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Kinematic and Electromyographic Evaluation of Locomotion on the Enhanced Zero-gravity Locomotion Simulator: A Comparison of External Loading Mechanisms

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Dedication

This work is dedicated to the memory of R. Donald Hagan, who left this world during the completion of this report. His mentoring, trust, and encouragement is evident in this paper, along with all of the reports published by anyone who worked in the NASA-Johnson Space Center Exercise Physiology Laboratory under his leadership. His memory lives on in this research.

Table of Contents

List of Tables

List of Figures

Abstract

Background: Horizontal suspension treadmill locomotion is often used as an analog for locomotion in microgravity. The enhanced Zero-gravity Locomotion Simulator (eZLS) is currently used to study locomotion in microgravity. External load (EL) can be provided by various means, including electro-mechanically, pneumatically, or via passive elastic devices. The present study compares EL applied with bungees (passive elastic device) and with a linear motor (active electro-mechanical) subject loading device. To date, there are no data that quantify differences in locomotion depending upon the external loading manner.

Purpose: The primary purpose was to determine how EL type affects locomotion patterns and muscular activity on the eZLS. Specifically, when using bungees or a linear motor subject loading device as the two EL types.

Methods: Eight subjects were suspended on the eZLS while walking at 1.34 $m{\cdot}s^{-1}$ (3 mph) and running at 3.13 $m{\cdot}s^{-1}$ (7 mph). The EL was provided by either bungees or a linear motor subject loading device (LM-SLD) at approximately 55% and 90% of body weight during the eZLS exercise. Joint kinematics, ground reaction forces (GRF), and electromyographical (EMG) activity of lower body musculature were measured during each condition. Repeated measures analysis of variance (ANOVA) were used to test for differences between EL types within load levels on the eZLS ($p<0.05$).

Results: There were few differences in locomotion patterns and muscular activity between loading mechanisms. GRF were greater with the LM-SLD than with bungees during eZLS locomotion. GRF magnitudes for both devices were lower than previously reported values obtained during upright locomotion in normal gravity, but similar to those found in actual microgravity.

Discussion and Operational Relevance: Greater GRF with the LM-SLD than with bungees suggests that the use of a constant-force SLD may be of potential benefit during treadmill exercise because locomotion patterns do not change, but subjects experience increased force magnitude and loading rates applied at the feet.

1.0 Introduction

1.1 Background and Significance

When performing locomotive exercise onboard the International Space Station (ISS), astronauts wear a harness attached to a vertical tether that pulls them back to the treadmill (Figure 1). Impact and propulsive forces are applied to the axial skeleton via ground reaction forces (GRF) that occur during foot and treadmill surface contact. The amount of external load (EL) used during each exercise session is variable and dependent upon the loading mechanism. Bungees are used as one mode of loading device. The EL delivered by bungees is dependent upon length and is varied by inserting one or more carabiner clips between the bungee and the attachment point of the treadmill. The Subject Loading Device (SLD) is a mechanical EL apparatus that applies an EL chosen by the astronaut. The current SLD used in-flight offers a wider selection of EL than bungees, but it has a greater stiffness.[1]

Studying locomotion in actual microgravity is difficult and expensive. While data can be collected directly from astronauts during spaceflight, or during parabolic flight experiments, there are obstacles. The drawbacks to spaceflight experiments include difficulty using necessary data collection hardware, and completing an experiment with adequate sample size. Parabolic flight offers a viable ground-based alternative, but periods of microgravity are limited to 20-30 seconds, which only allows for acute locomotion investigations. In addition, variables such as comfort level and steady-state metabolic rate are difficult or impossible to attain. Ground-based simulators such as the enhanced Zero-gravity Locomotion Simulator (eZLS) provide a relatively low-cost, longer duration simulation of locomotion in microgravity than parabolic flight.

Figure 1: Astronaut exercising on a treadmill onboard the International Space Station. The astronaut is wearing a harness around his waist and shoulders and is connected to the treadmill via bungee cords (vertical attachment between the harness and treadmill).

Horizontal suspension locomotion is an analog used to study walking and running in conditions similar to microgravity. With this analog, the subject is suspended by cables in a horizontal position, and the treadmill is oriented vertically, parallel to the direction of gravity. The system is arranged so that no gravitational forces are oriented between the treadmill surface and the subject, thus simulating microgravity. Researchers have used this arrangement to study the effects of load and harness treatments upon locomotion in microgravity.[2,3,4,5]

The eZLS is a horizontal suspension microgravity analog treadmill that is currently in use at the NASA Glenn Research Center in Cleveland, Ohio. The eZLS allows for locomotion studies in simulated microgravity, or partial gravity (e.g., lunar or Mars gravitational equivalents). For microgravity simulations, subjects are suspended horizontally while performing locomotion on a vertically mounted motorized treadmill. Latex cords are attached to fabric cuffs that support each shank, thigh, upper arm and lower arm. The cuffs are placed to approximate the location of the center of mass of each segment. The tension in each cord can be adjusted using an overhead pulley system to balance the weight of each segment. The subject's entire torso and pelvis are supported by a cradle made of fabric, foam and sturdy plastic. Each subject wears a safety helmet to support the head and neck and to shield the face from potential hazards. A foam safety mat is placed on the ground beneath the subject to protect against any fall (Figure 2).

A closed-loop, force feedback-controlled linear motor SLD (LM-SLD) is used to provide near constant EL on the subject in the eZLS. In contrast, the current in-flight SLD used on the ISS treadmill applies EL through a passive torsional spring and has inherent stiffness characteristics, where force is dependent on EL cable vertical motion (how far the cable is extended or retracted into the SLD mechanism). The Series Bungee System (SBS bungees), used in-flight and in this study, also operate with some inherent stiffness. Because increased stiffness can result in an increase in EL variability due to vertical motion of the subject during locomotion, the constant-force model may improve exercise effectiveness and comfort. It is unclear, however, how a LM-SLD may influence locomotion patterns.

1.2 Purpose and Hypotheses

The primary purpose of this investigation was to determine how the EL mechanism affected locomotion on the eZLS. We hypothesized that when walking and running on the eZLS (controlling for static external load magnitude), locomotion kinematics, GRF, and lower extremity muscle activation patterns will differ between external load delivered by SBS bungees and an LM-SLD.

2.0 Methods

2.1 Human Subjects

Eight subjects (four men and four women) participated in this study (Table 1). To be eligible for the test subject pool, each subject passed a United States Air Force Class III-equivalent physical exam.

Table 1: Subject Demographics (Mean ± SD).

	Height (cm)	Weight (kg)	Age (yrs)
F (n=4)	159.4 ± 4.3	54.9 ± 7.4	33.5 ± 4.0
M (n=4)	175.9 ± 1.3	75.5 ± 6.2	34.0 ± 5.0
Total (n=8)	167.6 ± 9.3	65.2 ± 12.7	33.8 ± 4.2

The methodology of this investigation was reviewed and approved by the Johnson Space Center Committee for Protection of Human Subjects. Each subject was informed of the requirements of the study and the potential benefits and risks of participation. Each subject provided written informed consent prior to data collection, and each was free to withdraw from the study at any time. All trials were conducted in the Exercise Countermeasures Laboratory at NASA Glenn Research Center.

2.2 Data Collection

Each subject completed a locomotion session in simulated microgravity on the eZLS treadmill in the laboratory (Figure 2). During the trials, subjects wore Spandex running tights, a harness, and a protective helmet as motion capture, GRF, and electromyographical (EMG) data were collected.

