 (0.06)
	Mean	0.61 (0.13)	0.64 (0.08)*
Medium	3		0.81 (0.08)
	5		0.76 (0.04)
	7		0.75 (0.05)
	Mean		0.78 (0.06)*
Full	3		0.94 (0.08)
	5		0.90 (0.05)
	7		0.90 (0.04)
	Mean		0.91 (0.06)*

*Mean dynamic external loads for all load levels were significantly different from each other, $p < 0.05$.

3.1.2 External Load Standard Deviation

The standard deviation of the EL provides a measure of how much variance occurs during a trial. At low EL (no bungee data were available for other load levels), configurations using the extender straps seemed to provide a more variable load than those using bungees (Figure 5). No differences were found between mean standard deviations within speeds or within load levels, but the 3-mph trials had a lower standard deviation than the 5-mph and 7-mph trials (Table 2). These data suggest that as locomotion speed increases the variance in EL increases, and that extender straps may provide a more variable load than bungees.

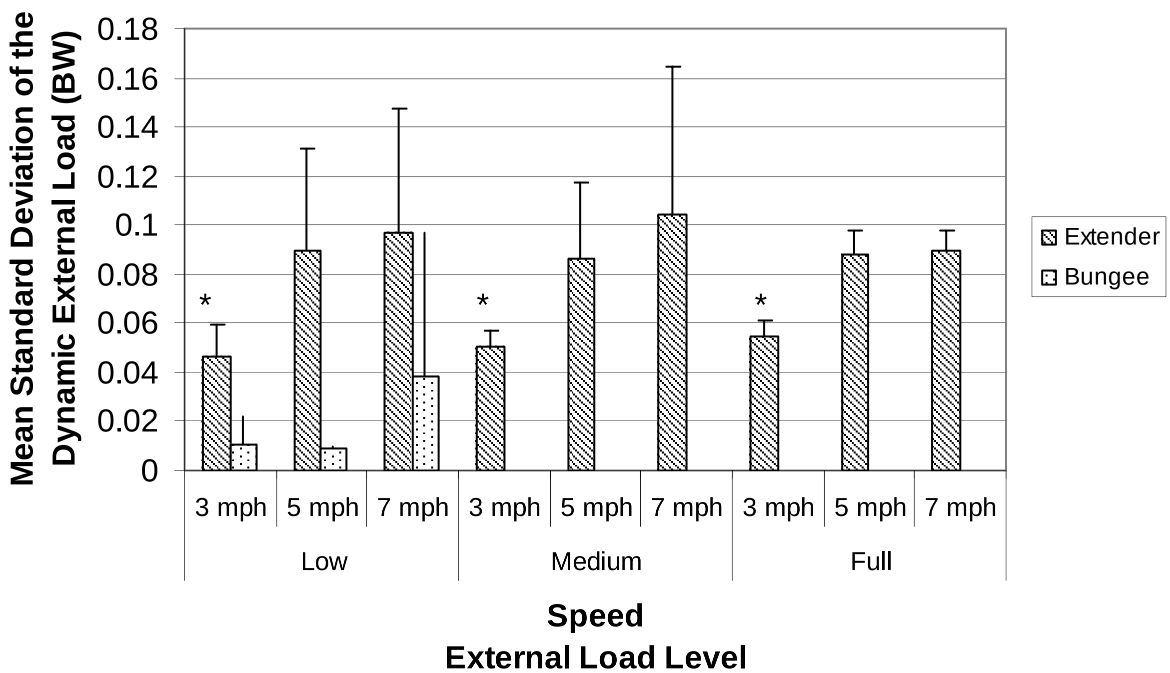

Figure 5. Mean standard deviation of the dynamic external load for each speed, load level, and loading configuration. *Significantly different from 5 mph and 7 mph, $p < 0.05$.

Table 2. Mean standard deviation of dynamic external load for each speed, load level, and loading configuration.

		Mean (SD) Dynamic External Load Standard Deviation (BW)	
Load Level	**Speed (mph)**	**SLD + Bungee Configuration**	**SLD + Extender Strap Configuration**
Low	3	0.01 (0.01)	0.05 (0.01)*
	5	0.01 (0.01)	0.09 (0.04)
	7	0.04 (0.06)	0.10 (0.05)
Medium	3		0.05 (0.01)*
	5		0.09 (0.03)
	7		0.10 (0.06)
Full	3		0.06 (0.01)*
	5		0.09 (0.01)
	7		0.09 (0.01)

*Significantly different from 5 mph and 7 mph, $p < 0.05$.

3.1.3 Change in External Load from Static to Dynamic Conditions

Because exercise prescriptions for crewmembers are based on the static load, it is important to quantify the amount of reduction in EL that occurs between static conditions (subject standing upright) and dynamic conditions (subject performing locomotion). Figure 6 and Figure 7 show the mean decrease in EL at the three load levels and three speeds for the two harness attachment configurations.

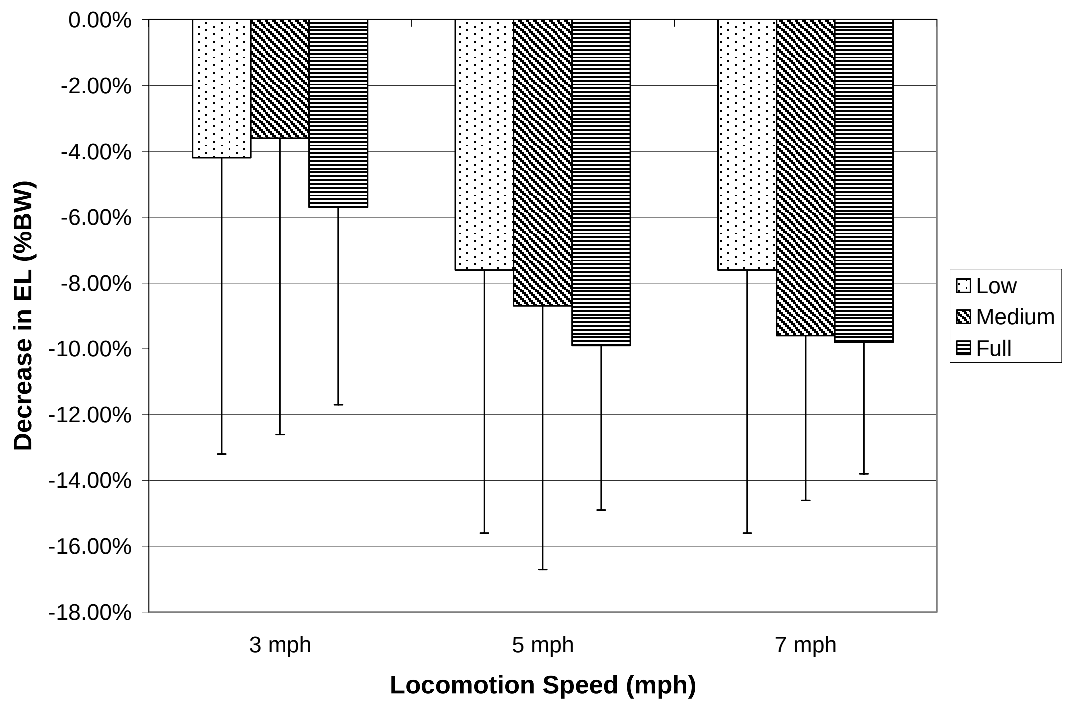

Figure 6. Mean decrease in external load (%BW) from static to dynamic conditions when the SLD was connected to the harness via an extender strap.

Figure 6 shows a consistent pattern in EL decrease when the harness was connected to the SLD via extender straps. A decrease of about -4% to -6%, occurred for walking (3 mph) and a larger decrease, about -8% to -10%, occurred during running (5 and 7 mph). Within this pattern, there also seemed to be a trend toward a greater decrease in EL with increasing load level.

