



Article

Effects of Simulated Hyper-Gravity on Lower Limb Kinematics and Electromyography During Walking

Christopher A. Malaya ^{1,2,*} , Pranav J. Parikh ², Dean L. Smith ³ and Charles S. Layne ²

¹ Grail Laboratory, Parker University, Dallas, TX 75229, USA

² Center for Neuromotor and Biomechanics Research, Department of Health and Human Performance, University of Houston, Houston, TX 77204, USA; pparikh2@uh.edu (P.J.P.); clayne2@central.uh.edu (C.S.L.)

³ Department of Kinesiology, Nutrition, and Health, Miami University, Oxford, OH 45056, USA; smithdl2@miamioh.edu

* Correspondence: cmalaya@parker.edu

Abstract: Background: Gravity profoundly influences human locomotion. Studies examining the effects of hyper-gravity on gait have largely relied on added external mass, potentially confounding results with changes in inertia and center of mass. This study attempted to isolate the effects of increased gravitational load on kinematics and electromyography during walking at several different levels of load. **Methods:** Fifteen healthy adults were exposed to simulated gravitational loads ranging from 100% to 130% of body weight using a novel harness and spring-based system that increased weight without the addition of external mass and without altering limb inertia. Participants walked on a treadmill at a self-selected speed through incremental loading and unloading. Lower limb kinematics and electromyography data were recorded. Traditional measures of gait, as well as more dynamical measures, including angle-angle analysis and phase portraits, were examined. **Results:** Data demonstrated that a 130% load is sufficient to induce kinematic changes at the hip and knee; however, these changes become significant only during the transition from 130% to lower load levels. Ankle kinematics and electromyography appeared to be unaffected. **Conclusions:** These findings suggest that the presence of external mass and alterations in limb inertias should be considered seriously as independent variables in future loading studies, and that weight and mass may need to be considered as separate effectors during locomotion. This study also found that the act of loading and unloading elicit distinct responses in the joints of the lower extremities, as well as that it may induce an adaptative after-effect.

Keywords: gait; gravity; loading; hysteresis; kinematics; EMG; treadmill



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1. Introduction

Gravity is a ubiquitous natural phenomenon that pervades every aspect of human experience on Earth. While the acceleration caused by gravity (g) varies from 9.763 to 9.833 m per second squared (m/s^2) depending on your terrestrial location, “standard g ” is often modeled—and assumed—at 9.81 m/s^2 [1]. Though humans experience this acceleration as their own weight—the force borne of their own mass accelerating—the influence of our gravitational environment extends far beyond perception and weightiness. Gravity guides the formation of human bone structure and density [2], influences the discharge rates and amplitudes of cortical and spinal neurons [3], and can alter cellular morphology and metabolism [4]. However, there is arguably no system, structure or behavior affected so demonstrably by gravity as that of human locomotion.

Locomotion studies utilizing environments with increased gravitational effects (hyper-gravity) are extremely limited. Previous work investigating the effects of increased weight on human locomotion have found changes in trunk angles, hip, and ankle range of motion and cadence [5], as well as equilibrium points, stability, efficiency, gait speed, and walk-to-run transitions [6–8]. However, the studies contained by these reviews utilized the addition of mass to participant's bodies in order to increase weight. While the goal of these studies was to examine the effects of externally carried loads, it highlights an important confound as these—and similar studies—relate to the effects of hyper-gravity. The addition of external mass changes an individual's center of mass location [9], induces a stabilizing forward lean [10,11], and adds impactful, unevenly distributed inertial differences to a body [12,13]. These factors, while related to mass and weight, may be unrelated or tangential to purely gravitational constraints. Therefore, the extent to which these changes are driven by increased weight versus increased inertia on the limbs and trunk remains unknown.

Similarly, it is unclear if the kinematic and electromyographic changes seen during loading are due entirely to the relative load on an individual, or if the previous level of load may work as a factor in these responses. In humans, hysteresis—the dependence of a system on its previous states—has been a dynamic factor exhibited at the cellular level [14–17], scaling all the way up to cortical networks [18,19], proprioception [20,21] and gait [22–26].

It is possible that the same level of load could elicit different kinematic and EMG changes, depending on whether an individual increased or decreased their relative weight to reach it. However, very few studies have sought to examine hysteresis in human gait through manipulation of the gravitational load. Without a clear understanding as to the specific and exact role of loading and unloading, as well as specific increases in load and mass in these adaptations, it is difficult, if not impossible, to optimize load carriage for real world conditions and outcomes. In this study, we sought to investigate the effects of increased gravitational loading on kinematic and electromyographic variables without the addition of external mass. Specifically, we were interested in examining the kinematics and neuromuscular activity of individuals in simulated gravitational environments ranging from 100%, up to 130% of gravitational load. In this study, these questions were approached through the use of zero-dimensional (traditional kinematic and electromyographic measures) and one-dimensional (angle-angle diagrams and phase portraits) analyses. Previous use of these methods in this lab has found that they provide complementary information not otherwise apparent given the use of a single set of measures [27,28].

