THE EFFECT OF INCREASING MASS UPON LOCOMOTION

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ABSTRACT

The purpose of this investigation was to determine if increasing body mass while maintaining bodyweight would affect ground reaction forces and joint kinetics during walking and running. It was hypothesized that performing gait with increased mass while maintaining body weight would result in greater ground reaction forces, and would affect the net joint torques and work at the ankle, knee and hip when compared to gait with normal mass and bodyweight. Vertical ground reaction force was measured for ten subjects (5M/5F) during walking (1.34 m s^{-1}) and running (3.13 m s^{-1}) on a treadmill. Subjects completed one minute of locomotion at normal mass and bodyweight and at four added mass (AM) conditions (10%, 20%, 30% and 40% of body mass) in random order. Three-dimensional joint position data were collected via videography. Walking and running were analyzed separately. The addition of mass resulted in several effects. Peak impact forces and loading rates increased during walking, but decreased during running. Peak propulsive forces decreased during walking and did not change during running. Stride time increased and hip extensor angular impulse and positive work increased as mass was added for both styles of locomotion. Work increased at a greater rate during running than walking. The adaptations to additional mass that occur during walking are different than during running. Increasing mass during exercise in microgravity may be beneficial to increasing ground reaction forces during walking and strengthening hip musculature during both walking and running. Future study in true microgravity is required to determine if the adaptations found would be similar in a weightless environment.

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INTRODUCTION

Specific Aims

Astronauts perform locomotive exercise on the International Space Station (ISS) as a countermeasure to the detrimental physiological losses associated with space flight. The locomotive-related ground reaction forces may provide mechanical loading that is a sufficient stimulus for bone adaptation (Turner, 1998), and thus reduce the bone loss that occurs. However, recent investigations have shown that the ground reaction forces developed during locomotion in microgravity are less than those occurring in normal gravity (DeWitt, Schaffner, Blazine, Bentley, Laughlin, Loehr & Hagan, 2003; DeWitt, Schaffner, Laughlin, Loehr & Hagan, 2004, Schaffner, DeWitt, Bentley, Yarmanova, Kozlovskaya & Hagan , 2005). Specifically, peak ground reaction forces measured in microgravity were less than those developed in normal gravity even when subjects were loaded at external load levels equal to or greater than their body weight. The decrement in peak force may have been due to inadequate loading, altered distribution of the applied external load, or fundamental differences in gait between microgravity and normal gravity.

The ground reaction force occurring between the foot and ground is equivalent to the resultant force applied to the body's center of mass. Since force is the product of mass and acceleration, an increase in body mass with a given acceleration should result in an increase in ground reaction force. The loading mechanisms used on ISS to date appear to be limited in allowing astronauts to achieve normal-gravity-like levels of peak ground reaction force. Thus, other approaches may be necessary to effectively create normal-gravity-like ground reaction force using the current treadmill onboard ISS. Adding mass to the astronaut could result in an increased ground reaction force. The addition of mass to the subject, however, may also affect the kinematics, kinetics and adaptations in normal locomotion.

Increasing the mass of the astronaut through a weighted vest offers a potentially economical and easily modifiable enhancement to current exercise countermeasures. The use of a harness capable of providing different levels of mass could save resources by reducing the need to develop new exercise devices. More importantly, the health and well-being of astronauts could be improved efficiently and inexpensively.

The goal of this investigation was to determine whether or not increasing subject mass can create more effective exercise countermeasures during long-term spaceflight and to better understand the motor accommodations made to gait. The primary purpose of this investigation was to determine how the manipulation of mass affects the ground reaction force during treadmill locomotion. The secondary purpose was to examine the kinetic adaptations to increased body mass during locomotion.

BACKGROUND AND SIGNIFICANCE

During long-term space flight, astronauts experience losses of bone mineral density and losses of muscular strength and mass (Iwamoto, Takeda and Sato, 2005). Although the specific amount of bone mineral density loss varies among astronauts, the location of greatest loss tends to be in the critical weight bearing areas of the proximal femur, tibial plateau and calcaneus (Schneider, Oganov, LeBlanc, Rakmanov, Taggart, Bakulin, Huntoon, Grigoriev & Varonin, 1995; LeBlanc et al., 2000). As a countermeasure to these losses, astronauts perform locomotive exercise to create a bone-remodeling stimulus. The ground reaction force associated with locomotive exercise is hypothesized to stimulate bone remodeling (Davis, Cavanagh, Sommer & Wu, 1996).

When performing locomotive exercise, the astronauts wear a harness attached to a vertical tether that pulls them back to the treadmill (see Figure 1). Impact and propulsive forces are applied to the axial skeleton via the ground reaction forces that occur during foot and treadmill surface contact. The amount of external load used during each exercise session is variable and dependent upon the loading mechanism. Bungees are used as one sort of loading device. The external load delivered by bungees is dependent upon length and is varied by inserting one or more caribiner clips between the bungee and the attachment point of the treadmill.

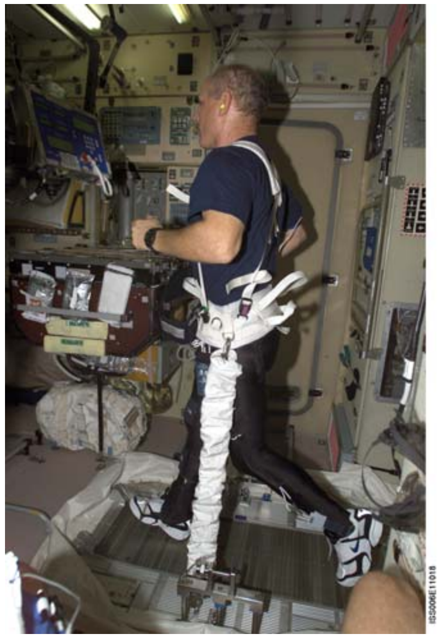

Figure 1. Astronaut exercising on a treadmill onboard the International Space Station. The astronaut is wearing a harness around his waist and shoulders and is connected to the treadmill via elastic bungee cords (vertical attachment between the harness and treadmill).

A mechanical subject-loading device can also be used to deliver external load. Loading is provided with a servo-motor system that is adjustable via a control panel. During current and past missions, the astronauts commonly use bungees to load themselves to about 50% to 60% of their body weight during a majority of their mission. However, despite the locomotive exercise, astronauts still experience loss of bone.

McCrory, Baron, Balkin and Cavanagh (2002) encourage maximizing the external load to 100% of body weight to generate ground reaction forces similar to those seen in normal gravity. Their conclusion was based on data collected during locomotion on a vertically-mounted treadmill used to simulate microgravity. However, in actual microgravity, increasing external loads may not be sufficient to recreate ground reaction force magnitudes experienced in normal gravity. In microgravity during parabolic flight, peak ground reaction force magnitudes were less than those occurring in normal gravity, even when the subject is loaded at levels near or greater than their bodyweight (Schaffner et al., 2005; DeWitt et al., 2004) (Figure 2).

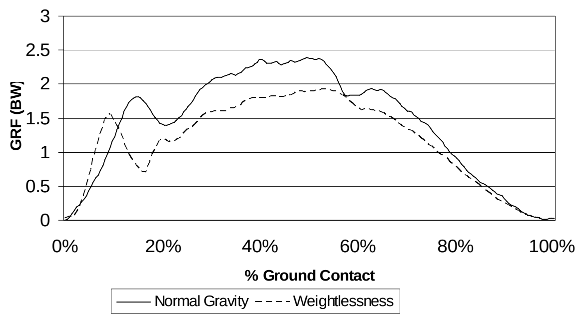

Figure 2. Typical GRF trajectories during running at 7 mph in normal gravity and microgravity. Weightless running was completed with the external tether load equal to one BW during quiet standing.

Since force is the product of mass and acceleration, increasing the mass of the subject could result in an increase in ground reaction force. While there have been no prior investigations of this hypothesis in microgravity, Grabowski, Farley and Kram (2005) and Chang, Huang, Hamerski and Kram, (2000) have examined the effects of varying mass and body weight during locomotion in normal gravity. These studies used an overhead unweighting apparatus that applied upward forces to the body at a variety of magnitudes through a harness. However, the findings of these studies are conflicting.