Figure 2: Typical data collection on the eZLS.

2.3 Instrumentation

2.3.1 Motion Capture Data

Lower body and trunk kinematics were measured at 60 Hz with a multi-camera motion capture system (Smart Elite motion capture system, BTS Bioengineering Spa, Milanese, IT). The three-dimensional positions of reflective markers were recorded relative to an inertial reference frame established during calibration. A reference trial was collected after calibration, but before the subject arrived at the lab, to establish a treadmill reference frame.

Reflective markers were attached to each subject's left side. Markers were placed laterally on the neck, level with the fifth cervical vertebrae, the posterior heel on the rear of the running shoe, and on the tip of the shoe over the distal end of the second metatarsal. Additional markers were placed arbitrarily near the proximal and distal lateral tibia and femur to approximate the long axes of the lower and upper leg. A final marker was placed on the harness near the greater trochanter and was used along with the neck marker to approximate the long axis of the trunk.

A static trial was recorded prior to any locomotion trials. With the EL disengaged, subjects straightened their legs and locked their ankles and knees in the neutral position as an investigator gently pushed them to contact the treadmill. The static trial was used to determine the baseline positions of each joint angle.

2.3.2 Ground Reaction Forces

Vertical GRF data were collected at 960 Hz during eZLS testing trials with a force platform mounted beneath the treadmill belt (9287BA, Kistler CO, Amherst, NY). Prior to data collection each day, each subject was weighed to allow normalization of GRF data to body weight.

2.3.3 Electromyography

Telemetry EMG (Myomonitor III Wireless EMG System, Delsys Inc., Boston, MA) was used to obtain muscle activation data of the tibialis anterior, calf (over the medial head of the gastrocnemius), quadriceps (over the rectus femoris), medial hamstrings and gluteus maximus. The electrodes were not moved between trials, and data were collected at 1000 Hz with a fixed gain of 1000.

Each electrode was attached directly to the skin using double-sided adhesive tape specifically designed for use with the system. The area on the skin where the electrodes were to be attached was prepared with rubbing alcohol and fine sandpaper. Three bipolar single-differential electrodes were placed over each muscle belly (DE-2.3, Delsys Inc., Boston, MA). The electrode for each muscle that produced the largest signal was used during data collection and the additional two electrodes were removed.

All motion capture, GRF and EMG data were synchronized via a global analog pulse recorded simultaneously by each hardware device. Figure 3 depicts a typical testing setup for a subject on the eZLS.

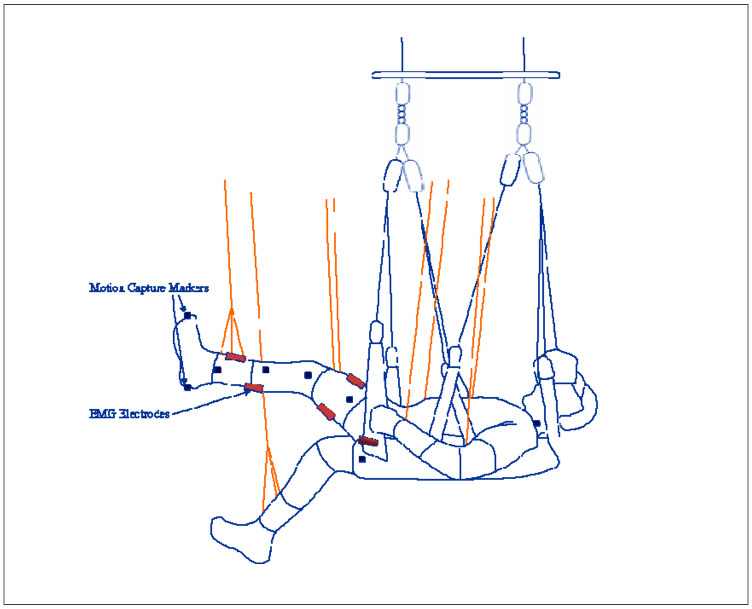

Figure 3: Typical subject setup on the eZLS, including motion capture marker and EMG electrode placement.

2.3.4 External Loading

Locomotion was completed at two EL levels. Subjects were loaded to approximately 55% (Low) and approximately 90% (High) of their bodyweight. The EL during this investigation was supplied by two methods: bungees and the LM-SLD. Elastomer bungees and carabiner clips similar to those currently used by astronauts onboard the ISS were arranged bilaterally to deliver EL. The EL magnitude delivered from bungees is dependent upon length, and whether single or dual bilateral bungees are used. Thus, the EL magnitude can be modified by arranging the clips and bungees in various configurations, but only a discrete number of loads are possible and are dependent upon the subject's leg length. In contrast, the LM-SLD system allows continuous initial load level setting and is not dependent upon the subject's leg length. The LM-SLD over-displacement limits are set and checked for safety prior to each test run, and the system also employs overspeed and overload protection for subject safety.

Subjects also completed trials while EL was delivered with the LM-SLD, a linear version with improved control algorithms and hardware than that used by Genc et al.[2] but conceptually based on the same closed-loop control approach. The LM-SLD (Figure 4) utilizes two linear servo motors (Trilogy Systems Corp., Houston, TX) with 45.7 cm (18 in.) stroke length that are controlled by a closed-loop force-feedback proportional-integral control. This device uses two in-line force transducers to provide a force feedback signal to the control system and a response measurement for measuring load directly on the LM-SLD to the subject harness; this allows the LM-SLD to maintain a relatively constant force on the subject.

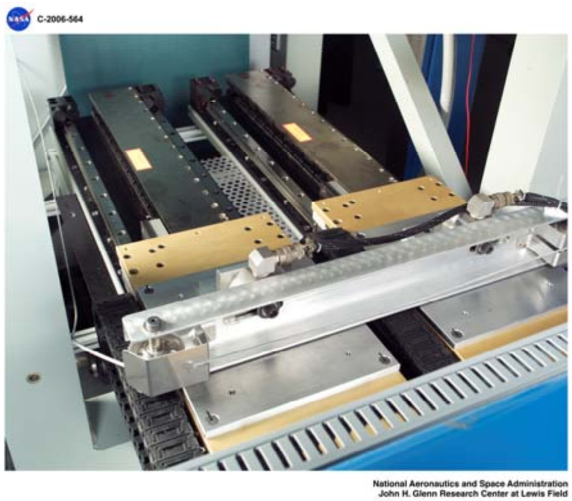

Figure 4: Linear motor subject loading device incorporated into the eZLS.

Table 2 shows the mean static EL values used for each loading condition. These were the loads measured as the subjects stood quietly on the force platform and do not account for any variation that may have occurred during actual locomotion.

Table 2: EL values for all loading conditions and types in microgravity and on the eZLS (Mean ± SD).

	External Load	
Load Type	Low (% body weight)	High (% body weight)
Bungees	58.0 ± 3.9	89.0 ± 4.2
LM-SLD	56.1 ± 4.5	89.6 ± 5.5

EL levels near the target loads were first found for each subject using bungees. The LM-SLD was then set to match the EL levels achieved with the bungees. This procedure was used because variation of the EL with the LM-SLD could be established much more quickly than with the bungee and clips. Subjects wore the Cleveland Clinic prototype harness that is similar in functionality to the waist and shoulder harnesses currently used by astronauts during exercise on the ISS treadmill.