2. Methods

2.1. Participants

This study examined 15 healthy adults (25.3 ± 4.7 years; 67.1 ± 4.0 inches; 172.3 ± 42.0 lbs.; 53% female). Participants were not knowingly pregnant and did not have a history of, or any current systemic, degenerative or neuromusculoskeletal injuries or disease that could affect their ability to walk with differential loading for 15 min. This study was conducted in accordance with the Declaration of Helsinki and approved by the Institutional Review Board at the University of Houston (IRB#:00002971). Informed consent was obtained from all participants prior to enrollment in the study.

2.2. Kinematic Sensors

Participants were fitted with seven inertial measurement units (IMUs; XSens Awinda—Movella Inc., Henderson, NV, USA) arranged in a lower-body configuration. These sensors were placed bilaterally over the insteps of both feet, as well as anteriorly over the tibia at mid-shank and laterally over the mid-thigh. The final sensor was placed over the sacrum,

centered at the S2 tubercle. All XSens sensors were secured by proprietary neoprene straps with non-slip, rubber backings.

2.3. Electromyographic Sensors

Four dry surface electromyographic (EMG) sensors (Model SX230—Biometrics Ltd., Newport, RI, USA) were adhered—using hypo-allergenic, double-sided tape—over the right rectus femoris, biceps femoris, medial gastrocnemius and tibialis anterior. These sensors were placed over the belly of each respective muscle—conduction surfaces in line with the muscle fibers—after any body hair in that location was shaved, and the area was cleaned and scrubbed with an alcohol wipe. Sensor placements were performed in accordance with recommendations by the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) group. The control system (DataLOG MWX8—Biometrics Ltd., Newport, RI, USA) was mounted on each participant's low back using a stretchable, Velcro band around their waist.

2.4. Loading System and Walking Protocol

Participants were asked to wear a climbing harness with front and back D-ring attachments over their clothing. This harness allowed the participants to be attached to the loading system at two points of equal height, thereby creating an equivalent angle in the front and back ropes that would tether them to the system. This had the intended effect of canceling out any anterior or posterior forces from the system, leaving only a vertical component of load. Similarly, the loading system was connected to the harness by two springs, which allowed individuals in a small degree of normal displacement that would not be available if they were only connected to a taut rope. The entire harness, after being connected to the rope system, added 4 lbs. of weight distributed over the shoulders of the participant (see Figure 1). The entire loading system was built around a treadmill with embedded force plates (Bertec Corporation, Columbus, OH, USA); as such, after donning the harness and being connected to the loading system, an individual's weight could be calculated and loading parameters for 100%, 110%, 120%, and 130% of normal body weight were established.

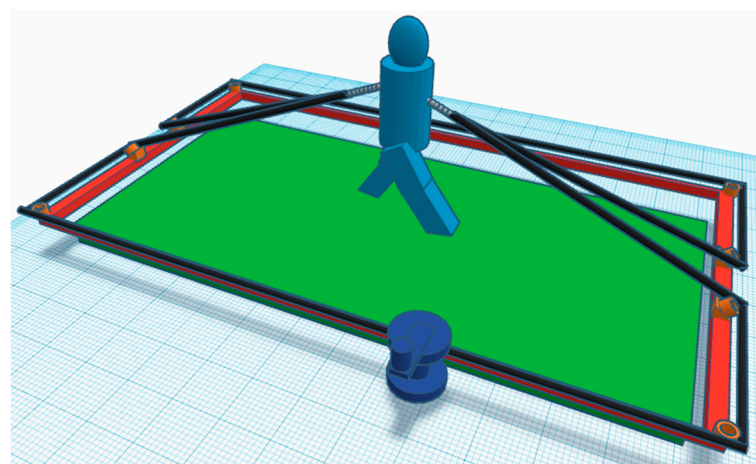


Figure 1. Loading system. In this figure, the participant (light blue humanoid) walks over an instrumented treadmill (green). They are connected to the rope system (black) by two springs (silver) that attach to a harness (not pictured). The ropes distribute vertical tension by way of 8 pulleys (orange) arranged around a metal frame (red). The tension in the rope system can be modulated by way of a crank pulley (dark blue) and vertical load is calculated by kinetic sensors embedded in the treadmill.

Participants were asked to walk at 100% of their normal load for 5 min at a self-selected, comfortable speed (mean speed: 0.78 ± 0.11 m/s). This gave each participant time to become familiar with the loading system, as well as for their gait to stabilize. Following the acclimation period, participants were loaded in 10% increments up to 130% of their body weight, spending 1 full minute at each level. After completing the full minute at 130% of body weight, the protocol was reversed, with participants walking for one minute at 120%, 110%, and 100% of normal load. For all levels of loading and unloading, the treadmill speed remained at the same speed each participant had selected earlier. There was no rest given in between each level besides the time it took to adjust the system to the desired load (≈ 10 s). Kinematic and EMG data were recorded for the final minute of the acclimatization period, as well as the full minute of walking at all levels of loading and unloading.