Grabowski et al. (2005) found that increasing body mass using a weighted vest without increasing body weight resulted in an increased metabolic cost during walking. This suggests that the increase in mass, which results in an increase in the work performed on the center of mass, incurs a significant metabolic cost. Thus, increasing mass without increasing body weight causes subjects to expend more energy.

Chang et al. (2000) examined the ground reaction forces, gait temporal variables, and the orientation of the ground reaction force vector during running at various weight and mass conditions. Their maximal increased mass condition was 130% of normal body mass. They reported that an increase in mass without a corresponding increase in body weight did not result in an increase in ground reaction force or changes in locomotion kinematics.

If increasing body mass without increasing body weight results in an increased metabolic cost during walking, but no change in ground reaction force magnitudes during running, there must be some other explanation for the increased energy expenditure. It is possible that the larger added mass conditions used by Grabowski et al. (2005) compared to Chang et al. (2000) resulted in the increased metabolic cost findings. It is also possible that walking and running are two distinctly different activities, and the adaptations by each to additional mass are different. Finally, there may be kinematic and/or muscular adaptations that occur to compensate for the increased mass. However, these internal adaptations result in little or no kinematic adaptations.

PURPOSE AND HYPOTHESES

The purpose of this investigation was to determine how increasing body mass affected the kinetics and kinematics of locomotion. Two issues were examined. The first was how increasing a person's mass while maintaining body weight affects the forces that are transferred to the body, and how these forces affect the resulting movement. The second was to examine the adaptations that occur to the addition of mass without increasing body weight. The results of this investigation will help to determine if the addition of mass during locomotive exercise can enhance current space flight exercise countermeasures. Two hypotheses were tested during this investigation.

Hypothesis 1: Performing gait with increased mass while maintaining body weight will result in greater ground reaction forces than gait performed with normal mass and body weight.

Hypothesis 2: Performing gait with increased mass will increase the net joint torques and work at the ankle, knee and hip when compared to gait with normal mass and bodyweight.

METHODS

HUMAN SUBJECTS

Ten subjects (five men and five women) participated in this study (see Table 1). The sample was drawn from the test subject pool at NASA Johnson Space Center in Houston, TX. The test subjects approximated the age range of the current astronaut population. In order to be eligible for the test subject pool, each subject had to be healthy and pass a United States Air Force Class III equivalent physical. In addition, because the weighted vest had a maximum capacity of 38.1 kg (84 lbs), all subjects had to weigh less than 200 pounds. Prior to being accepted to the sample, each potential subject was screened to ensure that they were healthy and free from injury.

Table 1. Subject demographics (mean ± SD).

	Height (cm)	Weight (kg)	Age (yrs)
M (n=5)	177.3 ± 5.8	77.2 ± 3.5	34.7 ± 4.5
F (n=5)	164.1 ± 7.9	59.6 ± 10.0	34.1 ± 9.3
Total (n=10)	170.7 ± 9.5	68.4 ± 11.7	34.4 ± 6.9

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 was free to withdraw from the study at any time. All trials were conducted in the Exercise Physiology Laboratory at NASA-Johnson Space Center. All subjects completed the testing protocol without problems.

EXPERIMENTAL SETUP

Vertical ground reaction force data were collected during the testing trials with a force-measuring treadmill (Kistler Gaitway, Amherst, NY) at 480 Hz. The treadmill was equipped with two force plates beneath the running tread arranged so one plate rested in the front and one in the rear of the locomotion area. Each plate contained four piezoelectric load cells that measured vertical ground reaction force and allowed for a determination of the center of pressure during each sample. Each force plate was calibrated prior to the study, and the location of the origin of the force plates was found.

Three-dimensional position data from reflective markers placed upon the subject were collected at 60 Hz with eight cameras. (Smart Elite motion capture system, BTS Bioengineering Spa, Milanese, IT). Prior to each day of data collection, the motion capture system was calibrated to within 0.44 ± 0.03 mm of marker reprediction accuracy. All three-dimensional data were expressed relative to an inertial reference frame that was established during calibration. A reference trial was collected after calibration but before the subject arrived at the lab to establish a treadmill reference frame. An electronic pulse was output by the force treadmill upon the initiation of data collection. The signal was recorded by the motion capture system and was used to synchronize the data during post processing.

Mass was added to each subject using a weighted exercise vest (X-Vest, Perform Better, Cranston, RI). The vest was worn over the shoulders and had top and bottom pockets on the fore and aft of the subject. Each pocket was fitted with slots where up to twenty-one individual 0.4545 kg masses could be placed. Slots for weight placement were located on the inside and outside inner surface of each pocket (10 on the outside; 11 on the inner side). During added mass (AM) trials, masses were added equally to the front and rear of the vest. The masses were always added to the inner-lower slots first, followed by inner-upper, outer-lower, and outer-upper slots. Within each pocket, weights were added to the center of each row of slots first, and then fanned outwards. This method was used to keep the additional mass close to the subject's natural center of mass and to ensure systematic application of all AM condition. Body weight was maintained with an overhead unweighting system (H/P/Cosmos Airwalk, Nussdorf, Germany). The system provided a constant upward force via a pneumatic pump. The subjects wore a harness about their waist and thighs that was provided by the unweighting system manufacturer (see Figure 3).

Figure 3. Typical data collection with the H/P Cosmos Airwalk Unweighting System and Weighted X-Vest.

Data were collected during five AM treatments at two speeds. Subjects walked at 1.34 m s^{-1} (3 mph) and ran at 3.13 m s^{-1} (7 mph). In addition to a control condition with no added mass (0% AM), 10%, 20%, 30% and 40% of additional mass was added while body weight was maintained. At each AM condition, subjects had their weight relieved with an unloading system so the net force between the subject and treadmill remained equal to 100% body weight during quiet standing.

All trials at each speed were completed during a single data collection session for each subject. Prior to actual data collection, each subject participated in a familiarization session during which they had the opportunity to practice walking and running at each speed and treatment condition.

Subjects completed all added mass treatments at one speed before completing treatments at the other speed. The speed order was randomized for each subject by a coin flip prior to their first testing session. Treatment randomization occurred independently for each speed. In order to assure that there was a balance of increased mass conditions between subjects, a balanced Latin square random assignment was used (Portney and Watkins, 2000). The design allowed for a balance of treatment orders so that no two testing sequences were the same for different subjects within each speed. Each subject was randomly assigned a sequence from the table with only one subject completing each specific order. Trial order assignment occurred separately for each speed. The subjects wore the unweighting harness during all conditions, including the 0% AM trial.

Preliminary Procedures

Upon arrival to the laboratory, each subject was provided with running shoes (Xccelerator TR, Nike, Inc, Beaverton, OR) and completed a general health questionnaire. After the subject changed into spandex running shorts, reflective markers were attached to the subjects' left side to approximate the lateral malleolus (ankle), lateral femoral condyle (knee), and greater trochanter (hip). Additional markers were placed on the lateral neck, level with the fifth cervical vertebrae (C5), the posterior heel on the rear of the running shoe, and on the tip of the shoe over the distal second metatarsal.

A static trial was recorded prior to any locomotion trials while the subject stood upright with each joint in the anatomical neutral position. The static trial was used to determine the zero positions for each joint angle. Once the unweighting harness had been donned and the markers attached, the subject was weighed on the force treadmill. The weight of the subject was noted and used to compute the appropriate AM magnitudes.

Data Acquisition

The following procedure was repeated prior to each trial. The treadmill load sensors were reset while the subject was not in contact with the treadmill belt to ensure that the output voltage for each sensor corresponded to a net force of zero Newtons. The subject then stood in the middle of the treadmill belt so that their feet were in contact with both plates. The front half of each foot was placed on the front plate and the back half of each foot was placed on the rear plate. The unweighting harness was worn, but no upward force was applied. The force treadmill data acquisition software provided an instantaneous readout of the force applied to the treadmill, which in this case was the subject's bodyweight. Masses were added to the weighted vest to the appropriate magnitude calculated during the preliminary procedures. The weighted vest was then placed on the subject with the help of two assistants, and the subject's weight, including that of the weighted vest, was noted. Additional masses were added or removed from the vest until the total weight of the subject approximated the target weight within 1 kg.