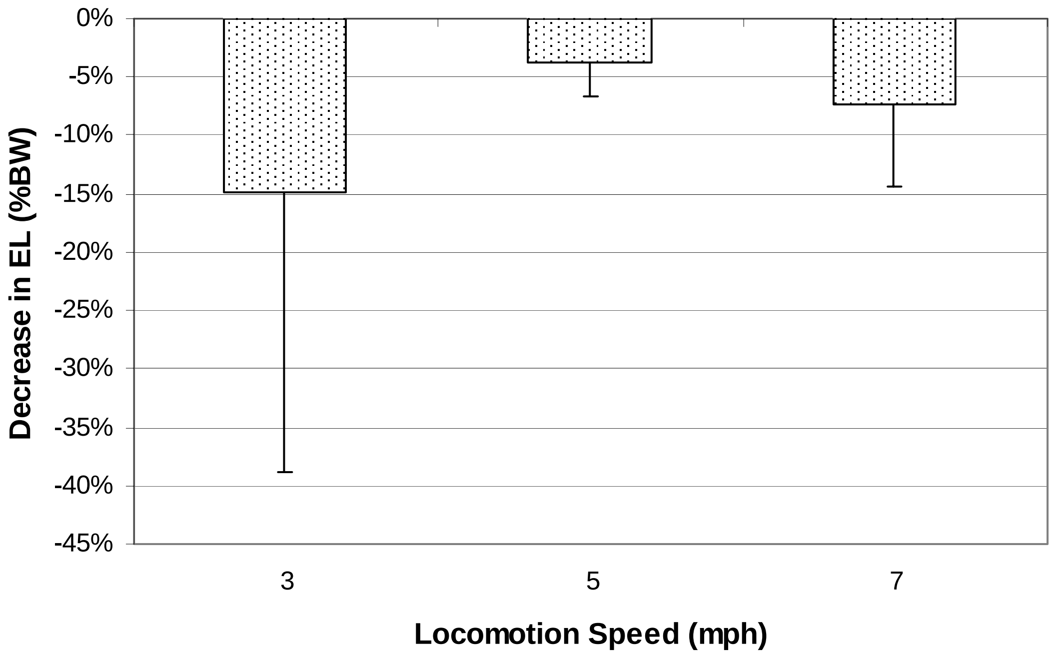

Figure 7. Mean decrease in external load (%BW) from static to dynamic conditions when the SLD was connected to the harness via a bungee.

Figure 7 shows the Low (0.7 BW) condition for the bungee. Unlike the results when the extender strap was attached, the decrease in EL with bungee attachment did not show a consistent pattern with increasing speed and ranged from about -4% at 5 mph to about -15% at 3 mph.

3.1.4 Dynamic External Load Asymmetry

Dynamic external load asymmetry is a measure of how unevenly the EL is distributed between the left and right side. A balanced load results in a load asymmetry score of 0. Load asymmetry is the absolute difference between the mean loads on the two sides of the harness.

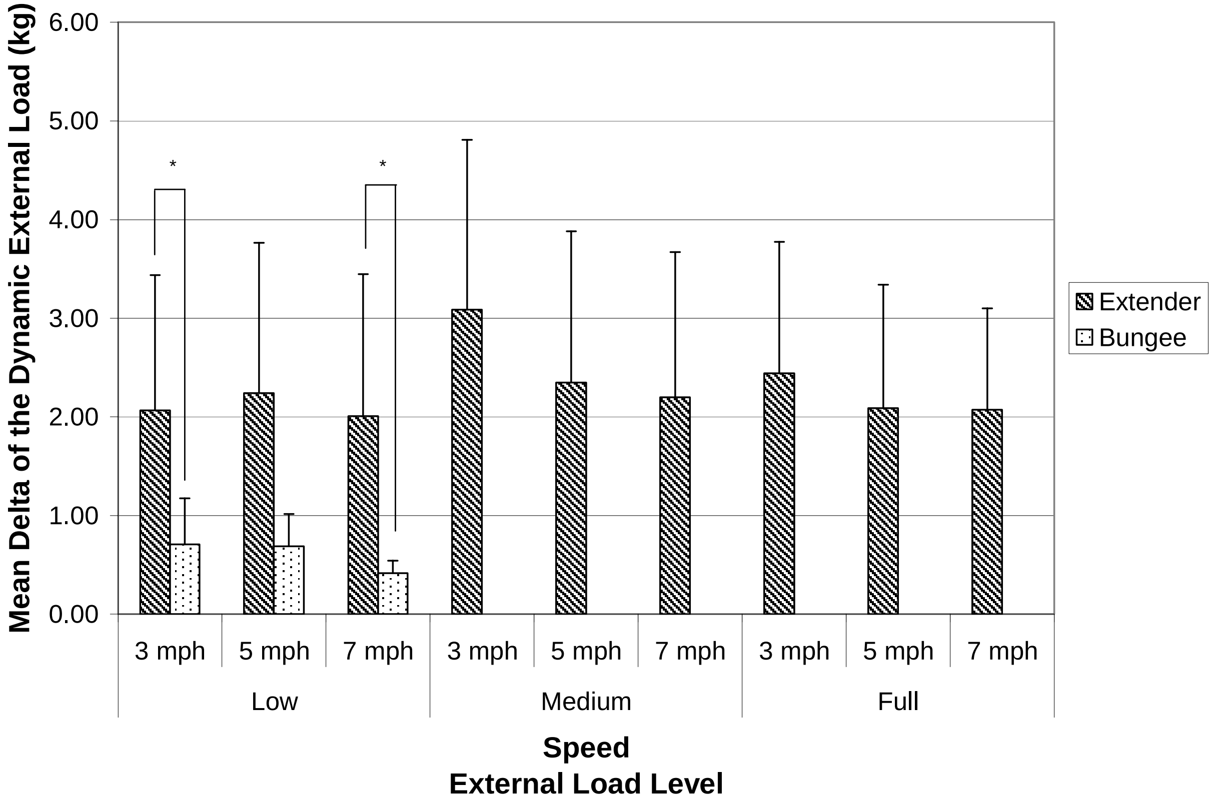

Figure 8. Mean dynamic external load asymmetry of the right and left sides during 0G trials. *Significantly different from bungee configuration, $p < 0.05$.

Table 3. Mean (SD) dynamic external load asymmetry provided by SLD during locomotion.

Load Level	Speed (mph)	Mean (SD) Dynamic External Load Asymmetry (kg) SLD + Bungee Configuration	Mean (SD) Dynamic External Load Asymmetry (kg) SLD + Extender Strap Configuration
Low	3	0.708 (0.466)	2.067 (1.371)*
	5	0.688 (0.329)	2.243 (1.522)
	7	0.417 (0.126)	2.008 (1.438)*
	Mean	0.585 (0.314)	2.109 (1.423)*
Medium	3		3.087 (1.721)
	5		2.346 (1.535)
	7		2.201 (1.471)
	Mean		2.533 (1.587)
Full	3		2.441 (1.332)
	5		2.087 (1.254)
	7		2.075 (1.025)
	Mean		2.194 (1.183)

*Significantly different from bungee configuration, $p < 0.05$.

A one-way ANOVA was used to determine if any differences existed between extender strap and bungee trials for all speeds (Figure 8 and Table 3). The load asymmetry was found to be greater during extender strap trials than during bungee trials, but asymmetries for different speeds with extender straps were not significantly different from each other. Post hoc t tests with a Bonferroni adjustment revealed that in the low load level trials, asymmetry was greater with the extender strap than with bungees at 3 mph and 7 mph, but not at 5 mph.

A second analysis was performed using only the extender strap trials in a one-way ANOVA with factors of speed and load. No interactions were found within the extender strap trials. These data suggest that asymmetries in loading occurred and that extender strap trials tend to have greater asymmetries than bungee trials during walking at 3 mph and running at 7 mph.

3.1.5 Dynamic External Load Stiffness

Dynamic external load stiffness is a measure of how much the EL changes as a function of the displacement of the SLD attachment point. Figure 9 shows that stiffness measures were greater during extender strap trials than during bungee trials. No differences in stiffness were found between speeds or load levels, and no significant interaction was found between speed and load level, suggesting that stiffness does not change as a result of change in external load level or locomotion speed (Table 4).