2.5. Data Processing

Kinematic data were streamed wirelessly from the XSens IMUs at 60 Hz to a computer running a data collection software suite (MVN Awinda ver. 2022.1). This software collected and internally calculated joint angles for the hip, knee, and ankle, bilaterally. Joint angle waveforms were separated into strides and normalized to 100 points using the peak knee as a reference. Mean, maximum, and minimum angles, as well as range of motion (ROM) were extracted for all joints. Data were exported, separated into strides using peak knee as the reference point, and statistically analyzed in MATLAB (R2019b: 9.7.0.1296695) using custom scripting.

Four channels of EMG data were simultaneously recorded by the waist-mounted control unit, as well as streamed to a computer running a data collection software suite (DataLOG ver: 10.27—Biometrics Ltd., Newport, RI, USA). The 12 g DataLOG control unit was set to sample at 1000 Hz, and provided $1000\times$ amplification gain as well as an automatic anti-aliasing filter prior to streaming. Data collected were exported into MATLAB for processing. Each channel was individually bandpass filtered (20 to 450 Hz) using a 2nd order Butterworth filter. Waveforms were then full wave rectified and enveloped using a low pass filter with an additional 2nd order Butterworth filter utilizing a cutoff frequency of 40 Hz [29]. EMG data were separated into strides and normalized to 100 points using the kinematic peak knee timestamps as a reference. After processing, peak values, root-mean-square (RMS) and integrated areas were calculated for all muscles. We calculated RMS as the square root of the mean of all values squared for each trial. This provides a metric representing the amplitude of the EMG signal [30]. We also calculated integrated areas for each trial to appraise the total electrical signal or drive from the central nervous system to the motorneuron [31–34].

This study made use of both zero- and one-dimensional analyses, representing traditional kinematic and electromyographic measures as well as phase portraits and angle-angle diagrams. These were created to examine the state spaces of and coordination between the joints of the lower extremity, respectively. Areas were calculated from mean phase portraits using a custom MATLAB script in order to quantify and compare the range of available behaviors.

2.6. Statistical Analysis

Kinematic and electromyographic variables were tested for normality and sphericity using the Kolmogorov–Smirnov and Shapiro–Wilk tests, as well as Mauchly’s test, respectively. Mean, maximum, minimum angles, and range of motion for each joint, as well as peak value, RMS, and integrated areas were compared across all levels of loading using repeated measure ANOVAs. Post hoc testing was performed with paired *t*-tests, as appropriate.

3. Results

Kolmogorov–Smirnov and Shapiro–Wilk tests revealed all data were normally distributed and Mauchly’s test showed sphericity was preserved.

3.1. Kinematics

Hip, knee, and ankle average joint angle waveforms by loading level are presented in Figure 2. Results from repeated measures ANOVA showed loading level had a statistically significant effect on hip mean ($F(6,84) = 2.447, p = 0.0314$) and max ($F(6,84) = 3.073, p = 0.0091$) values, as well as knee mean ($F(6,84) = 3.172, p = 0.0074$), and min ($F(6,84) = 4.647, p = 0.0004$) values. Pairwise comparisons for the hip and knee are depicted in Table 1. No differences in the ankle variables or any ROM values were found to be significant.