The unweighting system was then engaged to apply an upward force approximately equivalent to the extra weight added to the subject. The investigator operated a dial on the suspension system while monitoring the instantaneous weight measured by the treadmill. The upward force was adjusted until the weight measured by the treadmill was within 1 kg of the original body weight. The final weight was recorded, the suspension system was locked to provide a constant upward force, and data collection began.

Subjects completed approximately 1 minute of treadmill locomotion at each AM condition. Data collection began once the subject's gait appeared to achieve steady-state. Immediately following the 1 minute of data collection, the weighted vest was removed, the unweighting harness was released and the subject completed 3 minutes of walking. This exercise period was used to eliminate any adaptation to gait that may have occurred during the test condition. The subjects were then given additional rest until they felt that they were ready to continue with the next AM condition.

Data Analysis

The first ten strides of the left leg were analyzed in 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 written in Visual Basic for Applications interfaced with Microsoft Excel 2003 SP1 (Redmond, WA) and MATLAB Version 7.2.0.232 (R2006a) (Natick, MA) were used for the entire analysis. Initial processing was completed on the ground reaction force data and motion capture data separately. Once initial processing was completed, the data were combined into a single file for each trial with ground reaction force and motion capture data synchronized.

Ground Reaction Force Data Initial Processing

Custom software converted the output from each force sensor to net ground reaction force and center of pressure location. Raw voltage data from the eight load sensors in the treadmill force platforms were transformed into forces using calibration factors and equations provided by the treadmill manufacturer. The total vertical ground reaction force during each sample was then found as the sum of the forces measured by each sensor. Center of pressure during each sample was found relative to the force platform reference frame using the force outputs from each sensor along with the dimensions of the force sensors relative to one another. The treadmill manufacturer supplied the center of pressure determination equations. Different equations were used depending upon the foot being in contact with one or both force platforms.

Once center of pressure locations were found, the data were rotated into the laboratory reference frame by multiplying the lateral coordinates by minus one (–1). The position of the origin of the laboratory reference frame expressed in the force platform reference frame was then subtracted from each center of pressure sample to translate the data.

Motion Capture Data Initial Processing

Raw motion capture data were rotated and translated into the laboratory reference frame. The laboratory reference frame was oriented so the x-axis approximated the mediolateral axis, the y-axis approximated the vertical axis, and the z-axis approximated the longitudinal axis of the treadmill. After rotation, the data were translated so the origin corresponded with a marker placed on the rear right corner of the treadmill. This reference frame orientation allowed for all motion capture marker coordinates to be expressed in positive numbers.

After rotation and translation, the data were examined for any missing points due to marker dropout. All marker trajectories were examined and any gaps were filled via cubic spline interpolation. After gaps were filled, the data were filtered using a fourth-order recursive, zero-phase-shift Butterworth low-pass filter. In order to objectively determine the most appropriate filtering cutoff frequencies for each marker, 20 motion capture files (2 from each 5 AM conditions × 2 speeds), were chosen randomly across subjects and treadmill speeds for analysis.

A Fast Fourier Transform (FFT) analysis was used on each individual coordinate for each marker in each file to determine its frequency content. Prior to the FFT analysis, each marker dataset was examined to identify missing samples. If more than 20% of the samples were missing due to marker dropout, the FFT analysis was not completed on that marker for that trial. The resulting power spectrum was analyzed and the area beneath the curve was computed. The threshold frequency at which 90% of the original total area was contained was noted. The mean threshold frequency was computed for each marker (mean of x, y and z coordinates as examined individually) and recorded as the appropriate cutoff frequency for that marker.

The analysis revealed a variety of cutoff frequencies for the data set (2.34-11.14 Hz. Filtering the data at too low of a cutoff frequency could result in oversmoothing, where true data are removed along with excessive noise. In contrast, undersmoothing occurs when too high of a cutoff frequency is chosen and noise remains in the signal.

Since it would have been of greater issue to oversmooth the data than to undersmooth, it was decided before any additional processing to smooth all markers at the highest mean cutoff frequency for all of the markers at each speed. Therefore, the marker dataset was filtered with a cutoff frequency of 11.14 Hz.

Data Analysis

Foot contact was defined as the time between heel strike and toe off. Determination of foot-treadmill contact was important because it was necessary to discern whether ground reaction forces were caused by the left or right foot. It was also necessary to verify correct synchronization of the data between the devices. For that reason, the determination of heel strike and toe off was completed independently for the ground reaction force and motion capture data.

Determination of Foot-Treadmill Contact

Ground Reaction Force Data

Ground reaction force at the time of heel strike was found based on the criterion of Chang et al. (2000). An automated algorithm using ground reaction force data found heel strike as the sample at which a positive change in the force greater than 1 $N \cdot s^{-1}$ occurred when the force magnitude was less than 100 N.

Motion Capture Data

The motion capture samples at which heel strike occurred were found by inspection of the heel motion along the treadmill longitudinal axis. Acceleration and the derivative of the acceleration (jerk) of the heel marker were found for each sample using finite central differences. Heel acceleration along the longitudinal axis was examined for a local minimum as determined when jerk was equal to zero. The heel marker vertical position was examined for local minimums, which occur near the time of heel strike. The location of minimum heel longitudinal axis acceleration was then located between maximum and minimum heel vertical position. An 8 sample window near the time of minimum heel acceleration was then examined for the samples at which jerk turned from negative to positive. The last negative jerk sample number was noted as the sample of heel strike.

The actual time of heel strike (when jerk was equal to zero) may have occurred between motion capture samples. Therefore, a linear interpolation equation was used to estimate the time at which jerk was equal to zero (Eq. 1).

Time of heel strike = $t_1 + (J(t_1)/(J(t_1) - J(t_2)) * t_{int}$ (1)

In equation 1, t_1 was the time of the sample at which the last negative vertical value occurred, t_2 was the time of the sample at which the first positive jerk occurred, t_{int} was the time between samples (1/60 s), and $J(t_1)$ and $J(t_2)$ were the jerk values at $t_1$and t_2. The sample at which toe off occurred during each stride was found in a similar manner using equation 1 by determining the time of maximum acceleration of the toe marker along the vertical axis.

Synchonization

Prior to synchronization, the ground reaction force data were down-sampled to 60 Hz. A comparison of heel strike times found with the motion capture data with those found with the ground reaction force data revealed a slight, but inconsistent, offset of approximately 0.06 sec. However, this offset is based on the assumption that the true sample of heel strike was found using motion capture data. There is the potential that the exact time could have been estimated incorrectly because of compliance in the heel of the shoe that resulted in the heel marker continuing to travel forward (positive z direction) even though the foot was in contact with the ground.

After synchronization, the difference in time between heel strike identified using motion capture data and ground reaction force data increased as a function of data collection time. This occurred for all trials at a relatively constant rate. Although the ground reaction force workstation was set to collect data at 480 Hz, it is possible that the true sampling rate was slightly different. This seemed apparent because a repeated event that occurred at a single moment in time (e.g. heel strike) was found to occur at different time intervals depending on the type of data (motion capture vs. ground reaction force) being analyzed.

After comparing the time of heel strike computed with motion capture data to the corresponding times computed with ground reaction force data, the mean (±S.D.) ground reaction force sampling frequency was 481.2 ± 4.6 Hz. In order to determine if the recomputed sampling frequency was accurate, time of heel strikes as found using ground reaction force data were recomputed using 481.2 Hz. The mean differences between motion capture and ground reaction force heel strikes still exhibited an offset, but the systematic offset that increased as a function of trial time was eliminated. Therefore, for all calculations, the force platform true sampling frequency was taken as 481.2 Hz.