2.4 Experimental Protocol

Subjects walked at 1.34 $m \cdot s^{-1}$ (3 mph) and ran at 3.13 $m \cdot s^{-1}$ (7 mph) during each EL and loading mechanism condition. Therefore, subjects completed eight trials (two speeds × two EL × two loading mechanisms). Subjects completed one 60-second trial for each discrete condition.

Walking trials were always completed before running trials at each EL. However, EL level and type were randomized across subjects. A balanced randomization was used to ensure that testing orders were different for each subject. Assignment of the EL level order was made arbitrarily with a coin toss.

2.5 Data Analysis

The first 10 strides of the left leg were analyzed for each trial. The chosen epoch began with the first heel strike of the left foot, and ended with the eleventh heel strike of the left foot. Software programs written in MATLAB Version 7.2.0.232 (R2006a; Natick, MA) were used for the entire analysis. Processing was completed, separately, on the GRF, motion capture, and EMG data.

2.5.1 Ground Reaction Force

GRF data were analyzed to determine the number of times that the left foot was in contact with the treadmill belt. Center of pressure coordinates were computed and analyzed to determine if the foot in contact with the treadmill was on the right or left side of the belt. Once the left footfalls were identified, contact time, stride time, peak impact force, peak propulsive force, average loading rate, and impulse were found for each step.

Heel strike and toe off were found as described by Chang, et al..[6] Contact time was the length of time that the left foot was in contact with the treadmill belt during each stride, and was found as the duration between heel strike and toe off for each footfall. Stride time was the length of time between successive heel strikes of the left foot. Peak impact force was the magnitude of the first distinct peak in the ground reaction force trajectory. Peak propulsive force was the magnitude of the second distinct peak. Loading rate was the peak impact force divided by the time between heel strike and time of peak impact force. The impulse for each footfall was computed as the integral of the ground reaction force trajectory over contact time. Peak impact force, loading rate, peak propulsive force and impulse were all normalized to body weight to allow inter-subject comparisons (Figure 5).

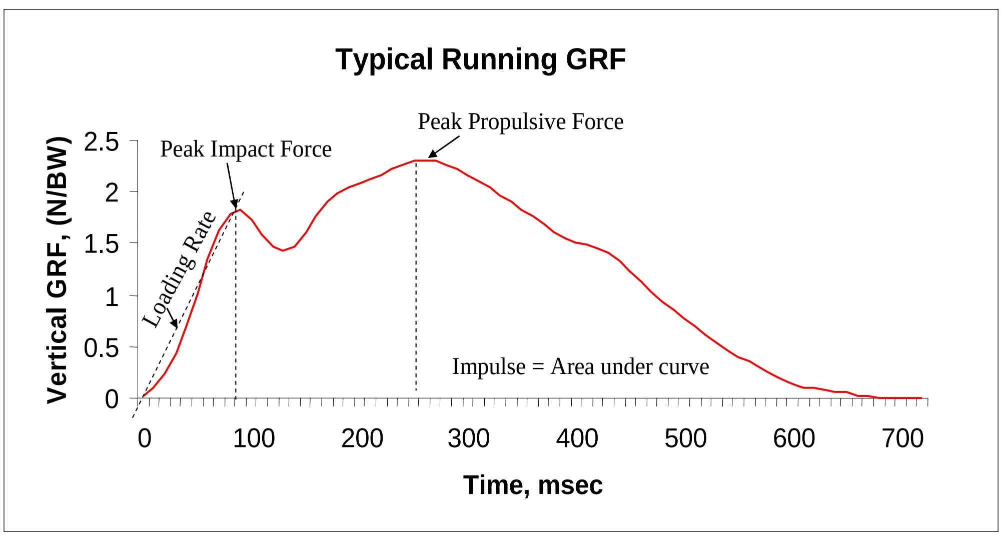

Figure 5: Typical running ground reaction force and dependent variables.

2.5.2 *Motion Capture*

Raw motion capture data were examined for missing points, which were replaced using cubic spline interpolation. The motion capture data were then filtered using a fourth-order Butterworth low-pass filter with an optimal cutoff frequency for each marker determined using an autocorrelation procedure.[7] The autocorrelation was executed independently for each coordinate of each marker, and the highest cutoff frequency determined for each coordinate was used for each marker. Cutoff frequencies ranged from 6-28 Hz (mean=16.65 Hz).

Joint angle trajectories of the ankle, knee and hip were found for each sample of every trial. Hip angle was defined as the angle separating the thigh and trunk. Knee angle was defined as the angle separating the shank and thigh. Positive hip and knee angles indicated flexion. Ankle angle was found as the angle separating the shank and foot segments. In order to relate the ankle angle to the anatomical angle, 90 degrees was subtracted from the computed ankle angle. Positive ankle angles represented plantarflexion of the left foot. All joint angles were corrected relative to the anatomical position by subtracting joint angles found during a static (standing) trial.

Segment angle trajectories were also found for the foot, shank, thigh and trunk. Segment angles were the angles separating each segment from the reference frame axis directed normal to the treadmill surface. These angles provide a measure of each segment’s orientation in space independent of the orientation of the adjacent segment.

2.5.3 *Electromyography*

EMG data were processed according the methods of Browning, Modica, Kram and Goswami.[8] Initially, the direct current offset was removed from each signal by subtracting the mean from the entire channel. The data were then filtered with a zero-lag fourth-order Butterworth band-pass filter of 16-499 Hz. The filtered data were full wave rectified, and then filtered with a Butterworth low-pass with a cutoff frequency of 7 Hz. Each signal was then objectively examined for a period where muscle activity appeared to be absent to establish a baseline. The mean and standard deviation of the signal during this time period was found. The threshold for muscle activity was established to be three standard deviations greater than the mean. Muscle activity was defined to occur when the signal was greater than the threshold for at least 100 ms. Similarly, if the signal fell below the threshold for at least 100 ms, the muscle was defined as inactive. The analysis procedure was completed independently for each muscle within each trial to minimize the possibility of applying an inadequate threshold to a channel. Muscle activation onset and duration were found and time-normalized to each stride.

Once all activations were identified, each stride for each trial was examined to determine the time and duration of the first activation of each muscle. There were trials in which more than one activation was found for each stride, although there was no consistent trend across subjects. In order to account for the fact that multiple activations may have occurred during a single stride, the mean duration of each activation was found. Finally, the total activation time for each stride was computed.

2.6 Statistical Analysis

The mean of each dependent variable over 10 strides was found for each trial. Statistical analyses were conducted utilizing NCSS 2004 statistical software (NCSS, Kaysville, Utah). Two separate, but related, hypotheses were tested during this evaluation. Walking and running were analyzed separately because they are distinct locomotive tasks that require different kinematics.

We used a two-factor repeated measures analysis of variance (ANOVA) with EL load type and EL level as main effects. We were interested in determining the differences in locomotion due to loading mechanism within EL level. Therefore, Tukey-Kramer Multiple Comparisons tests were examined to determine EL type effects within EL level. We only reported a main effect if we found a post-hoc difference between load type after finding a significant overall load type effect. In both evaluations, statistical significance was achieved at an alpha level of 0.05.

3.0 Results

The purpose of this study was to determine the similarities and differences between locomotion on the eZLS with the EL delivered by bungees and a linear motor SLD. Kinematic, GRF and EMG comparisons are presented in the following section.

During each trial, test termination criteria included a restriction on each subject’s heart rate. A chest-band wireless heart rate monitor (Polar Electro Inc., Lake Success, NY) was worn by each subject. If heart rate exceeded 85% of the subject’s age-predicted maximum for more than 30 seconds, the trial was terminated. Very few trials were affected, and all terminations occurred during running with the high EL.