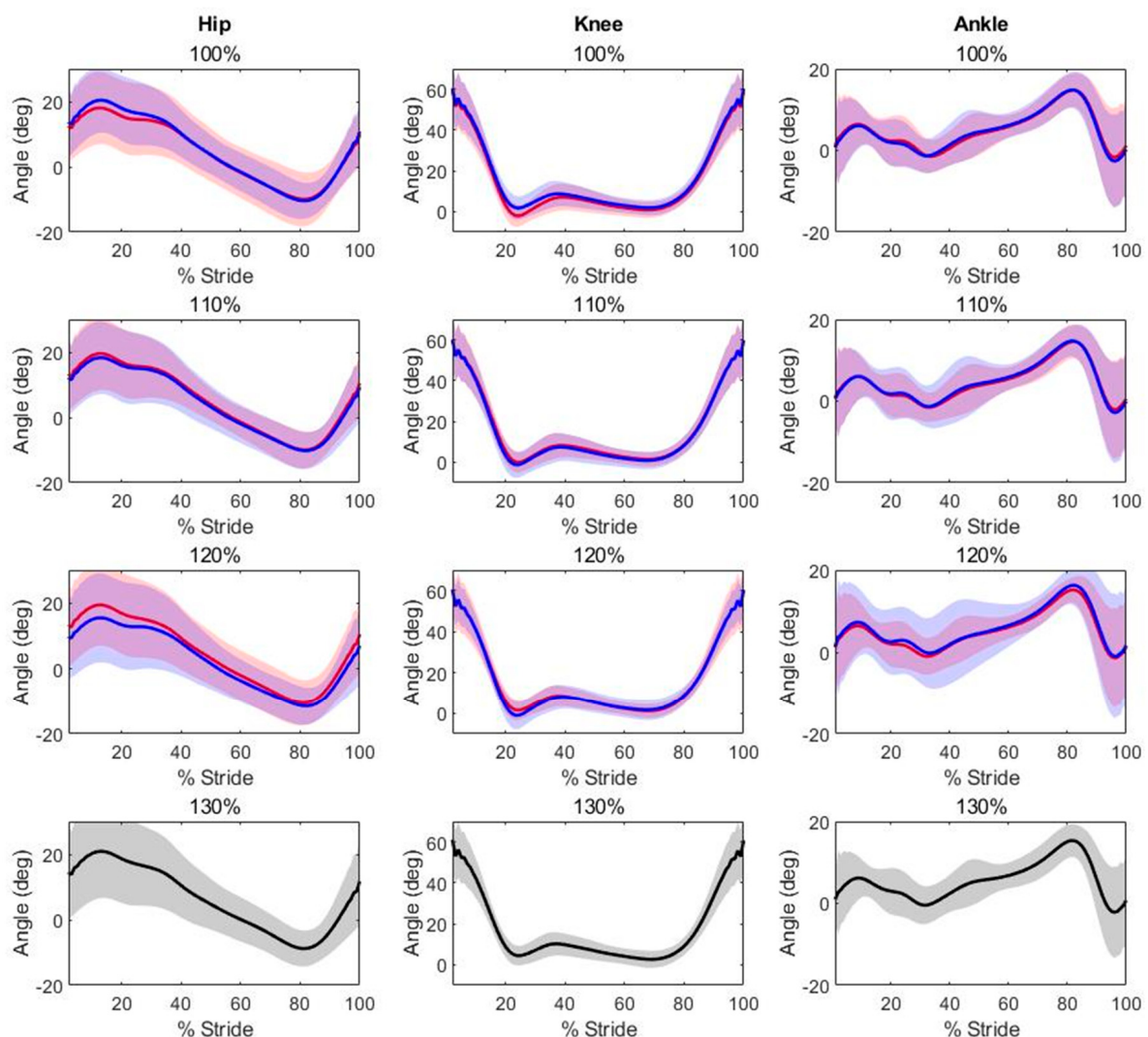


Figure 2. Hip, knee, and ankle joint angles by loading level. Each plot contains the kinematic waveforms for its respective loading (in red) and unloading (in blue) condition, along with a 2-standard deviation shaded area around each waveform. All 130% load conditions are in black to avoid any confusion, as only a single waveform is present.

Table 1. Hip and knee joint angles—pairwise comparisons.

		Condition	$\mu^\circ \pm std$	<i>p</i> Value
Hip	Mean	100	4.6 $\pm$ 7.6	0.1241
		110	5.2 $\pm$ 5.9	0.4919
		120	4.8 $\pm$ 8.6	0.0647
		130	6.4 $\pm$ 8.5	
		120 U	2.6 $\pm$ 8.3	0.0138 *
		110 U	4.5 $\pm$ 7.0	0.0642
		100 U	5.3 $\pm$ 6.5	0.3423
	Max	100	19.3 $\pm$ 10.9	0.0463 *
		110	20.5 $\pm$ 10.2	0.4931
		120	20.3 $\pm$ 12.1	0.1595
		130	22.0 $\pm$ 13.5	
		120 U	16.7 $\pm$ 12.9	0.0311 *
		110 U	19.2 $\pm$ 11.0	0.0488 *
		100 U	21.2 $\pm$ 9.8	0.6390
Knee	Mean	100	15.1 $\pm$ 5.1	0.0001 *
		110	16.2 $\pm$ 4.9	0.0068 *
		120	16.4 $\pm$ 4.6	0.0113 *
		130	17.8 $\pm$ 4.9	
		120 U	16.1 $\pm$ 3.8	0.1316
		110 U	15.4 $\pm$ 5.1	0.0082 *
		100 U	16.7 $\pm$ 4.7	0.1020
	Min	100	−3.1 $\pm$ 5.3	0.0012 *
		110	−1.7 $\pm$ 4.2	0.0055 *
		120	−1.1 $\pm$ 3.7	0.0027 *
		130	1.0 $\pm$ 3.9	
		120 U	−2.2 $\pm$ 6.3	0.0554
		110 U	−2.6 $\pm$ 5.6	0.0081 *
		100 U	−0.6 $\pm$ 4.8	0.0782

* denotes significance ($p < 0.05$). All pair-wise comparisons depicted represent the specified measure at that level of load versus 130% load. “U” denotes the specified level of load as it was unloaded to.

3.2. Electromyography

There were no significant differences in levels of load for peak muscle activity, root-mean-square, or integrated areas for any muscle. Mean and standard deviation EMG values by muscle, variable, and condition can be found in the Supplementary Materials.