Ground Reaction Forces

Ground reaction force data were used to find contact time, stride time, peak impact force, loading rate, peak propulsive force and impulse for each trial. All analyses were completed using raw ground reaction forces to ensure that peak values were not dampened during smoothing. Visual inspection of each footfall was used to identify anomalous data.

Contact time was the length of time that the left foot was in contact with the treadmill 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 (see Figure 4).

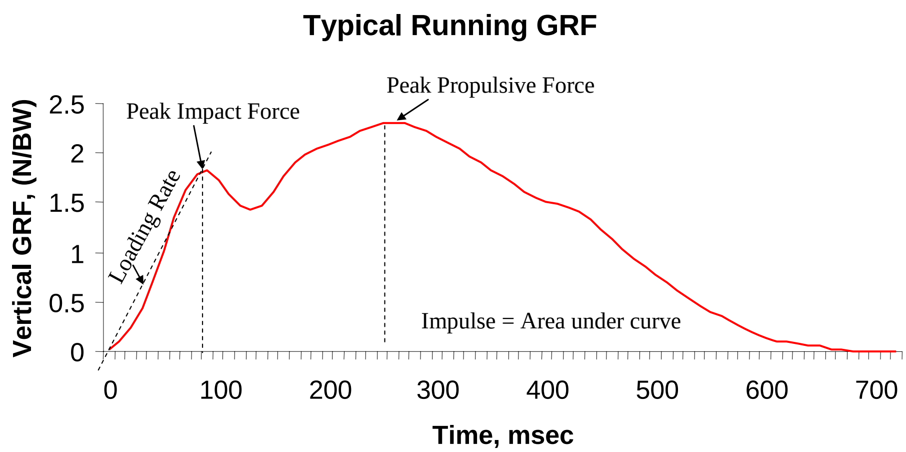

Figure 4. Typical running ground reaction force and dependent variables.

Joint Kinematics

Joint angle trajectories of the ankle, knee and hip were found for each stride of each trial. Hip angle was the angle separating the thigh and trunk vectors. Positive hip angles indicated flexion. The knee angle was the angle separating the thigh and shank vectors. Positive angles indicated extension. The ankle angle was the angle separating the foot and a vector perpendicular to the shank vector. Positive angles indicated plantar flexion. All joint angles were corrected using mean joint angles computed during the static trial.

Joint Torques

Synchronized data were used to compute the net torques at each joint. Segment inertial properties were found for each subject using body segment parameter tables (De Leva, 1996). Standard inverse dynamics using Newton-Euler equations of motion were used to compute net joint torques (Hof, 1992). Ground reaction forces were used as an input to the model when foot-treadmill belt contact was occurring. Positive torques represent hip and knee extension and ankle plantar flexion. Flexor and extensor angular impulse were computed as the area under the corresponding torque curves using the trapezoid rule. Separate analyses were completed during each stride for stance and swing phases.

Work and Power

Joint power trajectories were computed for each stride of each trial as the product of joint torque and joint angular velocity. Positive work was computed as the area under the positive joint power curve, and occurred when an extensor torque was present as the joint sped up while extending, or when a flexor torque was present when the joint sped up while flexing. Likewise, negative work was found as the area under the negative joint power curve. Separate calculations were made for stance and swing phases of locomotion.

Dependent Variables and Trial Means

There were eighteen dependent variables computed for each stride during each trial. However, for the joint torque and power dependent variables, separate analyses were conducted during the stance and swing phase of each stride. Therefore, there were thirty dependent variables for each trial. A trial mean was found over all strides for each variable. Tables 2 and 3 summarize the dependent variables used in this study.

Table 2. Summary and description of gait parameter and ground reaction force dependent variables.

Dependent Variable	Description
Gait Parameters	
Contact Time	Time between heel strike and toe off
Stride Time	Time between successive heel strikes
Ground Reaction Forces	
Peak Impact Force	First distinct peak in the ground reaction force trajectory
Loading Rate	Average slope of the ground reaction force trajectory from heel strike to peak impact force
Peak Propulsive Force	Second distinct peak in the ground reaction force trajectory
Impulse	Area under ground reaction force curve

Table 3. Summary and description of joint torque and power dependent variables.

Dependent Variable	Description
	Joint Torque and Power
Hip Flexor Angular Impulse	Area under hip flexor torque curve
Hip Extensor Angular Impulse	Area under hip extensor torque curve
Knee Flexor Angular Impulse	Area under knee flexor torque curve
Knee Extensor Angular Impulse	Area under knee extensor torque curve
Ankle Plantarflexor Angular Impulse	Area under ankle plantarflexor torque curve
Ankle Dorsiflexor Angular Impulse	Area under ankle dorsiflexor torque curve
Hip Positive Work	Area under hip positive power curve
Hip Negative Work	Area under hip negative power curve
Knee Positive Work	Area under knee positive power curve
Knee Negative Work	Area under knee negative power curve
Ankle Positive Work	Area under ankle positive power curve
Ankle Negative Work	Area under ankle negative power curve

STATISTICAL ANALYSIS

The mean of each dependent variable over ten strides was found for each trial. Statistical analyses were conducted utilizing NCSS 2004 statistical software (NCSS, Kaysville, Utah). Trial means were tested using a repeated measures analysis of variance (ANOVA) with AM level as a single factor. Walking and running were analyzed separately because they are two different tasks that require different kinematics. Tukey-Kramer Multiple Comparisons tests were used to determine differences between AM levels if a significant main effect was found. Statistical significance was achieved at an alpha level of 0.05.

RESULTS

GAIT PARAMETERS

The results of this investigation are presented in this section, including gait parameters, ground reaction forces, joint angular impulses, and the positive and negative work at each joint. Ensemble average plots of the ground reaction forces during walking and running are presented, along with angular velocity, joint torque, and joint power curves for the hip, knee and ankle during walking and running. The plots are for a single stride and are the mean trajectories across all subjects.

Contact Time

There was an AM effect upon contact time during running (see Table 4). The addition of any mass affected contact time, as indicated by the 0% AM being different from all other conditions. In addition, the 40% AM condition had a greater contact time than the 10% and 20% AM conditions, and the 30% AM condition had a greater contact time than the 10% AM condition.

Table 4. Contact time and stride time (Mean±SD) for each AM condition for walking and running.

Speed	0% AM	10% AM	20% AM	30% AM	40% AM	Differences
			Contact Time (s)			
Walk	0.64±0.04	0.63±0.04	0.63±0.04	0.63±0.04	0.64±0.04	
Run*	0.25±0.02	0.26±0.02	0.26±0.02	0.27±0.02	0.28±0.02	[0]<[10,20,30,40]
						[10]<[30,40]
						[20]<[40]
			Stride Time (s)			
Walk*	1.05±0.06	1.05±0.06	1.04±0.06	1.05±0.06	1.06±0.06	[20]<[40]
Run*	0.72±0.05	0.72±0.05	0.73±0.06	0.74±0.05	0.75±0.05	[0,10]<[30,40]
						[20]<[40]

*significant main effect, p<0.05.

Stride Time

Stride times were affected by AM condition during both walking and running. During walking, stride time was less during the 20% AM condition than the 40% AM condition. During running, stride times were longer as mass was increased. The stride times were longer during the 30% and 40% AM conditions than the control and 10% AM conditions. The 40% AM condition stride time was longer than the 20% AM condition (see Table 4).

GROUND REACTION FORCES

The addition of mass was found to affect peak impact force and loading rate during both walking and running. Peak propulsive force was only affected during walking and impulse was only affected during running (see Figure 5). Table 5 shows the mean peak impact force, peak propulsive force, loading rate, and impulse for each AM condition during walking and running.