3.1 Kinematics

3.1.1 Joint Range of Motion

Figure 6 shows the average trajectories of the hip, knee and ankle for a single subject. Table 3 shows the mean hip, knee and ankle angles during walking and running for the group. The only significant difference between loading type occurred in hip flexion angle during walking. Greater hip flexion was attained with the bungees during the Low condition.

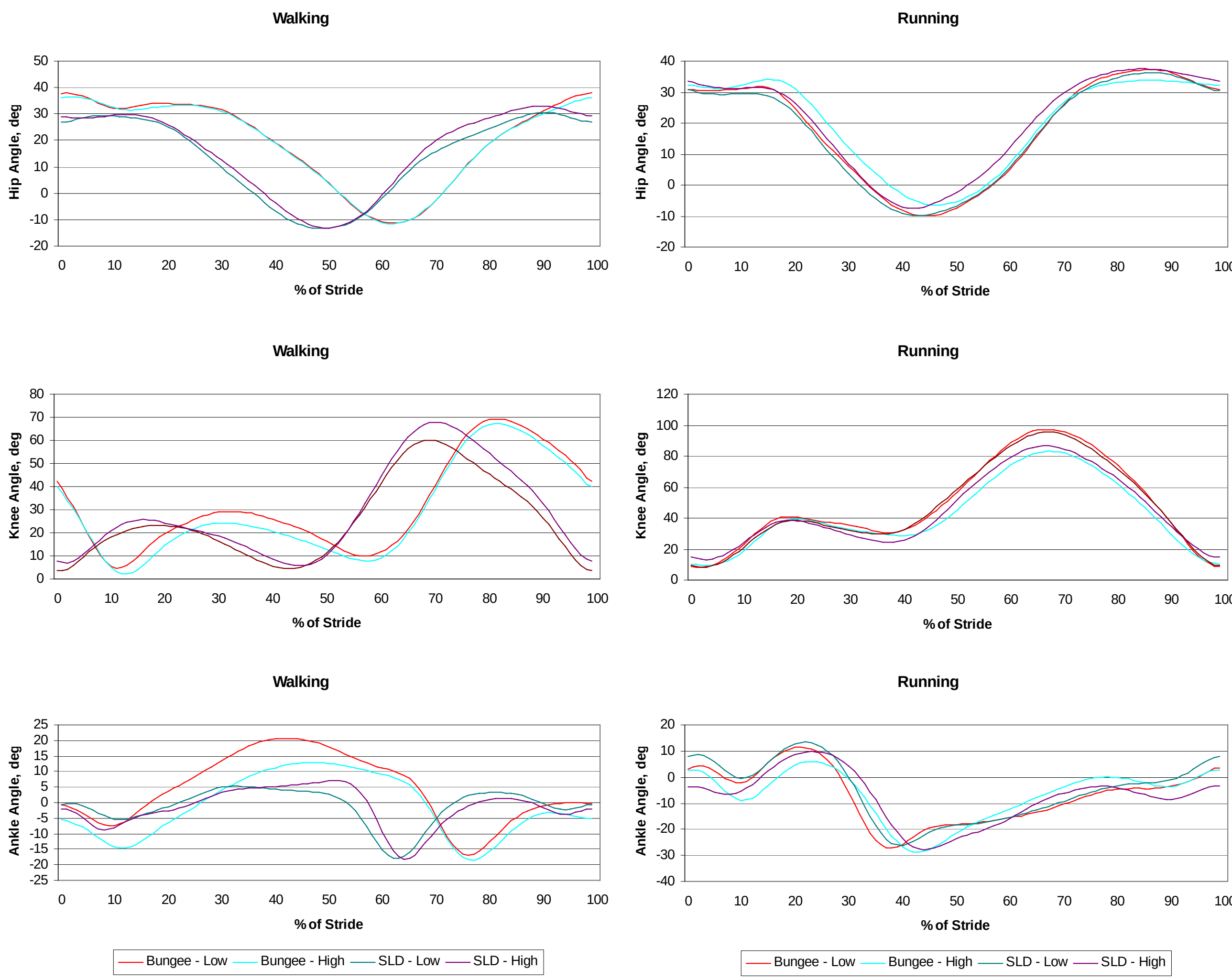

Figure 6: Joint ensemble average trajectories for a single stride for a single subject during walking and running on the eZLS while loaded with bungees and LM-SLD.

Table 3: Hip, knee and ankle minimum, maximum and ROM angles during walking and running on the eZLS while loaded with bungees and the eZLS while loaded with the LM-SLD at low and high EL (Mean ± SD).

		Hip Angle (deg)			Knee Angle (deg)			Ankle Angle (deg)		
					Walking					
Level	Load Type	Min	Max	ROM	Min	Max	ROM	Min	Max	ROM
Low										
	Bungee	-18.19 ± 7.73	28.85 ± 6.18*	47.04 ± 6.61	-0.25 ± 6.77	61.27 ± 6.35	61.52 ± 4.87	-19.63 ± 4.19	11.00 ± 8.36	30.63 ± 6.66
	LM-SLD	-20.68 ± 9.26	25.87 ± 7.72	46.55 ± 1.97	-1.13 ± 6.87	56.76 ± 4.97	57.89 ± 4.22	-23.42 ± 4.05	6.37 ± 5.15	29.79 ± 6.83
High										
	Bungee	-20.04 ± 7.57	30.44 ± 5.10	50.48 ± 4.66	-1.95 ± 6.16	58.51 ± 5.31	60.46 ± 4.78	-20.72 ± 5.11	7.60 ± 4.30	28.31 ± 3.53
	LM-SLD	-21.43 ± 9.39	28.87 ± 6.86	50.30 ± 3.68	-1.42 ± 7.65	59.32 ± 5.53	60.74 ± 4.04	-22.57 ± 4.86	6.59 ± 3.59	29.16 ± 4.46
					Running					
Level	Location	Min	Max	ROM	Min	Max	ROM	Min	Max	ROM
Low										
	Bungee	-18.30 ± 7.02	31.43 ± 4.68	49.73 ± 5.45	3.79 ± 5.52	90.92 ± 10.11	87.13 ± 13.63	-26.94 ± 4.54	17.58 ± 5.72	44.51 ± 9.41
	LM-SLD	-18.70 ± 7.12	31.62 ± 3.87	50.32 ± 4.16	3.24 ± 4.12	91.71 ± 8.75	88.47 ± 11.17	-28.51 ± 6.71	17.05 ± 3.93	45.56 ± 9.22
High										
	Bungee	-17.40 ± 7.77	32.42 ± 5.98	49.82 ± 4.76	8.25 ± 7.29	89.10 ± 7.84	80.85 ± 13.95	-28.90 ± 4.10	16.97 ± 5.14	45.87 ± 7.14
	LM-SLD	-18.50 ± 11.20	31.43 ± 6.58	49.92 ± 5.74	6.43 ± 7.46	86.32 ± 6.32	79.89 ± 11.93	-27.65 ± 3.84	17.49 ± 6.17	45.14 ± 6.99

*LM-SLD and bungee loading conditions significantly different within load level, $p<0.05$

3.1.2 Segment Range of Motion

Table 4 shows the mean trunk, thigh, shank and foot angular minimum, maximum and ROM during walking. There were no differences between EL types within EL levels in any measure with the exception of minimum thigh angle. The minimum thigh angle was less in the Low condition with bungees than with the LM-SLD.