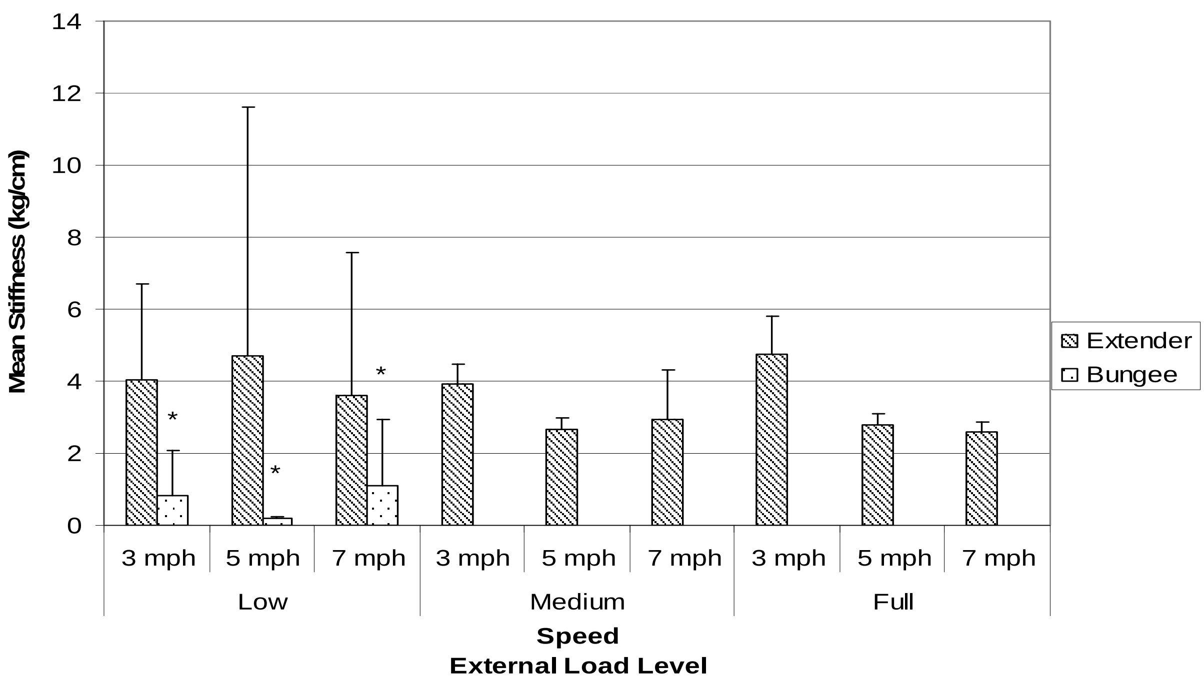

Figure 9. Mean dynamic external load stiffness during SLD trials.

Table 4. Mean stiffness of the dynamic external loads during all conditions.

		Mean (SD) Dynamic External Load Stiffness, (kg/cm)	
Load Level	**Speed (mph)**	**SLD + Bungee Configuration**	**SLD + Extender Strap Configuration**
Low	3	0.84 (1.23)	4.03 (2.66)
	5	0.20 (0.05)	4.72 (6.88)
	7	1.10 (1.83)	3.59 (3.97)
Medium	3		3.91 (0.56)
	5		2.65 (0.33)
	7		2.93 (1.38)
Full	3		4.75 (1.07)
	5		2.77 (0.34)
	7		2.58 (0.28)

The mean stiffness for all bungee trials was 0.71 kg/cm whereas the mean stiffness for all extender strap trials was 3.55 kg/cm. Consequently, the bungees lowered the loading system stiffness by approximately a factor of five.

3.1.6 <u>Gait Parameters</u>

Gait parameter data are shown in Table 5. The trends for each gait parameter are described in the sections below.

Table 5. Gait parameter outcome variables for locomotion at 0G and 1G.

	Speed	**Load [Mean (SD)]**			
Gait Parameter	**(mph)**	**Low**	**Medium**	**Full**	**1G**
Contact Time (s)	3	0.68 (0.04)	0.67 (0.03)	0.68 (0.05)	0.71 (0.05)
	7	0.28 (0.02)	0.29 (0.01)	0.29 (0.02)	0.29 (0.02)
Stride Length (cm)	3	161.86 (11.19)*	154.64 (8.43)	154.79 (10.93)	145.10 (4.89)
	7	261.35 (17.55)* †	251.09 (8.17)	236.59 (12.03)	234.19 (15.90)
Cadence (steps/min)	3	50.39 (3.65)*	52.58 (3.11)	52.51 (3.84)	55.56 (1.90)
	7	71.77 (3.21)* † ‡	75.43 (2.50)	79.72 (4.31)	79.39 (2.88)

* Significantly different from 1G ($p < 0.05$).
† Significantly different from full-body-weight loading ($p < 0.05$).
‡ Significantly different from medium-body-weight loading ($p < 0.05$).

3.1.6.1 Contact Time

Contact time was consistent for all loads at each locomotion speed, including normal gait at 1G. The contact time was about 2.3 times longer during walking than during running.

3.1.6.2 Stride Length

Mean stride length tended to decrease as load increased (Figure 10). The stride length at low loads was different from 1G walking; however, the stride lengths at medium and full loads were not different from the low or 1G condition. Stride length at low load was significantly greater than during 1G for both walking and running. In addition, during running trials, the stride length was longer during low loads than during full load.

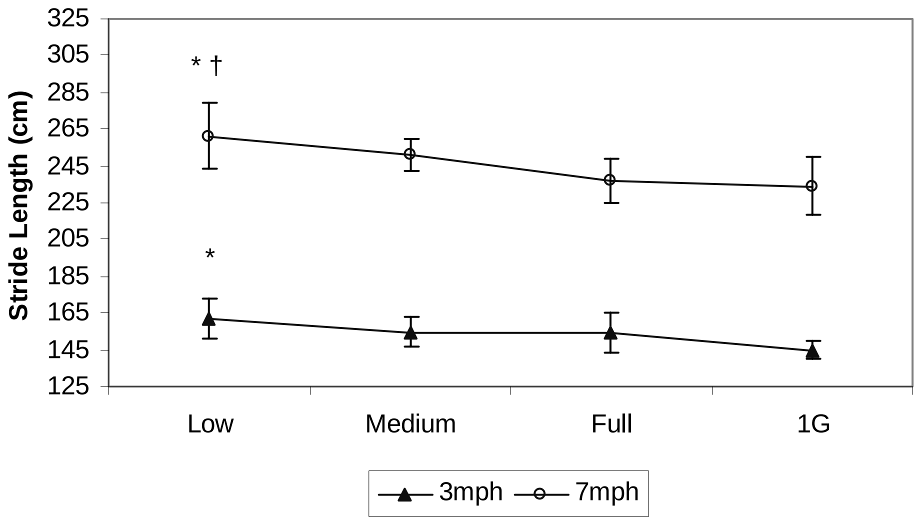

Figure 10. Mean stride length at all load levels for walking (3 mph) and running (7 mph). *Significantly different from 1G ($p < 0.05$). †Significantly different from full-body-weight loading ($p < 0.05$). ‡Significantly different from medium-body-weight loading ($p < 0.05$).

3.1.6.3 Cadence

Mean cadence tended to increase as load increased for both walking and running (Figure 11). For each speed, the cadence at low loads was different from cadence during locomotion at 1G. In addition, in the running trials the low-load cadence was significantly different from cadence with medium and full loading. There was no significant difference in cadence between the 1G trials and 0G trials with full loading.

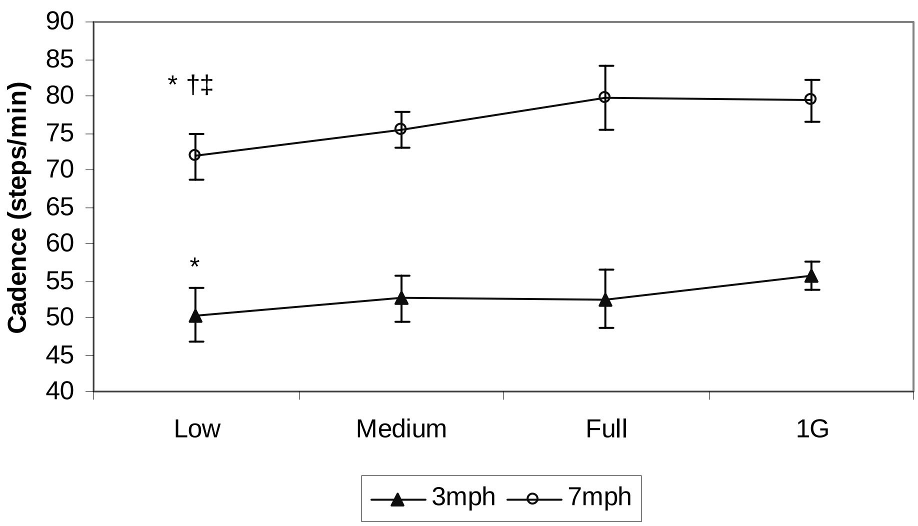

Figure 11. Mean cadence at all load levels for walking (3 mph) and running (7 mph). *Significantly different from 1G ($p < 0.05$). †Significantly different from full-body-weight loading ($p < 0.05$). ‡Significantly different from medium-body-weight loading ($p < 0.05$).