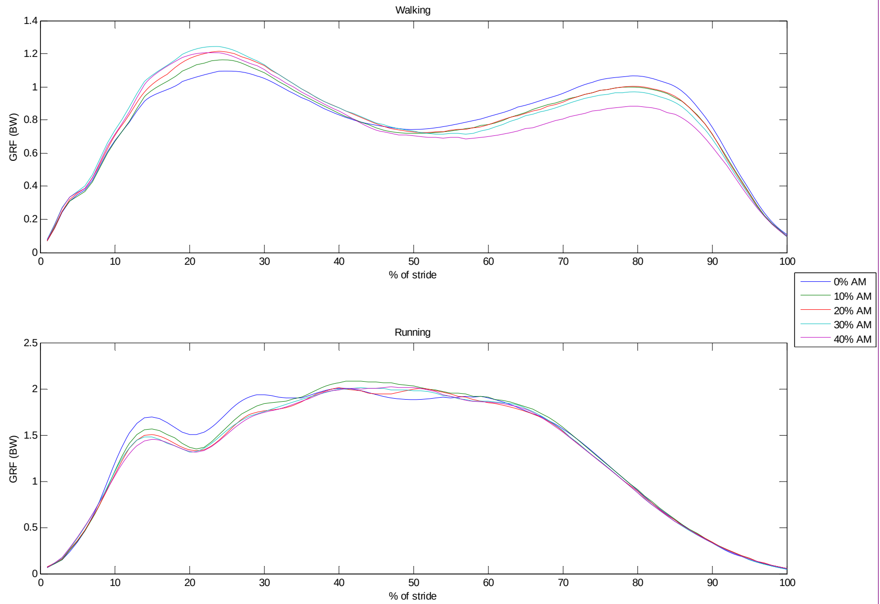

Figure 5. Normalized mean ensemble ground reaction force trajectories during walking and running at all AM levels.

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Table 5. Peak impact force, loading rate, peak propulsive force, and impulse (Mean±SD) for each AM condition for walking and running.

Speed	0% AM	10% AM	20% AM	30% AM	40% AM	Differences
			Peak Impact Force (BW)			
Walk*	1.13± 0.05	1.19±0.06	1.24±0.10	1.28±0.10	1.24±0.07	[0]<[20,30,40]
						[10]<[30]
Run*	1.75±0.17	1.62±0.13	1.54±0.15	1.52±0.15	1.47±0.15	[20,30,40]<[0]
			Loading Rate ($BW•s^{-1}$)			
Walk*	7.41± 0.62	8.08±0.51	8.79±0.94	9.50±1.40	9.02±1.17	[0]<[20,30,40]
						[10]<[40]
Run*	48.80±9.06	43.79±6.98	40.84±5.35	39.61±8.49	36.18±6.40	[20,30,40]<[0]
						[40]<[10]
			Peak Propulsive Force (BW)			
Walk*	1.10± 0.04	1.02±0.03	1.02±0.08	0.99±0.10	0.90±0.09	[30,40]<[0]
						[40]<[10,20,30]
Run	2.23±0.13	2.27±0.13	2.21±0.23	2.23±0.13	2.21±0.16	
			Impulse (BW•msec)			
Walk	524.21± 48.52	509.19±41.13	517.92±56.84	520.62±70.95	496.06±50.19	
Run*	324.49±23.29	333.99±21.83	330.52±23.28	338.04±28.56	342.67±31.69	[0]<[40]

*significant main effect, p<0.05.

Peak Impact Force

During walking, peak impact forces increased from 0% AM up to 30% AM, and then decreased at 40% AM. Peak impact forces at 20%, 30% and 40% AM were greater than without added mass, and those at 30% AM were greater than at 10% AM.
During running, a different phenomenon occurred. Peak impact forces decreased as AM increased. Peak impact forces during running with no added mass were greater than during 20%, 30% and 40% AM. There were no differences between added mass conditions.

Loading Rate

As with peak impact force, during walking, loading rates were lower without added mass than during the 20%, 30% and 40% AM conditions. Loading rates during the 10% AM was less than during 30% AM. During running, loading rates were greater without added mass than during 20%, 30% and 40% AM, and loading rates during the 10% AM were greater than during 40% AM.

Peak Propulsive Force

During walking, peak propulsive forces tended to decrease as AM increased. Peak propulsive forces during the control condition were greater than during 30% and 40% AM conditions. Peak propulsive forces during 40% AM were less than during all other AM conditions. There was no AM effect during running on peak propulsive force.

Impulse

There was a significant AM effect upon impulse during running, but not during walking. Mean impulse ranged from 496 – 524 BW·msec during walking and from 324 – 342 during running. During running, the mean impulse during the control condition was less than that developed during the 40% AM condition.

JOINT KINEMATICS

Joint motion was similar during walking and running between AM conditions (Fig. 6). Positive angles represent hip flexion, knee extension and ankle plantar flexion. Joint angle profiles suggest that range of motion increases between walking and running, as expected.

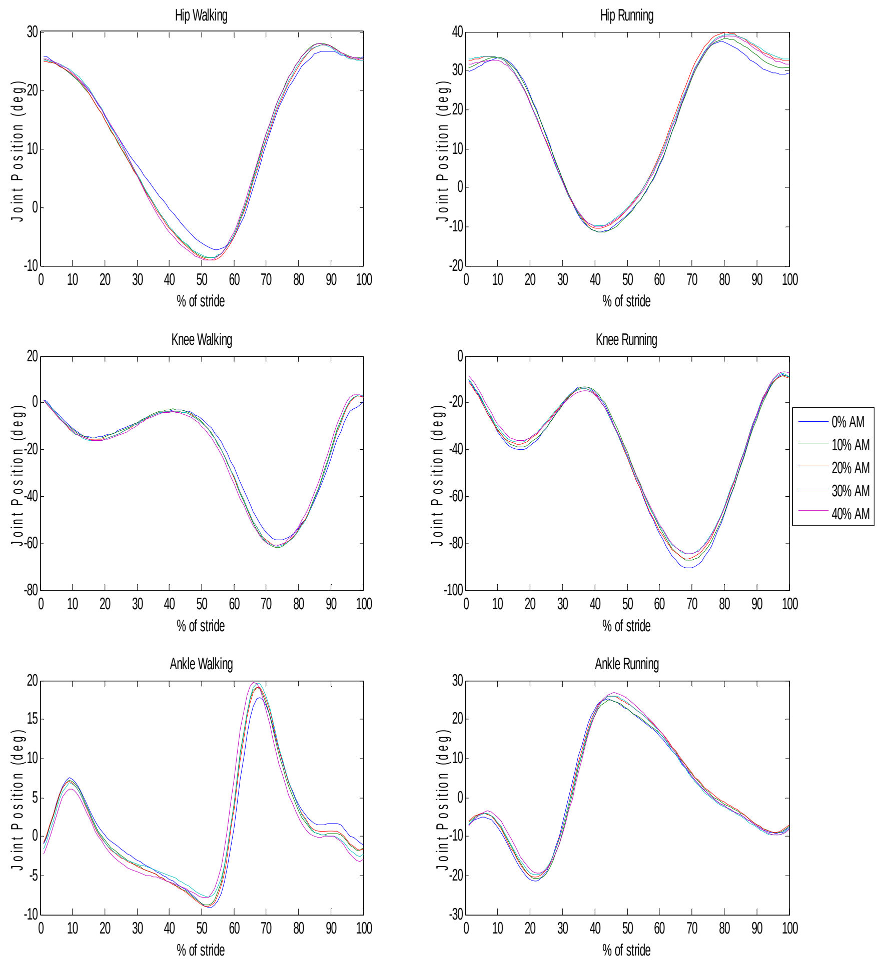

Figure 6. Mean hip, knee and ankle joint trajectories during walking and running with added mass.

JOINT KINETICS

Joint torques were computed using kinematic and ground reaction force data. The net joint torque is a measure of the net muscle action about each joint. Each joint torque curve has positive and negative phases. Positive torques represents hip and knee extension, and ankle plantarflexion. Angular impulse was computed as the area under the positive or negative phases of the joint torque curve.

Net work was computed as the area under the power curve. Joint powers were also computed using kinematic and ground reaction force data. Power was the product of the torque and angular velocity of each joint. The power curve for a given joint also has positive and negative phases. Positive values represent a net joint torque directed in the same direction as the joint angular velocity. Negative values represent net joint torque directed opposite as the joint angular velocity. Positive and negative phases of the power curves were analyzed separately. During the angular impulse and work analysis, the stance and swing phase of each stride were analyzed separately.