Table 4: Trunk, thigh, shank and foot minimum, maximum and ROM angles during walking on the eZLS while loaded with bungees and with the LM-SLD at low and high EL (Mean ± SD).

		Trunk Angle (deg)			Thigh Angle (deg)		
Level	EL Type	Min	Max	ROM	Min	Max	ROM
Low							
	Bungee	0.05 ± 1.40	1.59 ± 1.32	1.54 ± 0.37	-27.98 ± 5.40*	18.81 ± 7.61	46.79 ± 6.15
	LM-SLD	0.32 ± 1.21	1.66 ± 1.15	1.35 ± 0.17	-24.98 ± 7.18	21.52 ± 8.81	46.50 ± 2.12
High							
	Bungee	-0.43 ± 0.80	1.20 ± 0.77	1.63 ± 0.27	-29.94 ± 4.53	19.94 ± 7.26	49.88 ± 4.76
	LM-SLD	0.07 ± 1.02	1.67 ± 1.07	1.60 ± 0.19	-28.00 ± 6.37	21.94 ± 9.19	49.94 ± 3.85

		Shank Angle (deg)			Foot Angle (deg)		
Level	EL Type	Min	Max	ROM	Min	Max	ROM
Low							
	Bungee	-22.40 ± 4.60	53.18 ± 4.31	75.59 ± 2.31	-24.86 ± 3.19	70.72 ± 3.94	95.59 ± 6.33
	LM-SLD	-22.17 ± 3.08	52.24 ± 3.84	74.41 ± 1.99	-25.41 ± 2.14	73.91 ± 5.19	99.32 ± 6.09
High							
	Bungee	-25.08 ± 3.54	49.99 ± 2.21	75.08 ± 3.14	-24.94 ± 4.56	70.16 ± 5.47	95.10 ± 7.49
	LM-SLD	-22.62 ± 3.64	51.27 ± 4.73	73.88 ± 3.55	-25.19 ± 3.10	72.83 ± 5.25	98.01 ± 6.38

*LM-SLD and bungee loading conditions significantly different within load level, $p<0.05$

Table 5 shows the mean trunk, thigh, shank and foot angular minimum, maximum and ROM during running. There were no EL type effects within EL levels in any variable.

Table 5: Trunk, thigh, shank and foot minimum, maximum and ROM angles during running on the eZLS while loaded with bungees and with the LM-SLD at low and high EL (Mean ± SD).

		Trunk Angle (deg)			Thigh Angle (deg)		
Level	EL Type	Min	Max	ROM	Min	Max	ROM
Low							
	Bungee	-0.05 ± 1.56	2.81 ± 1.40	2.86 ± 1.39	-30.56 ± 4.43	20.66 ± 6.45	51.22 ± 5.73
	LM-SLD	-0.03 ± 1.64	2.27 ± 1.30	2.30 ± 0.88	-31.04 ± 3.62	20.61 ± 6.12	51.65 ± 4.23
High							
	Bungee	-1.10 ± 1.33	3.59 ± 2.73	4.68 ± 2.65	-31.93 ± 4.68	19.59 ± 6.42	51.52 ± 5.11
	LM-SLD	-0.74 ± 1.50	3.53 ± 1.85	4.27 ± 1.50	-30.95 ± 6.35	21.00 ± 10.31	51.95 ± 5.87
		Shank Angle (deg)			Foot Angle (deg)		
Level	EL Type	Min	Max	ROM	Min	Max	ROM
Low							
	Bungee	-20.41 ± 5.41	84.16 ± 10.11	104.58 ± 10.69	-28.60 ± 5.05	100.73 ± 11.77	129.34 ± 14.14
	LM-SLD	-21.10 ± 4.38	84.23 ± 10.51	105.33 ± 9.56	-28.77 ± 5.48	101.78 ± 11.79	130.55 ± 11.54
High							
	Bungee	-19.42 ± 5.58	80.27 ± 9.22	99.69 ± 11.41	-26.79 ± 6.49	96.56 ± 11.76	123.35 ± 14.03
	LM-SLD	-19.88 ± 3.68	77.86 ± 10.42	97.73 ± 11.20	-27.07 ± 6.24	93.46 ± 10.21	120.53 ± 13.51

3.1.3 Contact and Stride Time

Table 6 shows the contact times and stride times for each loading type and EL level for walking and running on the eZLS. There were no effects of load type on either dependent variable at any EL level.

Table 6: Contact time and stride time during walking and running on the eZLS with bungees and the LM-SLD at low and high EL (Mean ± SD).

		Walking	
Level	EL Type	Contact Time (s)	Stride Time (s)
Low			
	Bungee	0.82 ± 0.09	1.10 ± 0.09
	LM-SLD	0.83 ± 0.08	1.08 ± 0.07
High			
	Bungee	0.84 ± 0.08	1.07 ± 0.08
	LM-SLD	0.80 ± 0.07	1.06 ± 0.07
		Running	
Low			
	Bungee	0.33 ± 0.04	0.91 ± 0.08
	LM-SLD	0.34 ± 0.02	0.90 ± 0.06
High			
	Bungee	0.36 ± 0.09	0.82 ± 0.06
	LM-SLD	0.34 ± 0.08	0.80 ± 0.05

3.2 Ground Reaction Forces

Table 7 shows the GRF dependent variables on the eZLS. Peak impact forces during walking were larger with the LM-SLD than with bungees at both EL levels. Peak propulsive forces were greater with the bungees at the Low condition, but greater with the LM-SLD at the High condition. In addition, loading rate was greater with the LM-SLD than the bungees during the High trials. During High running, peak propulsive forces and impulse were greater when EL was supplied by the LM-SLD than by bungees.

Table 7: Peak impact force, peak propulsive force, loading rate and impulse during walking and running on the eZLS at low and high EL (Mean ± SD).

Level	EL Type	Peak Impact Force (BW)	Peak Propulsive Force (BW)	Loading Rate (BW/s)	Impulse (BW·msec)
		Walking			
Low					
	Bungee	0.93 ± 0.06*	0.59 ± 0.09*	9.55 ± 3.49	290.47 ± 34.23
	LM-SLD	1.12 ± 0.07	0.51 ± 0.11	11.21 ± 3.33	276.40 ± 116.52
High					
	Bungee	1.09 ± 0.06*	0.85 ± 0.09*	8.93 ± 2.34*	426.56 ± 29.62
	LM-SLD	1.38 ± 0.07	0.90 ± 0.07	11.53 ± 2.21	482.00 ± 50.16
		Running			
Low					
	Bungee	1.39 ± 0.25	1.56 ± 0.14	29.42 ± 13.10	240.53 ± 16.78
	LM-SLD	1.31 ± 0.24	1.65 ± 0.13	26.45 ± 4.63	258.82 ± 19.05
High					
	Bungee	1.78 ± 0.17	1.85 ± 0.31*	40.44 ± 8.11	304.91 ± 38.49*
	LM-SLD	1.82 ± 0.14	2.12 ± 0.29	32.17 ± 16.47	358.90 ± 30.37

*LM-SLD and bungee loading conditions significantly different within load level, $p<0.05$

3.3 Electromyographical

Typical muscle activation patterns for each muscle are shown in Figures 7-10. Each pattern is depicted along with the linear envelope used to define the presence or absence of muscle activity. All plots are normalized to a single stride.