3.2 Ground Reaction Force

3.2.1 <u>Typical Ground Reaction Force Trajectories</u>

In general, GRF during locomotion has a bimodal trajectory. An initial peak represents the impact of the foot on the ground. This peak can be referred to as the impact peak because it occurs without the effort of the subject; it results solely from the forces pulling the subject to the running surface. The second peak occurs as a result of the subject propelling the body upward and forward during the latter stages of foot-ground contact. This peak can be referred to as the propulsive peak as it is directly related to the muscular effort of the subject. The relative magnitudes of the peaks depend on locomotion speed.

3.2.1.1 Typical GRF – Walking

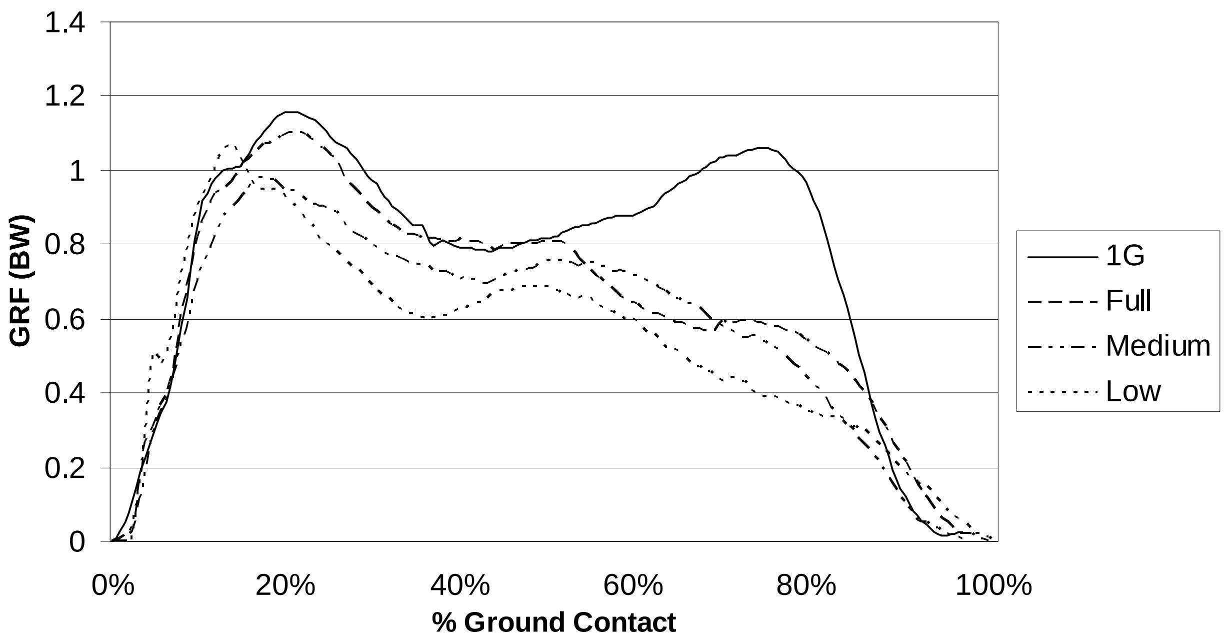

Figure 12. Typical GRF curves for walking at 3 mph during 1G and 0G at low, medium, and full EL.

Figure 12 shows typical GRF curves from single footfalls for walking at 3 mph. During the 1G trials, the active and passive peaks had similar magnitudes. In 0G trials, a different trend was seen. Although EL magnitude made no noticeable difference in the impact peak magnitude, the propulsive peak was not apparent during the 0G trials.

3.2.1.2 Typical GRF – Running

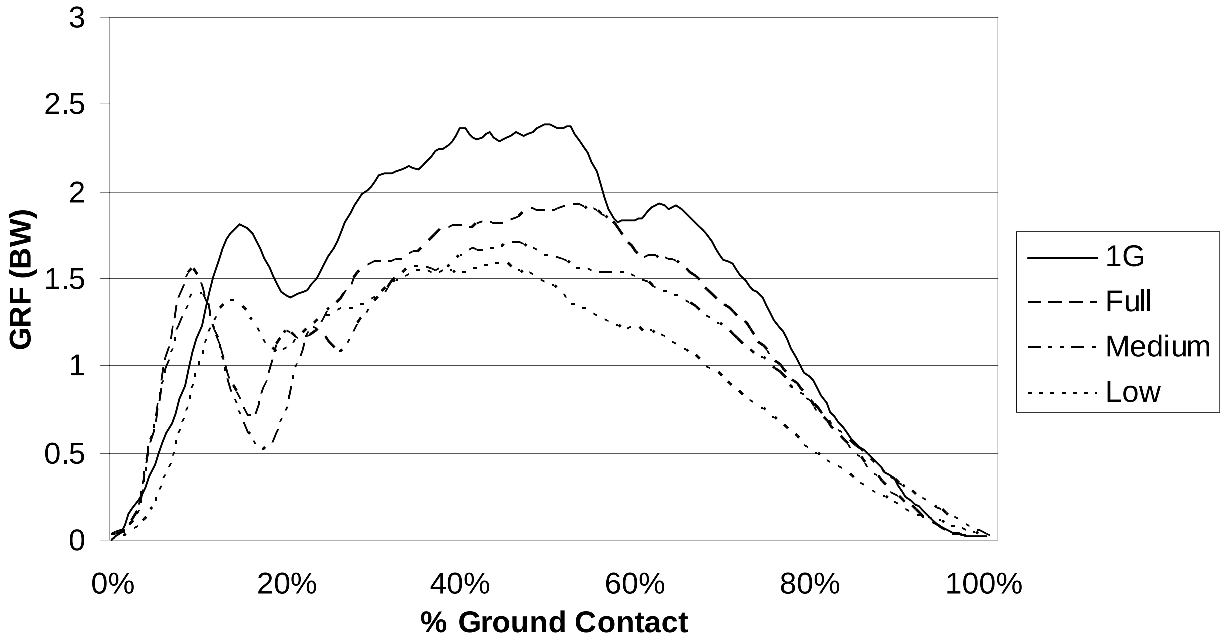

Figure 13. Typical GRF curves for running at 7 mph during 1G and 0G at low, medium, and full EL.

During running at 1G (Figure 13), the impact peak was about 75% that of the propulsive peak. In running at 0G with the SLD, the same characteristic shape of the GRF occurred, except that the magnitudes of the impact and propulsive forces were less than those seen at 1G, even at the full load level.

3.2.2 <u>Effect of Harness – U.S. vs. Russian Harness</u>

3.2.2.1 Bungee-SLD Configuration

An initial analysis of each harness attachment method (bungees and extender straps) was conducted to determine if harness type had any effect on GRF variables. Each speed was examined independently. For bungee trials, SLD loads were set only at low load, so a two-sample t test was executed and revealed that outcome variables for different harness types were not significantly different from each other (Table 6). It should be noted that throughout the entire study, only 2 bungee trials of the 3-mph condition were collected with the Russian harness. These limited data suggest that using bungees instead of extender straps in combination with the SLD does not affect GRF outcome variables

Table 6. Mean (SD) for each outcome variable in 0G locomotion trials using bungees in combination with SLD – Russian vs. U.S. Harness

Outcome Variable	Speed (mph)	Russian Harness	U.S. Harness
Peak Impact Force (BW)	3	1.02 (0.09)	0.93 (0.21)
	7	1.34 (0.24)	1.30 (0.29)
Peak Propulsive Force (BW)	3	0.61 (0.13)	0.57 (0.18)
	7	1.66 (0.16)	1.67 (0.24)
Peak GRF (BW)	3	1.02 (0.08)	0.93 (0.21)
	7	1.66 (0.16)	1.68 (0.24)
Loading Rate (BW/s)	3	7.76 (1.66)	8.51 (1.07)
	7	40.04 (13.10)	41.17 (8.88)
Impulse (BW*msec)	3	375.00 (9.10)	336.90 (96.60)
	7	273.63 (25.06)	258.22 (24.99)

3.2.2.2 Extender Strap-SLD Configuration

A two-factor ANOVA (load × harness type) was conducted with the trials using the extender straps in combination with the SLD for each speed. Only low and medium load levels were used with the Russian harness, so the statistical analysis was limited to those load levels. No significant difference between results for the two harnesses was found for any of the outcome variables (Table 7). From these analyses, it can be concluded that at static external loading levels of low and medium, the type of harness does not affect GRF outcome variables when extender straps are used with the SLD during walking at 3 mph or running at 7 mph. Furthermore, regardless of the SLD attachment configuration, harness type does not affect GRF outcome variables.