Figures 7-9 illustrate the ensemble averages of joint angular velocity, joint torque and joint power for the entire subject pool. The plots are the ensemble averages for each variable for all subjects. Tables 6-9 show the mean peak values for each variable over the entire stride.

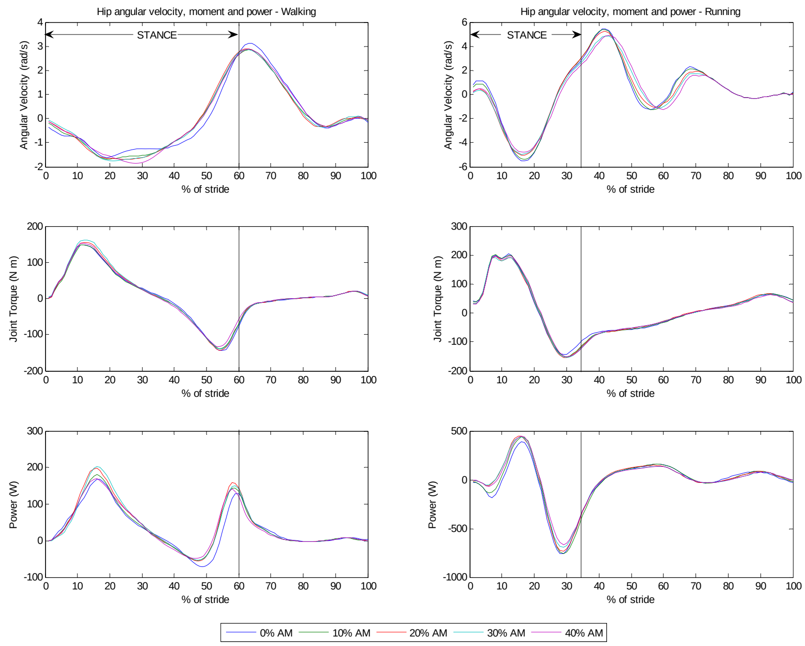

Figure 7. Ensemble average trajectories of the angular velocity, net torque, and power at the hip joint during walking (left) and running (right) at all added mass conditions.

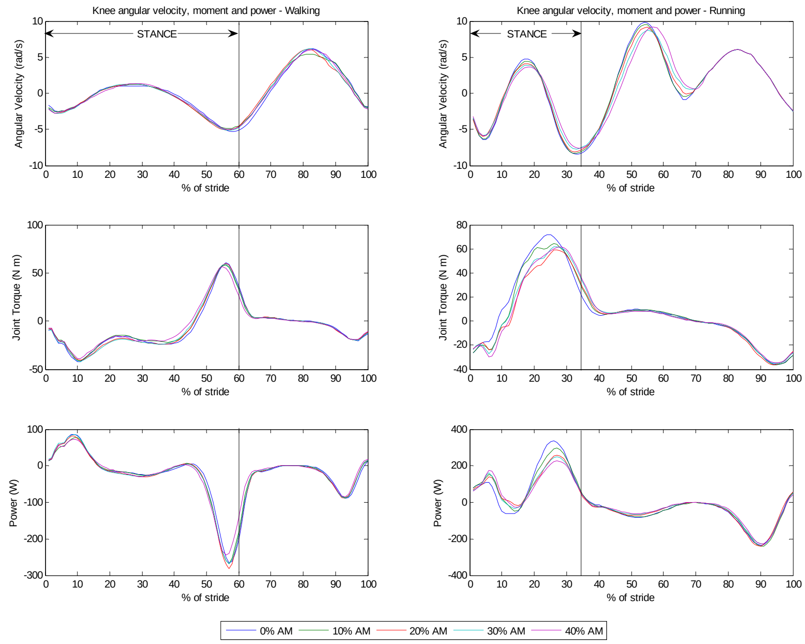

Figure 8. Ensemble average trajectories of the angular velocity, net torque, and power at the knee joint during walking (left) and running (right) at all added mass conditions.

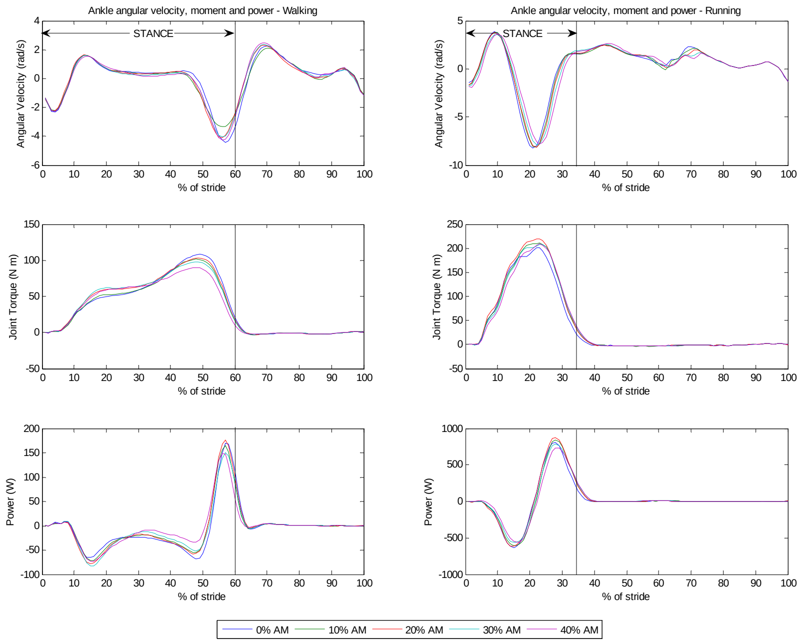

Figure 9. Ensemble average trajectories of the angular velocity, net torque, and power at the ankle joint during walking (left) and running (right) at all added mass conditions.

Table 6. Flexion and extension angular impulse angular impulse (Mean±SD) during the stance phase running at each AM condition.

Speed	0% AM	10% AM	20% AM	30% AM	40% AM	Differences
Hip Extension (N•m•s)						
Walk*	26.58 ± 8.61	25.71 ± 6.99	27.56 ± 8.04	28.98 ± 9.78	27.19 ± 7.58	[10]<[30]
Run	18.94 ± 5.72	19.57 ± 5.67	18.58 ± 5.46	19.17 ± 6.246	20.16 ± 6.74	
Hip Flexion (N•m•s)						
Walk	-21.20 ± 6.17	-20.75 ± 5.49	-21.19 ± 6.21	-21.17 ± 5.85	-20.54 ± 5.52	
Run*	-13.27 ± 3.01	-14.49 ± 3.33	-15.35 ± 3.37	-15.73 ± 3.17	-15.63 ± 3.50	[0]<[20,30,40]
Knee Extension (N•m•s)						
Walk	6.06 ± 1.55	6.05 ± 1.38	6.22 ± 1.44	6.35 ± 1.62	6.07 ± 1.43	
Run	9.29 ± 3.21	8.95 ± 3.38	7.84 ± 2.92	8.89 ± 3.81	8.87 ± 3.49	
Knee Flexion (N•m•s)						
Walk	-11.42 ± 5.27	-10.44 ± 3.86	-11.34 ± 4.49	-11.53 ± 4.51	-10.10 ± 5.02	
Run*	-1.16 ± 0.57	-1.61 ± 0.97	-1.64 ± 0.88	-1.72 ± 0.98	-1.96 ± 1.07	[40]<[0]
Ankle Plantar Flexion (N•m•s)						
Walk	37.09 ± 10.66	35.57 ± 7.84	37.58 ± 9.75	3706 ± 10.32	35.00 ± 10.67	
Run	27.62 ± 7.69	29.74 ± 10.11	31.36 ± 9.00	30.12 ± 9.01	29.94 ± 9.70	
Ankle Dorsiflexion (N•m•s)						
Walk	-0.07 ± 0.02	-0.06 ± 0.01	-0.06 ± 0.02	-0.06 ± 0.03	-0.06 ± 0.02	
Run	-0.06 ± 0.02	-0.06 ± 0.02	-0.05 ± 0.02	-0.06 ± 0.03	-0.06 ± 0.03	

*significant main effect, $p<0.05$.