4. Discussion

This study examined the effects of simulated gravitational loading (in this case, increased weight without the addition of extra, external mass) between 100% and 130% of body weight on kinematics and electromyographic variables during walking. We were interested in investigating if 130% of body weight was a sufficient load to induce kinematic and EMG changes, as well as examining the individual effects of increased weight without the addition of external mass. Our data revealed that 130% load is sufficient to elicit kinematic changes; however, these changes only appear significant when unloading from 130% to lesser loads. This suggests that walking at 130% and then unloading leads to gait alterations, while simply loading up to 130% does not. It is thus potentially the act of loading or unloading that can elicit changes at these levels, in addition to the actual borne weight.

Preceding studies have demonstrated that human proprioception diminishes in hypogravitational environments [35–37]. Indeed, anticipatory postural adjustments disappear below normal gravitational conditions [38] and kinesthetic responses to vibration dimin-

ish [39], with these changes being displayed not only kinematically, but also in the human cortical waveforms [40,41]. These studies indicate that alterations in human proprioception due to hypogravity are far reaching, and prevalent. This study found that increased loading at 130% of body weight was sufficient to elicit kinematic changes, but these changes were only clear as participants were unloaded to 120% of their body weight. To be clear, all levels of load from 100% increasing to 130% were not statistically distinct, yet 130% was significantly different from 120% in the knee and hip as participants were unloaded. This suggests that the level of load may not be the only operative factor in our findings; rather, the acts of loading or unloading may elicit distinct kinematic responses. Studies examining the drivers of hysteresis found that hysteretic effects were highest in the situations in which sensory information was the weakest [42] and that perceptual judgements are affected by the lack of or availability of information about an impending action [22].

In the case of this study, movement from a higher level of load (130%) to a lower level of load (120%) would reduce the relative amount of available sensory information. This, in turn, would invoke hysteretic changes in which participants based their expectation of movement in the new environment less on actual environmental cues, and instead more on internal models and expectation.

This concept is supported by Kostyukov and Cherkassky [43], which found muscle spindle discharge rates were higher after stimulation rate increases, and lower after decreases. It is also possible that some of these effects are modified by plantar pressure. Work by Kozlovskaya et al. [44] found that the removal of plantar support led to reflexive decreases in muscle activity and the eventual atony of extensor musculature with concurrent reductions in proprioception [45,46]. Exposure over longer time frames has led to decreased muscle strength-speed properties and motor control alterations [47,48]. Further, some of these alterations were entirely mitigated with plantar pressure stimulation [49]. As load increases, the relative increase in environmental-based proprioceptive information will drive gait behaviors more strongly; on the contrary, as the relative availability of proprioceptive information decreases, the reduction of sensory information will facilitate the use of information from previous levels of loading. This suggests that the effects of unloading, and loading are kinematically distinct, and should likely not be considered equivalent factors, even if used to reach the same level of load.

Phase portraits graphically represent all of the potential states of a dynamic system [50]. In this case, phase portraits depict all of the potential positions (i.e., angles) of a joint, as well as their velocities at that moment. A direct comparison of the portraits for 100% and 130% load (see Figure 3) shows that the movement structure of the joints is mostly preserved, with some stretching and translation as load increases. In combination with the angle-angle diagrams (see Figure 4), we can also see that the coordination between the joints is relatively similar, but also expanded and translated. This suggests that walking-type gait is relatively robust from 100% to 130% of body weight. Interestingly, this mirrors previous investigations of unloading down from full body weight and strengthens the idea that gait is a behavior centered around and suited to our particular gravitational environment. While this study did not present enough load to examine if an entirely new locomotive behavior would emerge at very high percentages of body weight (analogous and opposite to the sub-volitional shift into bounding-type gait found on the moon, for example), the durability of walking-type locomotion appears to be strong up to a 130% load.