Table 7. Mean (SD) for each outcome variable for 3-mph trials using extender straps in combination with the SLD – Russian vs. U.S. Harness

Outcome Variable	Speed (mph)	Russian Harness		U.S. Harness		
		Low	Medium	Low	Medium	Full
Peak Impact Force (BW)	3	0.91 (0.11)	1.11 (0.05)	0.93 (0.18)	1.09 (0.14)	1.22 (0.14)
	7	1.39 (0.40)	1.32 (0.43)	1.30 (0.28)	1.39 (0.33)	1.50 (0.22)
Peak Propulsive Force (BW)	3	0.61 (0.10)	0.74 (0.03)	0.66 (0.17)	0.73 (0.12)	0.90 (0.09)
	7	1.47 (0.28)	1.73 (0.19)	1.61 (0.12)	1.69 (0.20)	1.94 (0.18)
Peak GRF (BW)	3	0.91 (0.11)	1.11 (0.05)	0.93 (0.18)	1.09 (0.14)	1.22 (0.14)
	7	1.58 (0.28)	1.74 (0.17)	1.61 (0.11)	1.72 (0.16)	1.95 (0.17)
Loading Rate (BW/s)	3	7.55 (0.97)	9.52 (2.94)	7.63 (2.67)	8.16 (1.58)	8.06 (1.10)
	7	42.2 (15.5)	45.3 (10.7)	43.0 (14.2)	46.7 (12.4)	47.8 (9.3)
Impulse (BW*msec)	3	374.8 (34.6)	434.0 (35.8)	381.6 (64.6)	442.6 (41.3)	530.1 (42.7)
	7	263.9 (39.7)	288.5 (0.7)	264.8 (30.2)	290.3 (24.7)	332.6 (23.9)

3.2.3 Effect of Harness Attachment Method – Bungee vs. Extender Straps

Because no differences were found between harness types (Russian and U.S.) within each harness attachment method, data from each harness type were pooled to determine if harness attachment method affected GRF variables. A two-sample t test was conducted to identify any interactions because the only bungee data available were for low SLD load levels. Each speed was examined independently. No significant difference between attachment methods was found for any outcome variable (Table 8).

Table 8. Mean (SD) for each outcome variable for trials using extender straps and bungees – low external loads.

Outcome Variable	Speed (mph)	Bungees	Extenders
Peak Impact Force (BW)	3	0.95 (0.18)	0.92 (0.15)
	7	1.31 (0.26)	1.34 (0.32)
Peak Propulsive Force (BW)	3	0.58 (0.16)	0.63 (0.14)
	7	1.67 (0.20)	1.55 (0.20)
Peak GRF (BW)	3	0.95 (0.18)	0.92 (0.15)
	7	1.67 (0.20)	1.60 (0.19)
Loading Rate(BW/s)	3	8.29 (1.16)	7.59 (2.04)
	7	40.72 (10.07)	42.68 (14.06)
Impulse (BW*msec)	3	347.80 (81.09)	378.58 (52.07)
	7	264.38 (24.89)	264.43 (32.73)

At a static external loading level of 70% BW, the attachment type (extender straps vs. bungees) did not have an effect on GRF outcome variables during walking at 3 mph or running at 7 mph. This result, coupled with the lack of differences between harnesses, prompted the pooling of trials by harness type and extension type within load levels for further analysis at both speeds.

3.2.4 Effect of External Load Level on GRF Variables

An ANOVA was executed to assess the effect of external load level on GRF variables. Separate analyses were executed for each outcome variable for each speed condition. Tukey-Kramer post hoc t tests were used to identify significant differences between load levels. Significant load effects were found for each outcome variable except loading rate (Table 9). Each outcome variable will be discussed separately.

Table 9. GRF outcome variables summarized by external load level (0G and 1G).

Outcome Variable	Speed (mph)	Low Load	Medium Load	Full Load	1G Total
Peak Impact Force (BW)	3	0.93 (0.16)* †‡	1.10 (0.12)	1.22 (0.14)	1.14 (0.05)
	7	1.33 (0.29)*	1.37 (0.32)*	1.50 (0.22)	1.80 (0.49)
Peak Propulsive Force (BW)	3	0.62 (0.15)* †	0.73 (0.11)* †	0.90 (0.09)*	1.09 (0.07)
	7	1.60 (0.21)* †	1.69 (0.19)* †	1.94 (0.18)*	2.37 (0.09)
Peak GRF (BW)	3	0.93 (0.16)* ‡†	1.10 (0.12)	1.22 (0.14)	1.17 (0.05)
	7	1.63 (0.19)* †	1.73 (0.15)* †	1.95 (0.17)*	2.38 (0.09)
Loading Rate (BW/s)	3	7.79 (1.84)	8.46 (1.82)	8.06 (1.10)	7.29 (0.71)
	7	41.8 (12.2)	46.4 (11.6)	47.8 (9.3)	46.0 (15.3)
Impulse (BW*msec)	3	370.0 (61.4)* †‡	440.7 (38.1)* †	530.1 (42.7)	546.6 (29.5)
	7	264.4 (28.8)* †‡	289.9 (21.8)* †	332.6 (23.9)*	375.9 (16.4)

* Significantly different from 1G ($p < 0.05$).
† Significantly different from full body weight loading ($p < 0.05$).
‡ Significantly different from medium body weight loading ($p < 0.05$).

3.2.4.1 Peak Impact Force

Walking – The low load condition resulted in significantly lower impact forces than the other 0G loading conditions and the 1G condition (Figure 14). However, no differences were found between the medium and full loading conditions and 1G.

Running – Peak impact force tended to increase as external load increased, and peak impulse forces for the low and medium external load conditions were significantly different from the force in the 1G conditions (Figure 14). Peak impact force for the full body weight condition was not significantly different from the force at 1G.

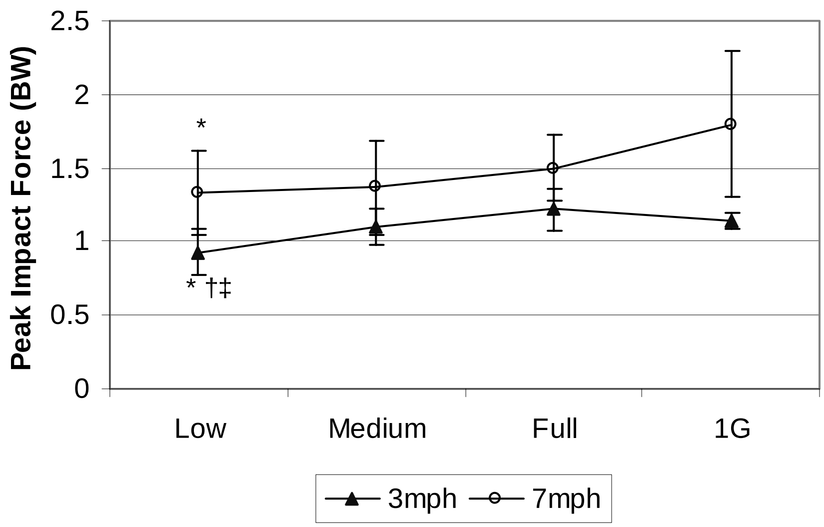

Figure 14. Peak impact force during locomotion at 0G and 1G – walking and running. *Significantly different from 1G ($p < 0.05$). †Significantly different from full-body-weight loading ($p < 0.05$). ‡Significantly different from medium-body-weight loading ($p < 0.05$).

3.2.4.2 Peak Propulsive Force

For both locomotion speeds, the peak propulsive force was significantly less during all 0G trials than during 1G trials (Figure 15). This force was less in low-load and medium-load trials than in trials with the full body weight load.