Table 7. Positive and negative work (Mean±SD) during the stance phase for walking and running at

Speed	0% AM	10% AM	20% AM	30% AM	40% AM	Differences
Hip Positive Work (J)						
Walk*	38.54 ± 9.85	41.64 ± 7.82	45.56 ± 9.48	45.62 ± 13.37	41.82 ± 9.35	[0]<[30,40]
Run*	18.87 ± 4.01	24.27 ± 5.37	26.95 ± 6.20	27.79 ± 5.81	28.11 ± 7.24	[0]<[10,20,30,40]
Hip Negative Work (J)						
Walk*	-7.40 ± 3.03	-5.60 ± 2.76	-5.55 ± 2.63	-5.87 ± 2.64	-5.49 ± 1.77	[0]<[10,20,30,40]
Run*	-63.89 ± 14.87	-62.10 ± 18.56	-57.54 ± 18.26	-57.54 ± 18.26	-55.12 ± 17.59	[0]<[30,40]
Knee Positive Work (J)						
Walk	9.04 ± 5.05	8.29 ± 4.59	8.61 ± 4.19	9.32 ± 4.51	8.20 ± 4.14	
Run	33.07 ± 7.97	32.48 ± 9.71	28.46 ± 9.65	29.02 ± 9.59	28.73 ± 9.78	
Knee Negative Work (J)						
Walk	-29.35 ± 8.90	-29.39 ± 6.27	-31.18 ± 7.62	-30.94 ± 9.46	-28.09 ± 7.60	
Run	-4.05 ± 2.97	-3.19 ± 2.40	-2.48 ± 1.73	-3.06 ± 1.63	-2.41 ± 1.15	
Ankle Positive Work (J)						
Walk	12.30 ± 2.55	12.03 ± 2.55	12.92 ± 3.16	11.59 ± 3.21	11.37 ± 3.65	
Run	51.79 ± 18.73	55.28 ± 24.44	59.57 ± 20.70	55.81 ± 19.42	52.01 ± 22.13	
Ankle Negative Work (J)						
Walk*	-18.56 ± 7.67	-16.86 ± 6.02	-17.92 ± 5.84	-17.39 ± 7.99	-14.48 ± 4.79	[0,20]<[40]
Run	-39.96 ± 9.55	-40.51 ± 10.39	-40.68 ± 11.07	-38.06 ± 10.90	-37.92 ± 10.77	

*significant main effect, p<0.05.

Table 8. Flexion and extension angular impulse (Mean±SD) during the swing phase for walking and condition.

Speed	0% AM	10% AM	20% AM	30% AM	40% AM	Differences
Hip Extension (N•m•s)						
Walk	2.37 ± 0.67	2.37 ± 0.69	2.35 ± 0.82	2.42 ± 0.96	2.43 ± 0.90	
Run*	8.63 ± 1.77	8.38 ± 1.87	8.70 ± 1.89	8.38 ± 1.76	8.07 ± 1.84	[0,20]>[40]
Hip Flexion (N•m•s)						
Walk	-0.75 ± 0.41	-0.70 ± 0.38	-0.66 ± 0.39	-0.66 ± 0.26	-0.69 ± 0.35	
Run*	-7.61 ± 1.94	-7.05 ± 1.59	-6.71 ± 1.79	-6.54 ± 1.67	-6.57 ± 1.64	[0]<[10,20,30,40]; [10]<[30,4
Knee Extension (N•m•s)						
Walk*	0.30 ± 0.14	0.27 ± 0.16	0.27 ± 0.15	0.26 ± 0.15	0.21 ± 0.13	[0,10,20]>[40]
Run*	1.20 ± 0.37	1.15 ± 0.42	1.04 ± 0.37	1.01 ± 0.37	0.97 ± 0.39	[0]>[20,30,40]; [10]>[30,40]
Knee Flexion (N•m•s)						
Walk	-2.12 ± 0.40	-2.14 ± 0.41	-2.13 ± 0.42	-2.13 ± 0.44	-2.16 ± 0.42	
Run	-3.88 ± 0.81	-3.87 ± 0.83	-4.02 ± 0.80	-3.93 ± 0.67	-3.83 ± 0.76	
Ankle PlantarFlexion (N•m•s)						
Walk	0.07 ± 0.03	0.07 ± 0.03	0.07 ± 0.03	0.07 ± 0.03	0.07 ± 0.03	
Run*	0.13 ± 0.04	0.13 ± 0.05	0.13 ± 0.04	0.13 ± 0.04	0.12 ± 0.04	[0,10]>[40]
Ankle Dorsiflexion (N•m•s)						
Walk	-0.28 ± 0.10	-0.27 ± 0.10	-0.26 ± 0.09	-0.27 ± 0.09	-0.27 ± 0.09	
Run	-0.47 ± 0.16	-0.47 ± 0.15	-0.46 ± 0.16	-0.46 ± 0.17	-0.46 ± 0.16	

*significant main effect, p<0.05.

Table 9. Positive and negative work (Mean±SD) during the swing phase for walking and running at e

Speed	0% AM	10% AM	20% AM	30% AM	40% AM	Differences
Hip Positive Work (J)						
Walk	2.94 ± 0.98	2.70 ± 1.18	2.48 ± 1.14	2.61 ± 1.30	2.59 ± 1.17	
Run	25.37 ± 7.94	25.27 ± 6.92	24.70 ± 6.55	23.07 ± 7.19	22.86 ± 6.23	
Hip Negative Work (J)						
Walk	-0.20 ± 0.14	-0.26 ± 0.34	-0.22 ± 0.22	-0.31 ± 0.28	-0.25 ± 0.23	
Run	-2.82 ± 1.36	-2.57 ± 1.39	-2.42 ± 1.26	-2.52 ± 1.32	-2.20 ± 1.20	
Knee Positive Work (J)						
Walk	0.48 ± 0.26	0.47 ± 0.15	0.53 ± 0.20	0.53 ± 0.21	0.55 ± 0.28	
Run	1.59 ± 0.69	1.56 ± 0.96	1.67 ± 0.98	1.68 ± 1.07	1.42 ± 0.83	
Knee Negative Work (J)						
Walk	-8.36 ± 1.31	-8.38 ± 1.87	-8.14 ± 1.73	-8.18 ± 2.09	-7.90 ± 1.80	
Run*	-29.06 ± 5.92	-28.41 ± 5.59	-27.78 ± 5.92	-26.74 ± 6.26	-25.88 ± 6.39	[0]<[30,40]
Ankle Positive Work (J)						
Walk	0.29 ± 0.10	0.29 ± 0.11	0.30 ± 0.12	0.30 ± 0.12	0.30 ± 0.12	
Run	0.84 ± 0.25	0.83 ± 0.25	0.78 ± 0.21	0.82 ± 0.26	0.81 ± 0.26	
Ankle Negative Work (J)						
Walk	-0.10 ± 0.06	-0.12 ± 0.09	-0.10 ± 0.07	-0.10 ± 0.05	-0.11 ± 0.05	
Run	-0.08 ± 0.04	-0.07 ± 0.03	-0.09 ± 0.03	-0.08 ± 0.04	-0.08 ± 0.04	

*significant main effect, $p<0.05$.

Hip Joint Angular Impulse

During the stance phase, the repeated measures ANOVA found a significant main effect of added mass upon the hip extensor angular impulse during walking, but not during running. Hip extensor angular impulse during walking at 30% AM was greater than 10% AM. While there was no effect of AM upon hip flexor impulse during walking, there was an effect of AM during running. Hip flexor angular impulse during running was greater at 20%, 30% and 40% AM than at 0% AM.

During the swing phase, the repeated measures ANOVA found a significant main effect of added mass during running for both hip extensor and hip flexor angular impulse. There was no main effect of additional mass during walking. Hip extensor angular impulse at 40% AM was less than during 0% and 20% AM. Hip flexor impulse at 10%, 20%, 30% and 40% AM were less than during the 0% AM condition. In addition, the hip flexor impulse at 30% AM and 40% AM was less than during the 10% AM condition.