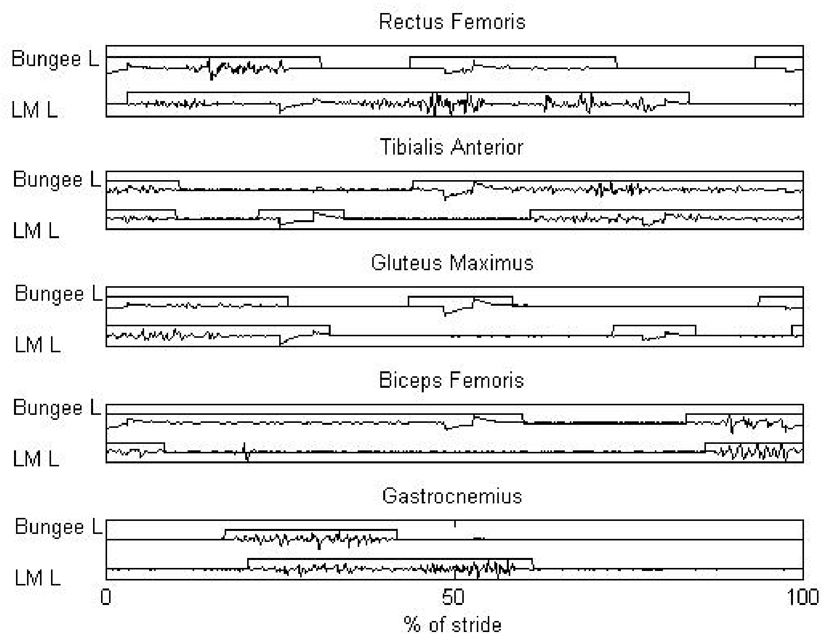

Figure 7: Typical EMG activity for the lower extremity muscles during walking on the eZLS with a low EL.

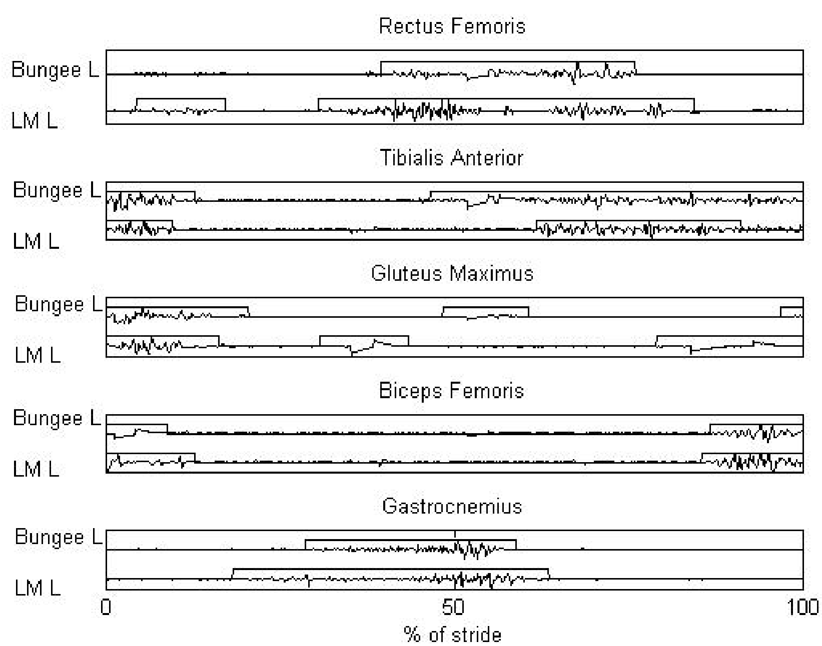

Figure 8: Typical EMG activity for the lower extremity muscles during walking on the eZLS with a high EL.

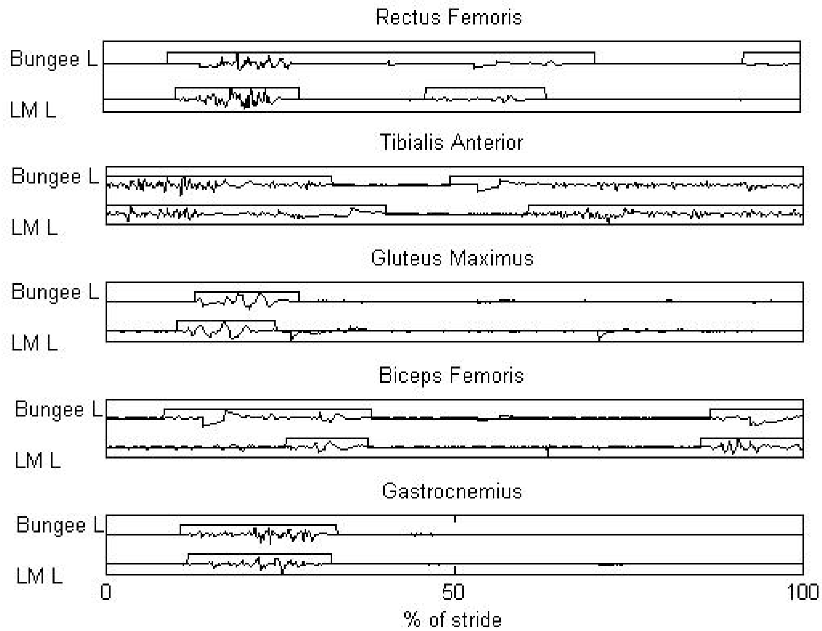

Figure 9: Typical EMG activity for the lower extremity muscles during running on the eZLS with a low EL.

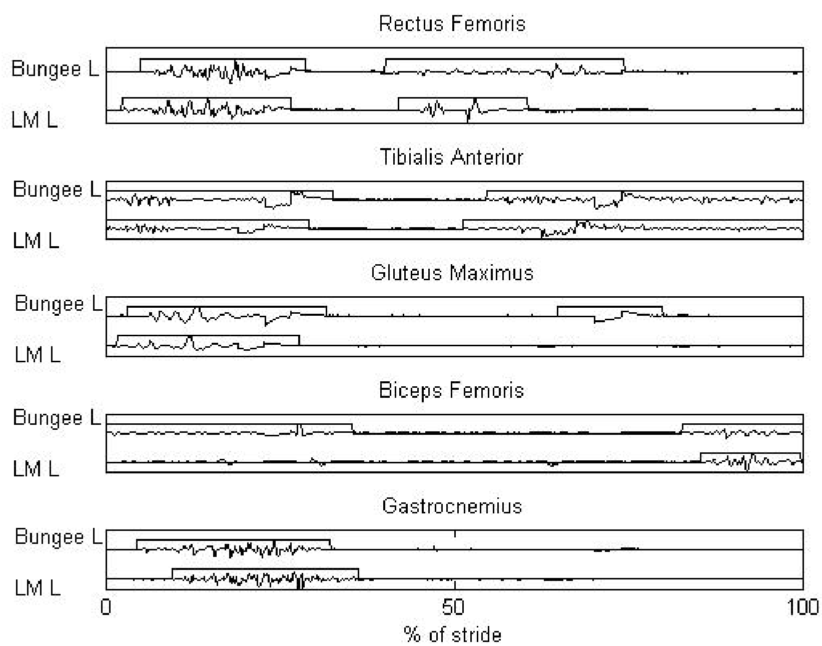

Figure 10: Typical EMG activity for the lower extremity muscles during running on the eZLS with a high EL.

Tables 8-12 show the first activation time, duration of the first activation, mean duration per activation and mean total duration of activation for the rectus femoris, tibialis anterior, gluteus maximus, biceps femoris and gastrocnemius during walking and running on the eZLS with EL provided by bungees and the LM-SLD. For each muscle, there were no significant differences in any dependent variable between EL types at either load during walking or running.