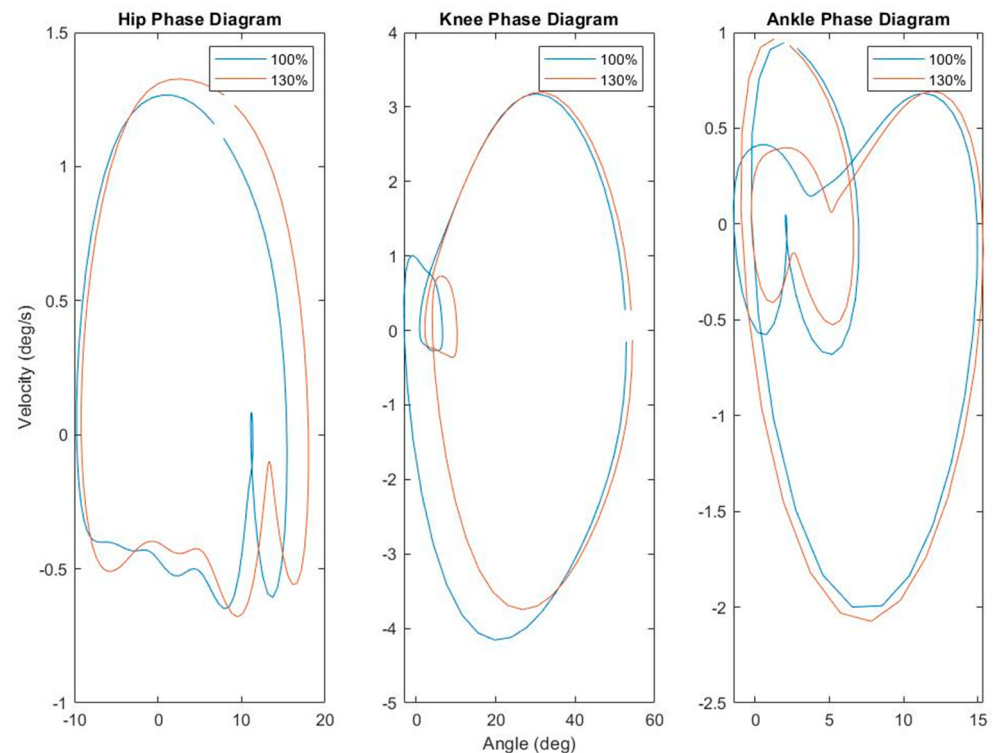


Figure 3. Phase portrait comparisons of 100% and 130% load. This graph displays phase portraits for the hip, knee, and ankle. Though there is some expansion of the range of available behaviors for the hip and knee—suggesting they are most sensitive to loading—the ankle appears to be mostly unaffected by the increase in load.

We also calculated the areas encircled by our phase portraits for every level of load by joint (see Figure 5). These areas are the two-dimensional spaces created by the outermost set of lines on each graph. These values provide a quantification of the state space of each joint at each level of load; in that way, they can be considered a means to numerically compare the contraction or expansion of the state space between different conditions. In this study, examination of the phase portrait areas of the knee reveals steady contraction of the state space as we increased to 130% of load, before a more than 10% expansion at 120% unloading. Interestingly, this expansion then contracts as we continue to unload, eventually settling at a smaller area than even the original (100%) load condition. The hip areas, on the contrary, consistently expand as we increase to a 120% load—drop slightly at 130%—before contracting significantly as we unload back to a 120% load. Analogously to the knee, the area of the hip phase portrait then continues increasing as load decreases, eventually ending at a larger area than even the original (100%) load condition. This suggests a crucially interesting relationship between the hip and the knee: as the range of available configurations of the hip expands, the knee, inversely, contracts. Similarly, as the hip contracts, the knee expands, and vice versa.

A consideration of the total areas across all three joints (see Table 2) finds a similar trend to the above. As loading increases to 130%, the overall summed areas of the three joints contract slightly before an almost 10% increase with unloading from 130% to 120%. This area remains relatively stable with unloading to 110% and then drops 4% with a return to 100% load. This, and the previous trending (in the hip and knee) highlight two primary ideas. First, this supports the previous assertion that loading and unloading do not appear to be equivalent phenomena. Second, while it is possible that there is an inflection point at 110% with unloading, it is also possible—given the similarity of 120% and 110% when unloading—that this is extinction of a loading induced after-effect. This has major

implications for populations in which load-carriage is common in that effective increases or decreases in weight can alter kinematics and movement structure, possibly even for time beyond the actual adding or subtracting of weight. Correspondingly, whether the individual was loaded or unloaded to a certain weight appears to induce specific changes that are not equal across similar loading levels.