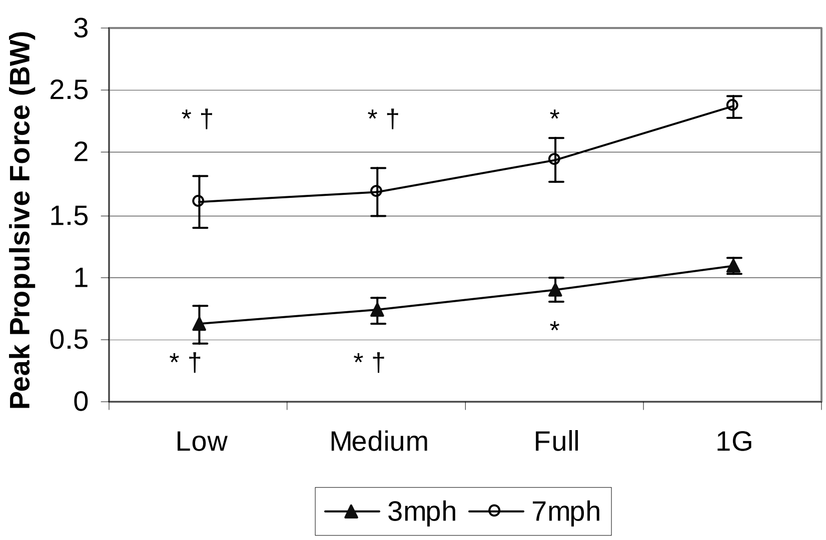

Figure 15. Peak propulsive force for locomotion at 0G and 1G – walking and running. *Significantly different from 1G ($p < 0.05$). †Significantly different from full-body-weight loading ($p < 0.05$).

3.2.4.3 Peak Ground Reaction Force

Walking – The peak GRF was less during the trials with a low load than during trials with any other load level or at 1G (Figure 16). No difference was found between peak GRFs at the medium, full, and 1G conditions.

Running – No significant difference in the peak GRF was found between the low- and medium-load trials, but peak GRF under each of these conditions was lower than for the full-load and 1G trials (Figure 16). In addition, the peak GRF for full-load trials was different from the peak GRF for 1G trials.

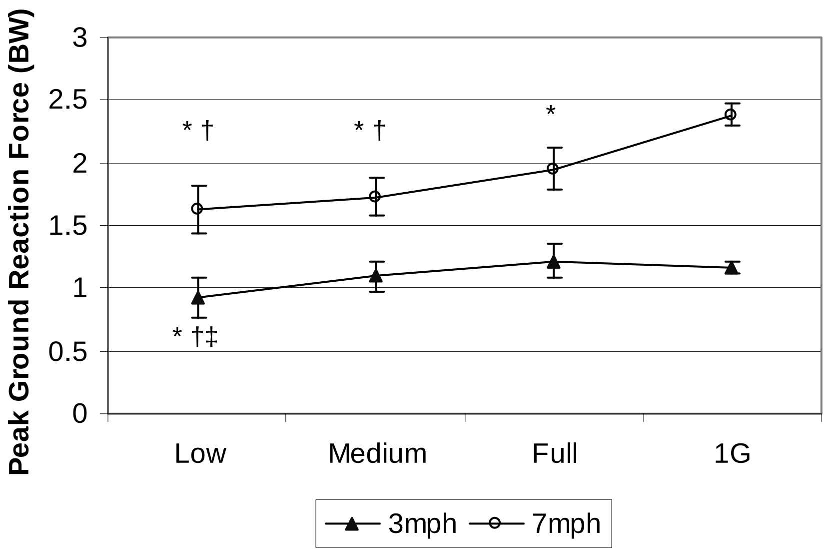

Figure 16. Peak ground reaction force for locomotion at 0G and 1G – walking and running. *Significantly different from 1G ($p < 0.05$). †Significantly different from full-body-weight loading ($p < 0.05$). ‡Significantly different from medium-body-weight loading ($p < 0.05$).

3.2.4.4 Loading Rate

No significant difference in loading rates was found between load levels in walking or running (Figure 17).

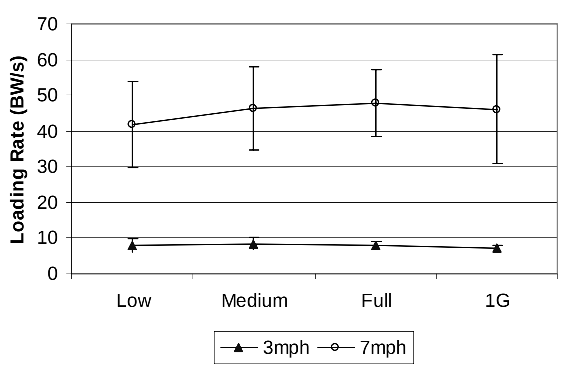

Figure 17. Loading rate for locomotion at 0G – walking and running.

3.2.4.5 Impulse

Walking - Impulse increased as external load increased (Figure 18). All load conditions were significantly different from each other, and impulse magnitudes were less at 0G than at 1G.

Running – Impulse at the low and medium loading conditions was less than impulse at the full body weight and 1G conditions (Figure 18). No difference in impulse was found between the full body weight loading condition and 1G.

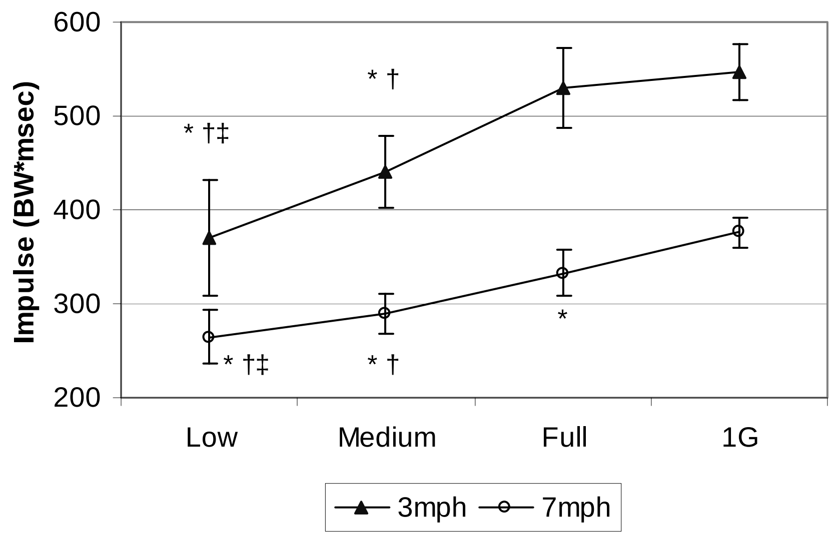

Figure 18. Impulse for locomotion at 0G and 1G – walking and running. *Significantly different from 1G ($p < 0.05$). †Significantly different from full-body-weight loading ($p < 0.05$). ‡Significantly different from medium-body-weight loading ($p < 0.05$).

3.2.5 Prediction of Peak Ground Reaction Force from External Load

Although tests were completed at static loads of low, medium, and full body weight for each subject, dynamic loads varied during each trial, ranging from 0.60 to 1.09 BW during 3-mph locomotion and 0.52 to 0.94 BW during 7-mph locomotion (Figure 19). Peak GRF ranged from 0.75 to 1.44 BW for 3 mph and 1.17 to 2.21 BW for 7 mph. A significant association between mean EL and peak GRF was identified for each speed (Table 10).

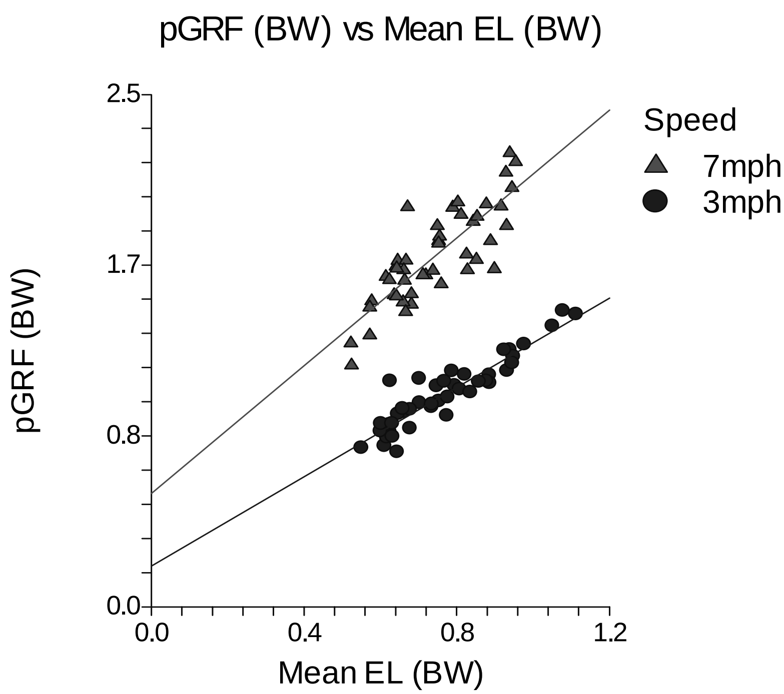

Figure 19. Scatter plot of peak ground reaction force vs. external load for all trials.