Table 8: Rectus femoris activity during walking and running on the eZLS at low and high EL with bungees and the LM-SLD at low and high EL (Mean ± SD).

		First Activation (% of stride)	Duration of First Activation (% of stride)	Mean Duration per Activation (% of stride)	Mean Total Duration of Activation (% of stride)
Level	EL Type	Walking			
Low					
	Bungee	0.24 ± 0.16	0.26 ± 0.10	0.25 ± 0.09	0.36 ± 0.19
	LM-SLD	0.38 ± 0.21	0.42 ± 0.13	0.41 ± 0.16	0.59 ± 0.18
High					
	Bungee	0.34 ± 0.24	0.46 ± 0.16	0.44 ± 0.15	0.54 ± 0.18
	LM-SLD	0.33 ± 0.24	0.41 ± 0.23	0.40 ± 0.22	0.54 ± 0.17
		Running			
Low					
	Bungee	0.42 ± 0.12	0.42 ± 0.22	0.42 ± 0.19	0.57 ± 0.14
	LM-SLD	0.42 ± 0.15	0.34 ± 0.19	0.35 ± 0.17	0.54 ± 0.16
High					
	Bungee	0.41 ± 0.17	0.32 ± 0.15	0.31 ± 0.14	0.47 ± 0.16
	LM-SLD	0.41 ± 0.13	0.41 ± 0.18	0.40 ± 0.17	0.55 ± 0.14

Table 9: Tibialis anterior activity during walking and running on the eZLS at low and high EL with bungees and the LM-SLD at low and high EL (Mean ± SD).

Level	EL Type	First Activation (% of stride)	Duration of First Activation (% of stride)	Mean Duration per Activation (% of stride)	Mean Total Duration of Activation (% of stride)
			Walking		
Low					
	Bungee	0.31 ± 0.29	0.62 ± 0.11	0.61 ± 0.12	0.53 ± 0.17
	LM-SLD	0.23 ± 0.06	0.54 ± 0.04	0.53 ± 0.07	0.58 ± 0.06
High					
	Bungee	0.21 ± 0.07	0.50 ± 0.15	0.48 ± 0.16	0.56 ± 0.10
	LM-SLD	0.22 ± 0.03	0.51 ± 0.10	0.50 ± 0.12	0.53 ± 0.07
			Running		
Low					
	Bungee	0.62 ± 0.29	0.57 ± 0.25	0.58 ± 0.25	0.57 ± 0.23
	LM-SLD	0.52 ± 0.34	0.65 ± 0.16	0.65 ± 0.15	0.68 ± 0.08
High					
	Bungee	0.78 ± 0.28	0.64 ± 0.13	0.64 ± 0.13	0.63 ± 0.13
	LM-SLD	0.78 ± 0.29	0.71 ± 0.05	0.71 ± 0.05	0.66 ± 0.06

Table 10: Gluteus maximus activity during walking and running on the eZLS at low and high EL with bungees and the LM-SLD at low and high EL (Mean ± SD).

Level	EL Type	First Activation (% of stride)	Duration of First Activation (% of stride)	Mean Duration per Activation (% of stride)	Mean Total Duration of Activation (% of stride)
			Walking		
Low					
	Bungee	0.28 ± 0.11	0.20 ± 0.05	0.21 ± 0.04	0.37 ± 0.06
	LM-SLD	0.30 ± 0.13	0.22 ± 0.09	0.24 ± 0.08	0.40 ± 0.08
High					
	Bungee	0.35 ± 0.16	0.20 ± 0.05	0.22 ± 0.04	0.37 ± 0.12
	LM-SLD	0.35 ± 0.15	0.24 ± 0.11	0.23 ± 0.08	0.37 ± 0.15
			Running		
Low					
	Bungee	0.40 ± 0.19	0.27 ± 0.13	0.28 ± 0.17	0.32 ± 0.10
	LM-SLD	0.44 ± 0.17	0.27 ± 0.16	0.27 ± 0.18	0.32 ± 0.20
High					
	Bungee	0.40 ± 0.12	0.34 ± 0.15	0.33 ± 0.15	0.35 ± 0.14
	LM-SLD	0.32 ± 0.10	0.37 ± 0.18	0.36 ± 0.17	0.43 ± 0.15

Table 11: Biceps femoris activity during walking and running on the eZLS at low and high EL with bungees and the LM-SLD at low and high EL (Mean ± SD).

Level	EL Type	First Activation (% of stride)	Duration of First Activation (% of stride)	Mean Duration per Activation (% of stride)	Mean Total Duration of Activation (% of stride)
			Walking		
Low					
	Bungee	0.36 ± 0.10	0.32 ± 0.19	0.35 ± 0.18	0.50 ± 0.14
	LM-SLD	0.36 ± 0.15	0.30 ± 0.13	0.30 ± 0.13	0.42 ± 0.14
High					
	Bungee	0.38 ± 0.10	0.26 ± 0.09	0.28 ± 0.08	0.42 ± 0.18
	LM-SLD	0.44 ± 0.10	0.25 ± 0.11	0.26 ± 0.10	0.28 ± 0.11
			Running		
Low					
	Bungee	0.28 ± 0.13	0.49 ± 0.14	0.50 ± 0.11	0.58 ± 0.12
	LM-SLD	0.27 ± 0.12	0.53 ± 0.16	0.53 ± 0.16	0.61 ± 0.13
High					
	Bungee	0.25 ± 0.12	0.45 ± 0.17	0.47 ± 0.16	0.55 ± 0.13
	LM-SLD	0.27 ± 0.11	0.40 ± 0.19	0.40 ± 0.19	0.47 ± 0.16

Table 12: Gastrocnemius activity during walking and running on the eZLS at low and high EL with bungees and the LM-SLD at low and high EL (Mean ± SD).

Level	EL Type	First Activation (% of stride)	Duration of First Activation (% of stride)	Mean Duration per Activation (% of stride)	Mean Total Duration of Activation (% of stride)
			Walking		
Low					
	Bungee	0.45 ± 0.24	0.34 ± 0.26	0.35 ± 0.26	0.40 ± 0.24
	LM-SLD	0.56 ± 0.17	0.42 ± 0.25	0.44 ± 0.23	0.52 ± 0.25
High					
	Bungee	0.56 ± 0.21	0.34 ± 0.15	0.35 ± 0.15	0.42 ± 0.12
	LM-SLD	0.54 ± 0.27	0.39 ± 0.18	0.40 ± 0.17	0.46 ± 0.19
			Running		
Low					
	Bungee	0.45 ± 0.19	0.34 ± 0.09	0.33 ± 0.09	0.40 ± 0.14
	LM-SLD	0.53 ± 0.16	0.31 ± 0.06	0.31 ± 0.06	0.33 ± 0.09
High					
	Bungee	0.51 ± 0.12	0.36 ± 0.12	0.36 ± 0.12	0.37 ± 0.13
	LM-SLD	0.53 ± 0.10	0.31 ± 0.07	0.31 ± 0.07	0.32 ± 0.07

4.0 Discussion

The primary purpose of this study was to determine the differences in locomotion when EL is supplied with bungees or a linear motor SLD on the eZLS. There were no kinematic or muscular activity effects due to EL type, except for hip flexion angle during walking. However, multiple GRF variables were affected by loading type. Peak impact force during walking was greater with the LM-SLD at both load levels. Peak propulsive force during walking was greater at the Low level with bungees, but greater with the LM-SLD at the High level. During running, peak propulsive force was greater with the LM-SLD at the High level. Loading rate during walking was greater with the LM-SLD at the Low level. Impulse during running was greater with the LM-SLD at high loads.