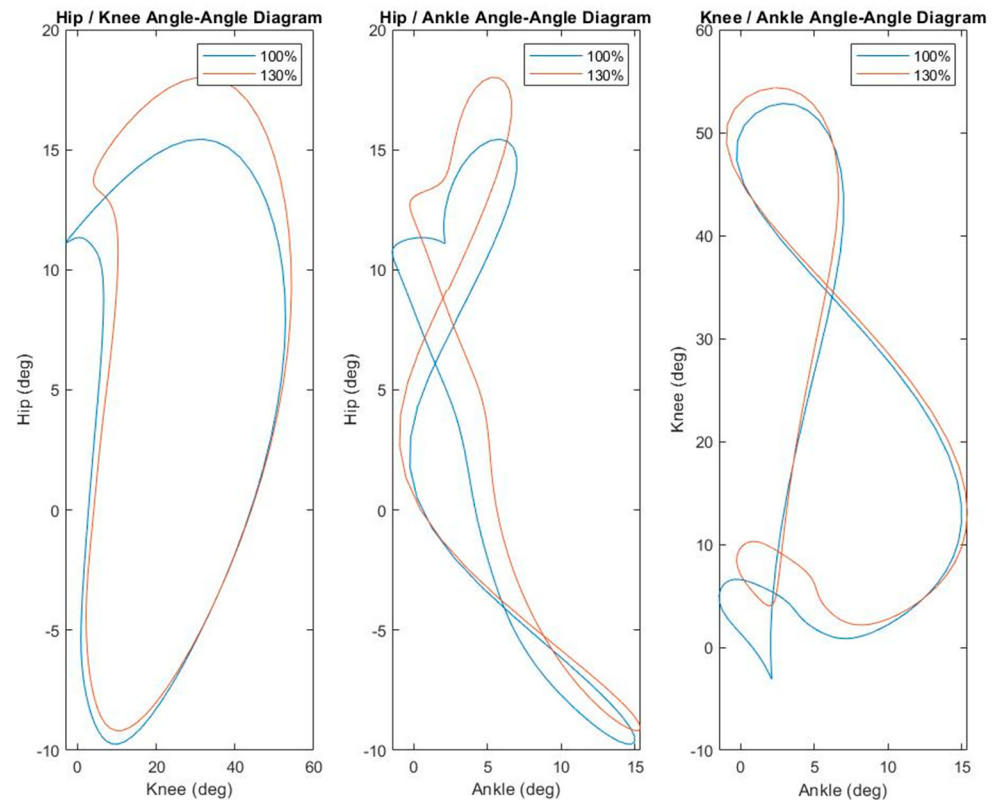


Figure 4. Angle-angle diagram comparisons of 100% and 130% load. This graph shows that the coordination strategies between the joints of the lower extremities are mostly preserved as load was increased. It is important to note, however, that there was distinct stretching and skewing of the shapes for all three graphs. This suggests that, although the general coordinative structure of movement between these joints was similar, they were not unaffected by load. Indeed, even 130% of body weight was enough to shift some aspects of the coordinative structure of the hip, knee, and ankle.

Interestingly, the kinematic changes seen in this study were not reflected in EMG data, in which no changes were found across any levels of load. This suggests that muscle activity and kinematic variables can decouple and respond to changes in load differentially. While previous investigations have found that kinematics can be accurately predicted from EMG data alone [51], others have found that kinematic and EMG variables correlate differentially depending on the activity being performed [52]. The results of this study suggest that loading and unloading are activities in which these variables do not track well with each other. It is important to note that this is also potentially due the absence of external mass. In this study, changes in limb inertia and center of gravity were bypassed through the use of our novel loading system. In this way, and in relation to load added over body weight, kinematic variables appear more sensitive than EMG to loading and unloading, and perhaps—given the results of other studies with positive EMG findings—EMG is more sensitive to changes stemming from changes in external mass/inertia. Alternatively, as we only recorded EMG data from four muscles, it is possible that other muscles exhibited changes in response to the loading protocol but were not captured.