Table 10. Regression equation and r-squared value for peak ground reaction force as predicted by external load, for each locomotion speed. Slopes and r^2 values were significant ($p < 0.05$).

Speed (mph)	Equation	r^2
3	Peak GRF = 0.20 + 1.08 × EL	0.80
7	Peak GRF = 0.58 + 1.53 × EL	0.69

NOTE: Peak GRF and EL are expressed in BW.

A secondary multiple regression analysis was completed using speed and mean EL as independent variables. Locomotion speed was categorized as 0 for 7-mph trials and 1 for 3-mph trials. A significant equation was found that related locomotion speed and EL to peak GRF (Table 11).

Table 11. Regression equation and r-squared value for both speeds relating external load to peak ground reaction force.

Equation	r^2
Peak GRF = 0.75 + 1.29 × EL - 0.71 × (locomotion coefficient) where locomotion coefficient = 0 for 7 mph and 1 for 3 mph	0.92*

These results suggest that the peak ground reaction force increases with an increasing external load. However, the increase in peak GRF depends on locomotion speed. Furthermore, the expected peak GRF based on external load is not directly proportional to the peak GRF obtained during 1G locomotion. The equations above can be used to predict the EL that would be required to achieve a 1G-like peak GRF. During walking at 3 mph, the EL required to achieve a peak GRF of 1.2 (1G mean) is 0.93. During running at 7 mph, the EL required to achieve a peak GRF of 2.4 (1G mean) is 1.19.

3.3 Kinematics

No significant difference between 1G and 0G was found in peak ankle angular displacement or velocity or peak knee angular displacement (Table 12). However, a significant difference between gravitational conditions was found in peak knee flexion (maximum) and extension (minimum) angular velocity (Table 13).

Table 12. Peak ankle angular displacement and peak ankle angular velocity (mean ± SD).

	Ankle Angular Displacement Max (º)	Ankle Angular Displacement Min (º)	Ankle Plantarflexor Angular Velocity Max (º/s)	Ankle Dorsiflexor Angular Velocity Min (º/s)
1G	44.26 ± 6.93	3.14 ± 3.20	425.28 ± 47.89	-166.87 ± 11.26
0G	44.77 ± 4.00	4.21 ± 3.43	381.73 ± 77.74	-176.09 ± 19.52
Difference	0.51 ± 1.75	1.07 ± 2.88	43.55 ± 42.59	9.22 ± 12.94

Table 13. Peak knee angular displacement and peak knee angular velocity.

	Knee Angular Displacement Max (º)	Knee Angular Displacement Min (º)	Knee Flexor Angular Velocity Max (º/s)	Knee Extensor Angular Velocity Min (º/s)
1G	102.47 ± 8.62	15.90 ± 2.29	537.50 ± 56.77	-648.04 ± 37.31
0G	98.89 ± 8.44	23.41 ± 3.95	466.33 ± 33.61	-511.95 ± 57.80
Difference	3.58 ± 5.65	7.52 ± 5.06	71.17 ± 30.28 *	136.10 ± 29.83 *

* Significant difference between 0G and 1G ($p < 0.05$).

Peak knee flexion angular velocities were less during 0G locomotion than during 1G locomotion. Furthermore, peak knee extension angular velocities were less at 0G than at 1G (Table 13). Therefore, the range of knee angular velocities was greater during 1G locomotion than during 0G. Peak flexion velocities occurred after toe off during the earlier portion of the swing, whereas peak extension velocities occurred before heel strike during the latter portion of the swing (Figure 20). These data suggest that during running at 7 mph, knee velocity at 0G and 1G may be slightly different, but there is little difference in range of motion.

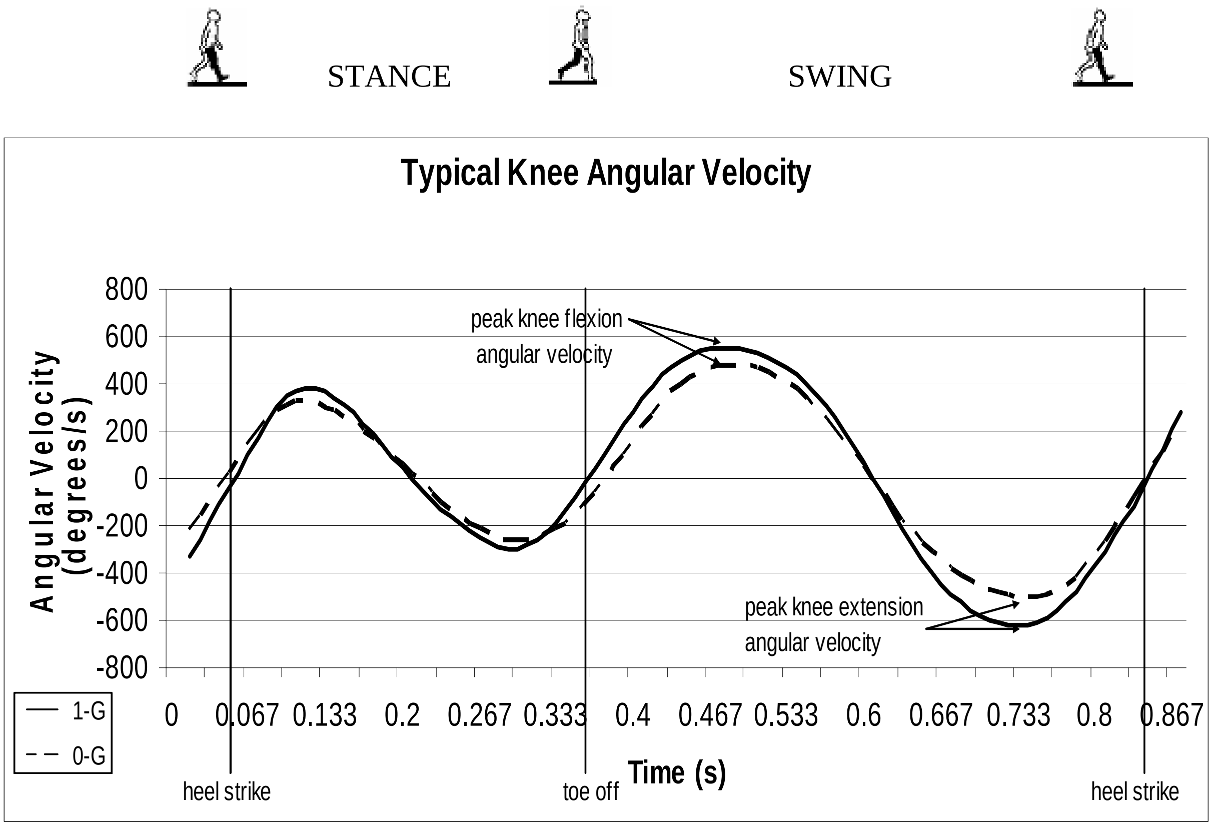

Figure 20. Knee angular velocity for one stride of one subject. Positive values indicate flexion angular velocity and negative values indicate extension angular velocity.

4 DISCUSSION

The purpose of this investigation was to compare the effects of modifying external loads, applied by the SLD, on locomotion at 0G and 1G. Subjects were tested at three load levels during walking and running at 0G and during normal treadmill walking and running at 1G. External load parameters and biomechanical variables were analyzed to determine the differences in these variables during locomotion at 0G and 1G. In addition, the effect of the Russian and U.S. harnesses and the use of bungees and extender straps with the SLD were studied.

Although the total external load level was not affected by locomotion speed, the variation in external load was consistently less during 3-mph walking than during 5-mph and 7-mph running, and this may make walking more comfortable than running. When subjects went from the static (standing upright) to the dynamic (walking or running) condition, the external load decreased more for running than for walking. So that exercise prescriptions for crewmembers can be more accurate, the static SLD load setting for running may need to be increased slightly from the setting for walking, to compensate for the larger decrease in dynamic external load when a subject runs.