4.1 Summary of Results

Multiple dependent variables were examined throughout this investigation. Table 13 summarizes the dependent variable results during the EL type comparison on the eZLS. All differences noted were statistically significant.

Table 13: Summary of the significant main effects of EL type within EL level upon all dependent variables during walking and running ($p<0.05$). B=Bungees.

	Walking		Running	
	Low	High	Low	High
Kinematics				
Hip Angle	Flexion: B>LM-SLD			
Thigh Angle	Min: B>LM-SLD			
Ground Reaction Forces				
Peak Impact Force	B<LM-SLD	B<LM-SLD		
Loading Rate		B<LM-SLD		
Peak Propulsive Force	B>LM-SLD	B<LM-SLD		B<LM-SLD
Impulse				B<LM-SLD

4.2 Effect of External Loading Mechanism Upon Locomotion

During the eZLS locomotion trials, EL was applied with two different mechanisms. Elastomer bungees, which are also currently used by crew members as a contingency loading device onboard the ISS, and the LM-SLD mechanism available to the eZLS.

The advantage of the bungees is that they are passive, simple and easy to use. They are essentially strong rubber bands that clip between the harness and the treadmill. Their principal weakness is that they provide a specific tension that is related to their stiffness and length, which limits the available levels of EL. Schaffner et al.[1] found the mean stiffness of bungees measured during locomotion in microgravity to be approximately 3 kg/cm. This stiffness results in potential EL variations of up to 10% of body weight.

Bungee loads are directly related to their length based on their stiffness. The distance between the center of mass and ground will be shortest during midstance.[9] Therefore, subjects will have to overcome a reduced EL to propel themselves upward from the treadmill. The reduction in EL may explain why peak propulsive GRF are less in microgravity than during normal locomotion.

The LM-SLD, as opposed to bungees, can provide a user-defined EL with considerably less stiffness. A characteristic of the LM-SLD is that it is active and, therefore, requires some electrical power to operate. In addition, the LM-SLD is a complex device that may be difficult to repair if a breakdown occurs.

The closed-loop force-feedback control of the LM-SLD will allow the EL to vary much less than with bungees, regardless of the length of the tether connecting the subject's harness to the LM-SLD. Therefore, as the bungee load decreases during midstance, the decrease in LM-SLD EL, if it occurs, will be much less. Our results indicate that peak impact and propulsive GRF during walking are greater with the LM-SLD than with bungees, and peak propulsive forces are greater during running with the LM-SLD. The lower variation in EL is probably the main reason for the increased GRF, though our results indicate that kinematics and muscle activation patterns are not affected.

Our results show that during locomotion on the eZLS, greater GRF were generated when the EL was supplied with the LM-SLD than with bungees, without appreciably affecting lower limb and trunk kinematics. There was a difference in maximum hip flexion during walking between devices of about 3 degrees, which we feel to be clinically insignificant. Researchers should expect greater GRF, and potentially a greater physiological benefit, when the LM-SLD is the chosen EL mechanism during eZLS exercise.

Although we did not measure GRF during upright locomotion in Earth's gravity, Schaffner et al.[1] reported peak impact forces during upright walking and running to be 1.14BW and 1.80BW. Peak propulsive forces were 1.09 BW and 2.37BW, respectively. They investigated locomotion at the same speeds as in our study. Our results suggest that the GRF developed during walking at either load and running with the low EL on the eZLS is less than that obtained during upright locomotion. However, during running with the high EL supplied with the LM-SLD, the peak propulsive force (2.12BW) was much closer to upright values than during the bungee conditions (1.85BW).

Schaffner et al. [1] also investigated walking and running in microgravity with EL levels of 60%, 80% and 90% BW supplied by bungees. With EL levels of 60% BW, peak impact and propulsive GRF magnitudes were 0.93BW and .062BW for walking and 1.33BW and 1.60BW for running. When the EL was 90% BW, they found peak impact and propulsive forces of 1.22BW and 0.90BW for walking and 1.50BW and 1.94BW for running. Their findings with bungee-loading in microgravity are very similar to our findings with bungee-loading on the eZLS.

We have no reason to believe that exercise in actual microgravity would result in an interaction effect upon the differences in GRF magnitudes between bungees and the LM-SLD. Therefore, we can reasonably speculate that the utilization of the LM-SLD during exercise in microgravity will result in an increase in GRF over those which are currently experienced by ISS crew members. Research and development of a flight-certified LM-SLD may be of benefit for the cardiovascular and bone health of future crew members during long-duration space flight.

4.3 Limitations

A primary limitation in this study was the use of multiple ANOVAs to examine the large number of dependent variables. This occurrence was unavoidable given the infrequent opportunities to capture data on microgravity locomotion simulators. Because we used many repeated measures ANOVAs to test differences in the means of the dependent variables between loading type, it is possible that the chance of Type I error increased. A Type I error occurs when a null hypothesis is rejected that should have been accepted. It is possible that we found statistical differences that were due to chance rather than to a true difference in means. Therefore, the variables in which significant effects of load type were found should be regarded with some caution. However, given the statistical significance value of $p<0.05$ combined with our use of the Tukey-Kramer Multiple Comparison tests, the chance of this error occurring was minimized. In addition, the use of the repeated measures ANOVA assumes that samples are normally distributed and that the variances are equal. It is possible that violations of the repeated measures ANOVA assumptions occurred given our small sample size.

4.4 Summary and Conclusions

The specific purpose of this investigation was to assess the differences between loading with bungees and a linear motor SLD on the eZLS. We found that providing EL with a linear motor SLD results in greater peak impact and peak propulsive forces, greater loading rates, and a greater impulse than when loaded with a bungee. EL type does not appear to appreciably affect kinematics or lower extremity muscle activity.

We were not able to compare locomotion with the LM-SLD in true microgravity to that on eZLS. Therefore, it is unknown if the differences that occurred between bungees and the LM-SLD extends to exercise in microgravity. However, LM-SLD loading on the eZLS during running results in GRF magnitudes nearer to those obtained during upright running. If increasing GRF magnitudes is an operational goal of astronaut health personnel, the use of the LM-SLD or other constant-force device may be beneficial for obtaining greater GRF.

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13. ABSTRACT *(Maximum 200 words)*

Purpose: Determine how External Load (EL) type affects locomotion patterns and muscular activity on the enhanced Zero-gravity Locomotion Simulator (eZLS) when using bungees or a linear motor subject loading device.

Eight subjects were suspended on the eZLS while walking at 3 mph and running at 7 mph. The EL was provided by either bungees or a linear motor subject loading device (LM-SLD) at approximately 55% and 90% of body weight during the exercise. Joint kinematics, ground reaction forces (GRF), and electromyographical activity of lower body musculature were measured during each condition. Repeated measures analysis of variance were tested for differences between EL types within load levels on eZLS.

There were few differences in locomotion patterns and muscular activity between loading mechanisms. GRF were greater with the LM-SLD than with bungees during eZLS locomotion. GRF magnitudes for both devices were lower than previously reported values obtained during upright locomotion in normal gravity, but similar to those found in actual microgravity.

Greater GRF with the LM-SLD suggests that use of a constant-force SLD may be of potential benefit during treadmill exercise because locomotion patterns do not change, but subjects experience increased force magnitude and loading rates applied at the feet.

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