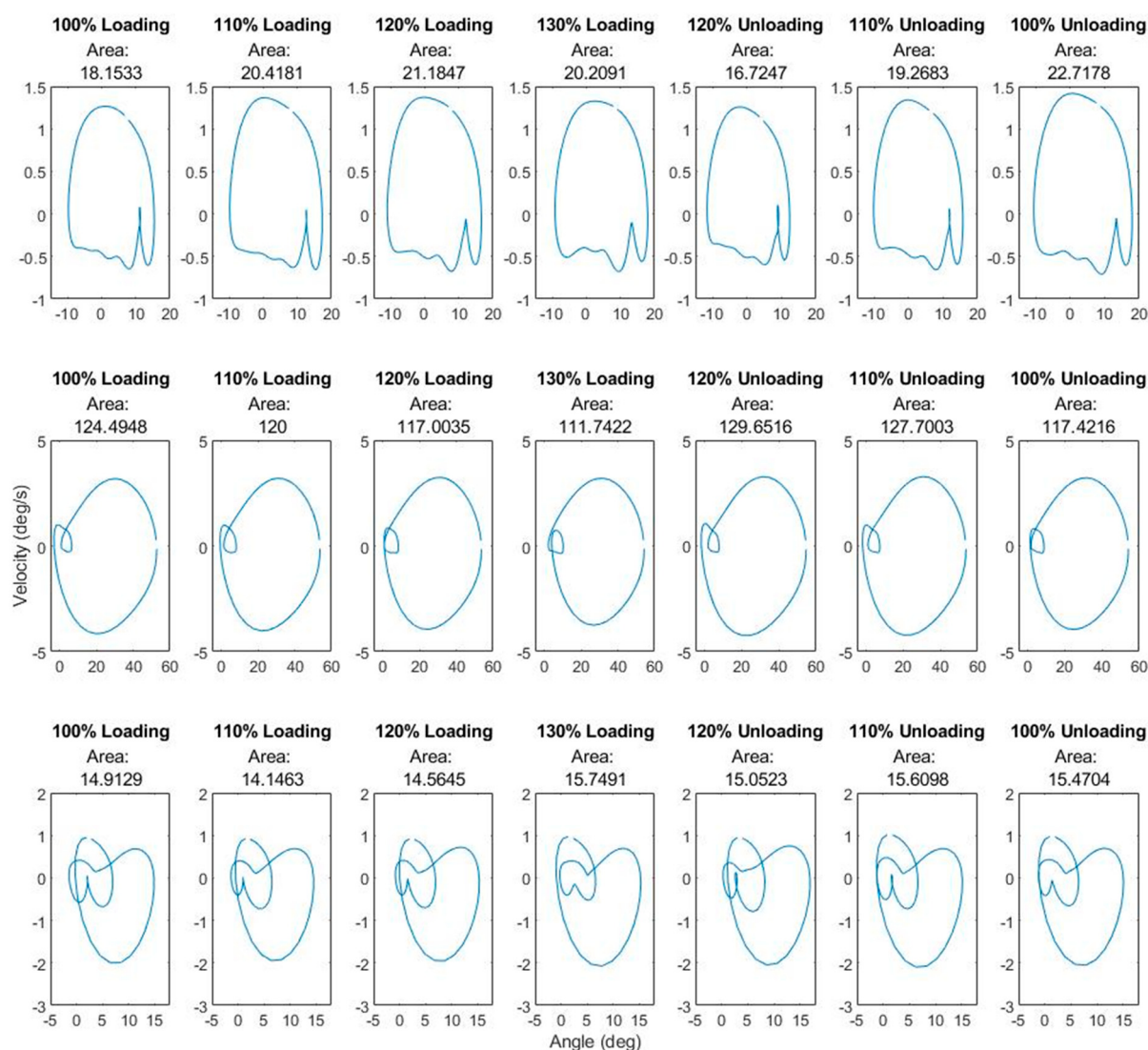


Figure 5. Phase portraits with calculated areas. The top row represents the hip phase portraits, the middle row is the knee phase portraits, and the bottom row is the ankle phase portraits. The area value represents the two-dimensional area—in pixels—occupied by each shape. These values can be considered a quantitative estimation of the range of available behaviors across each loading condition.

Table 2. Summed areas of phase portraits and percentage change.

Condition	% Load	Summed Areas of Hip, Knee and Ankle Phase Portraits	% Change from Previous Level of Loading
Loading	100%	157.561	
	110%	154.564	−2%
	120%	152.753	−1.1%
	130%	147.700	−3.3%
Unloading	120%	161.429	+9.2%
	110%	162.578	+0.7%
	100%	155.610	−4.2%

The results of this study have important implications for our understanding of the role that gravity plays in human locomotion. This study found that an increase in load can specifically affect both the knee and hip joints, as well as supporting the concept that loading and unloading are independent activities with specific responses. Even the same level of load, reached from higher or lower levels of weight, can elicit different responses.

In that way, increases in load appear to drive fewer hysteric changes than decreases (due to the relative availability of proprioceptive sensory information); as such, researchers should take care to ensure that their participants are responding to the correct level of load and should increase their load to the desired level, rather than unweight them.

There are also implications for using this system in long-duration spaceflight, where gravity may not be available to facilitate loading and gait. Indeed, the use of this system in a spacecraft could allow astronauts to maintain healthy levels of load for bone health and venous pumping, despite the absence of gravity, though research would clearly be needed to investigate this.

This study is not without limitations. There is a potential for this system to have influenced gait in some way, though no participant stated they felt the system interfered with their gait, arm swing, or ability to walk on the treadmill at any time. Similarly, participants were queried at all levels of load, and none expressed discomfort or fatigue with the system or any level of load. This study also did not compare its findings to more traditional loading studies examining the effects of external load on gait. As such, it is currently unclear how the effects seen in this study might compare with a heavy backpack or weight vest, for instance. However, while such a comparison was outside the scope of this study, future work should undoubtedly examine this. As the levels of load were not randomized, it is also possible that data could have been influenced by an order effect. This is a constraint of the system itself (individuals need to pass through any lower level of load in order to reach a specific level) which other potential users should be aware of.

This study was a novel investigation of an easily reproducible loading system that can increase the load of an individual, without the addition of external mass. Increased gravitational loading up to 130% of normal body weight can alter hip and knee kinematics but does not appear to affect the ankle joint nor does it appear to elicit changes in electromyographic variables. These findings suggest that the presence of external mass and alterations in limb inertias should be considered seriously as independent variables in future loading studies, and that weight and mass may need to be considered as separate effectors during locomotion. This study also found that the act of loading and unloading elicit distinct responses in the joints of the lower extremities, as well as that it may induce an adaptative after-effect.

Supplementary Materials: The following supporting information can be downloaded at: <https://www.mdpi.com/article/10.3390/biomechanics5020031/s1>, Table S1: EMG Peak by Muscle and Condition; Table S2: EMG RMS by Muscle and Condition; Table S3: EMG AUC by Muscle and Condition.

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Informed Consent Statement: Informed consent was obtained from all participants involved in the study.

Data Availability Statement: Data can be made available upon request.

Conflicts of Interest: The authors declare no conflicts of interest.

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