Stiffness was greater with the extender strap attachment than with the bungee attachment. The bungees themselves have relatively low stiffness, and when placed in line with the SLD, they lowered the loading system stiffness by a factor of approximately five. Stiffness did not seem to be affected by external load level or locomotion speed. Bungees also seemed to decrease EL asymmetries, which may occur because of slightly different harness attachment lengths. The elasticity of the bungees may compensate for length differences between the two sides. This suggests that it is highly desirable to use the bungees in conjunction with the SLDs to minimize load variation and load asymmetry during locomotion. On the other hand, the bungees do not allow an external load of more than about 70% BW, which is a significant limitation.

Gait parameters showed generally consistent patterns with regard to load level for both walking (3 mph) and running (7 mph). There was no difference in contact time for different loads. As load increased, stride length decreased and cadence increased. That is, subjects tended to take shorter steps, but at a faster rate. When a change in cadence is coupled with a consistent contact time, flight time must change. Therefore, a decrease in load resulted in an increase in flight time. Increased flight time resulted in a greater stride length (more tread passed beneath the foot during the airborne phase). All three gait parameters showed no difference between full loading at 0G and normal (1G) loading. This suggests that when subjects were loaded at nearly full body weight during locomotion in weightlessness, they achieved a gait pattern similar to the pattern at 1G.

GRF was not affected by harness type (Russian or U.S.) at low and medium load levels. Also, at low load levels GRF was not affected by harness attachment method (bungees or extender straps). Peak propulsive forces at 0G were less than those at 1G, even when subjects were loaded at close to full body weight. For all conditions, the loading rate at 0G was similar to the rate at 1G.

Peak impact force and impulse results differed slightly between walking and running in weightlessness. For walking, peak impact forces with low external load were less than those at 1G, but at medium and full loading, peak impact forces were similar to those at 1G. However, for running, both low and medium external loading in weightlessness produced peak impact forces that were less than those at 1G. For walking, impulse was less during low and medium loading in weightlessness than at full body weight and 1G, but impulse during full body loading was similar to impulse at 1G. For running in weightlessness, impulse increased as external load increased, but was less than impulse at 1G, even when subjects were loaded to full body weight.

The regression equations suggest that the peak GRF obtained during 0G locomotion can be reasonably predicted using the applied external load. However, the peak GRF obtained at 0G is not directly proportional to the peak GRF during 1G locomotion. For example, the peak ground reaction force during 7-mph running at 1G is 2.38 BW. If a runner is loaded to 1 BW at 0G, the 7-mph equation predicts the 0G peak GRF to be 2.11 BW. The equation for both speeds predicts the 0G peak GRF to be 2.04 BW. The measured 0G peak GRF at 1 BW EL was 1.95 BW. The differences between the actual and predicted values are 8% for the 7-mph equation and 4.6% for the more general equation. In each case, the peak GRF at 0G is less than that at 1G. For that reason, it would be inappropriate to expect the peak GRF obtained at 0G, with an EL of some percentage of BW, to be the corresponding percentage of peak GRF obtained during 1G locomotion.

The limited kinematic analysis conducted in this investigation compared ankle and knee displacement and velocity for 7-mph running in weightlessness versus 1G. Knee range of motion was similar for the two gravitational conditions. Peak knee velocity was slower at 0G than at 1G. The slower knee velocity at 0G was consistent with the cadence being slower at 0G with full loading than at 1G. Because cadence was decreased at 0G, the subjects had to move their legs more slowly to create the lower step rate. Only knee and ankle kinematics were examined in this investigation, so it is unclear if differences in hip motion were also present. The kinematic data suggest that the subjects created a slower cadence at 0G through decreased knee velocity during both the recovery and forward movement of the leg. This finding is interesting because the decreased cadence, coupled with a consistent foot-ground contact time, suggests that flight time increases at 0G and that the increase in flight time is affected by EL. As EL decreases, flight time increases because the force acting to pull the subject back to the running surface is decreasing. The decreases in peak knee velocity for both extensor and flexor motions, coupled with a relatively consistent range of motion, suggest that the subjects may scale their locomotion pattern to flight time. In other words, rather than adapt to the increased flight time by modifying a part of their running motion, the subjects adapt by maintaining the same pattern while decreasing the velocity of the knee. From a neuromuscular control perspective, this may indicate that a mechanism exists that is based on the maintenance of joint range of motion during locomotion. The subjects can adopt a variety of strategies to adjust to the altered loading condition at 0G, but choose to maintain a consistent running form by decreasing their speed of movement. This may suggest that running form remains consistent at 0G through modification of the forces produced at the knee.

A limitation of this study was that the treadmill used during the 0G trials was not the actual TVIS that is used by crewmembers aboard the ISS. The tread area of the testing device was modified to be the same size as the TVIS tread area, but the test treadmill was bolted to the floor of the aircraft. The TVIS used on the ISS floats in a pit in the service module and is connected to the structure only by soft springs, which provide vibration isolation from the ISS structure. TVIS is equipped with gyroscopes to maintain the stability of the treadmill during locomotion. The fact that the TVIS is floating results in decreased resistance of the treadmill surface to any applied forces. As the crewmember runs and applies impact and propulsive forces to the TVIS, some motion of the treadmill occurs. This may modify each of the parameters studied in this investigation. Gait timing patterns may change, kinematics may be affected, and ground reaction force parameters may be different. For these findings to be confidently applied to the TVIS, an experiment needs to be conducted in weightlessness on a device that reproduces the vibration isolation system used for TVIS.

5 CONCLUSIONS

The mean external load provided to subjects during locomotion is smaller than that predicted from static measurements. When a subject uses extender straps, the external load decreases by 4 to 10% as locomotion speed increases from 3 to 7 mph. Furthermore, as the load setting is increased from low (70% BW) to full (100% BW) loading, external load decreases approximately 2–3% at each speed. When a subject uses bungees, the load decreases by 14% for walking and by 4–8% for running.

The mean load stiffness when the extender straps were used was 3.55 kg/cm, whereas the mean load stiffness when bungees were used was 0.71 kg/cm. Thus, the bungees decreased load stiffness by a factor of approximately 5. This greatly reduced the variation in external load during locomotion. Bungees also seem to reduce the asymmetry in external load. Both the reduced stiffness and reduced asymmetry should improve crewmember comfort.

The similarity of gait and kinematic parameters obtained at 0G and 1G suggests that locomotion in weightlessness is similar in form to locomotion on the same device at 1G, especially when load levels approach full body weight loading. However, a more extensive kinematic analysis is required to further support this conclusion.

The ability of the SLD to reproduce 1G-like GRFs seems to be more limited. The SLD may be able to reproduce 1G-like peak impact force, and at higher load levels may also reproduce impulse. However, even at external load levels close to full body weight, locomotion using this device does not reproduce the peak active force or loading rate experienced at 1G. Because peak active force and loading rate are considered to be important factors for maintenance of bone mineral density, the development of next-generation subject loading devices should strive to replicate 1G levels of these parameters. In addition, more research should be conducted; possibly including long-term training and bed rest studies, to determine how the alterations in GRF that would be expected to occur at 0G might affect neuromuscular conditioning and bone mineral density.

Some recommendations may be made regarding future efforts to improve the performance of the SLD:

- The external load should not change from the initial load setting with changes in locomotion speed (including going from standing to running) and load level.

- The loading stiffness should be reduced. A target of 0.05 kg/cm or less is recommended (as captured in the treadmill requirements).

- The GRF during running should not be less than twice the initial external load setting.

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Prolonged exposure to weightlessness associated with spaceflight provokes profound physiological changes in humans. Two areas of significant concern are bone loss and neuromuscular deconditioning. Changes in lower-limb neuromuscular activation patterns have raised concern about the ability of astronauts to escape from their spacecraft in an emergency landing situation. Countermeasures that stimulate bone maintenance and locomotor control are vital to the success of human spaceflight.
It has been hypothesized that, to help maintain the bone mineral density that an astronaut has in normal gravity, the ground reaction force (GRF) achieved during locomotion in weightlessness should mimic the GRF that occurs on Earth. Currently, a crewmember exercising on the treadmill recieves an external load by means of an upper-body harness that attaches to the SLD via extender straps or bungees. GRF is imparted as the SLD pulls the crewmember toward the treadmill surface during locomotion.
The primary objective of this investigation was to determine if gait, GRF, an kinematics during locomotion at 0G, using a simulated ISS treadmill and SLD, are significantly different from those found during locomotion at 1G. The secondary objective of this investigation was to quantify the load magnitude, variability, and stiffness of the external load applied to the crewmember by the SLD during locomotion at -G.

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