Knee Joint Angular Impulse

During the stance phase, the repeated measures ANOVA found a main effect of AM for knee flexor angular impulse during running, with no effect during walking. Knee flexor angular impulse at 40% AM was greater than during 0% AM.

During the swing phase, a significant main effect of AM was found upon knee extensor angular impulse during both walking and running. The extensor angular impulse during walking was less at 40% AM than during 0%, 10% and 20% AM conditions. During running, the extensor angular impulse was less at the 30% and 40% AM conditions than during the 0% and 10% conditions, and less at the 20% AM condition than the 0% AM condition.

Ankle Joint Angular Impulse

The repeated measured ANOVA found that during the stance phase, there was no effect of added mass upon ankle plantarflexion and dorsiflexion angular impulse. During the swing phase, there was a main effect of added mass found upon ankle plantarflexion angular impulse during running. The plantarflexion angular impulse at 40% AM was less than during the 0% and 20% AM conditions.

Hip Joint Work

The repeated measures ANOVA found that the addition of mass affected the positive work during the stance phases of both walking and running. During walking, the positive work at the hip was greater at 20% and 30% AM than at 0% AM. During running, the positive work at 10%, 20%, 30% and 40% AM was greater than at 0% AM. Additional mass also affected the negative work at the hip during the stance phase of walking and running. The magnitude of negative work during walking was less at 10%, 20%, 30% and 40% AM than at 0% AM. The magnitude of negative work during running was less at 30% than at 0%. In addition, the magnitude of negative work was less at 40% AM than at 0% and 10% AM. There was no effect of AM upon hip work during the swing phase.

Knee Joint Work

There was no effect of additional mass upon the work at the knee during the stance phase of walking or running. There was an effect of additional mass upon negative work at the knee during the swing phase of running. Negative work was less at 40% AM than during 0% AM and 10% AM. There were no other effects of additional mass on the positive or negative work at the knee during the swing phase.

Ankle Joint Work

There was an effect of additional mass upon the negative work at the ankle during the stance phase of walking. Negative work at 40% AM was less than during 0% and 20% AM. There were no effects of additional mass upon the positive work at the ankle during the stance phase at either speed, or during the stance phase of running. There were also no effects of additional mass upon positive or negative work at either speed during the swing phase.

DISCUSSION

The purpose of this investigation was to determine how adding mass while maintaining body weight affected kinematic and kinetic measures during walking and running. The primary finding of this investigation was that the addition of mass resulted in increased peak impact forces and loading rates during walking, but decreased peak impact forces and loading rates during running. In addition, peak propulsive forces during walking decreased with added mass. In both locomotive modes, positive work at the hip increased as mass was added. However, the increase in positive work was due to increased hip extensor torque during walking, and increased hip flexor torque during running. The adaptations to additional mass during walking were different than those during running.

SUMMARY OF RESULTS

Table 10 illustrates the trends that occurred when a main effect of AM was found for each dependent variable. When the terms increasing or decreasing are used, they are meant to compare the effect of additional mass upon the variable being discussed as compared to the 0% AM condition.

Table 10. Summary of the significant main effects of AM upon all dependent variables during walking and running when compared to 0% AM

	Walking	Running
Gait parameters		
Stride Time	↑	↑
Contact Time	No Change	↑
Ground Reaction Forces		
Peak Impact Force	↑	↓
Loading Rate	↑	↓
Peak Propulsive Force	↓	
Impulse	No Change	↑
Joint Torque and Power - Stance		
Hip Extensor Impulse	↑	No Change
Hip Flexor Impulse	No Change	↑
Knee Flexor Impulse	No Change	↑
Hip Positive Work	↑	↑
Hip Negative Work	↓	↓
Ankle Negative Work	↓	No Change
Joint Torque and Power - Swing		
Hip Extensor Impulse	No Change	↓
Hip Flexor Impulse	No Change	↓
Knee Extensor Impulse	↓	↓
Ankle Plantarflexor Impulse	No Change	↓
Knee Negative Work	No Change	↓

Walking

During walking, stride time, impact forces, and loading rates increased as mass was added. Peak propulsive forces decreased with additional mass, but impulse was not affected.

Stance phase hip extensor torques increased as mass was added, with the 30% AM condition having the largest angular impulse. Positive work at the hip increased and negative work decreased. Negative work at the ankle joint also decreased as mass was added. During the swing phase, knee extensor torques decreased with the addition of mass. There were no differences at any joint in positive or negative work during the swing phase.

Running

During running, contact time increased and stride time increased as mass was added. Peak impact forces and loading rates decreased, and impulse increased as mass was added. There were no effects of additional mass on peak propulsive forces.

During the stance phase, hip and knee flexor angular impulse increased with additional mass. Positive work at the hip increased and negative work decreased. During the swing phase, hip extensor, hip flexor, knee extensor and ankle plantarflexor impulses all decreased as mass was added. Negative work at the knee decreased; otherwise, there were no added mass effects on any of the other work dependent variables.

GAIT PARAMETERS

During this investigation, the gait parameters that were investigated were contact time and stride time. Contact time was defined as the interval between heel strike and toe off for each step. Swing time was defined as the time during the stride that the foot is not in contact with the ground. Examination of these variables determined if general modifications in motor patterns occurred.

Walking

The longer stride time at the 40% AM condition compared to the 0% AM condition indicates that the gait cycle time was only affected by the largest increase in mass tested. The increase in stride time suggests that since contact time did not increase, swing time increased to adapt to the increase in mass. It is probable that the increase in stride time resulted in an increase in stride length since treadmill speed was constant between AM conditions.

Donelan and Kram (1997) found that when gravity levels were reduced using overhead suspension, contact time decreased and step lengths shortened during walking at speeds between 0.75 and 1.75 m s^{-1}. Swing time did not change as overhead support increased. In their experiment, the effective gravity was modified by lifting the subject upwards from the treadmill. Griffin, Tolani and Kram (1999) also found that stride frequency, which is the inverse of stride time, did not undergo significant changes with the reduction of gravity using an overhead suspension system during walking at 1.0 m s^{-1}. Finch, Barbeau and Arsenault (1991) found similar results during their investigation of the influence of body weight support on gait.

The findings of this study suggest that the contact time for a given speed may be dependent upon the net force between the subject and the ground. This is reasonable since Donelan and Kram (1997) and Griffin et al. (1999) employed an overhead suspension method similar to that used during this experiment. In their experiments, as the force between the subject and ground decreased, contact time was affected, but stride time was not. In our experiment, although mass was added, the net force between the ground and subject was maintained using overhead suspension, and contact time did not change.

Another way to compare the studies would be to recognize that in each investigation, gravity was reduced. In the two aforementioned studies, mass was maintained. In the present study, mass was increased. Taken together, the results of this study suggest that weight, and not mass, affect contact time during walking. Stride time, however, may be dependent upon mass, since increases occurred due to the addition of body weight and mass. It is possible that inertial increases due to increases in mass result in changes in stride time, while contact time is optimized for a given speed.

Running

The results of our investigation indicate that the adaptations in contact time and stride time to added mass during running are different than those that occur during walking. Furthermore, adaptations to an increase in mass, or inertial forces, differ from adaptations to increases in body weight, or gravitational forces. In contrast to walking, contact time increased during running as mass was added. Similar to walking, however, stride time also increased. When mass was increased, Chang et al. (2000) found similar increases in contact time and stride time.

The fact that adding mass increases contact time and stride time, while maintaining mass and decreasing weight decreased these times suggests that the effective gravity level is not the controlling factor of motor patterns during running.

The increased contact time that occurs with an increase in mass may allow the lower body musculature more time to develop force to propel the body upward. Rather than increasing force, contact time is increased to allow the forces to act for a longer period. The mass of the person, which affects the inertial forces acting upon the body, may be the critical control factor.

GROUND REACTION FORCES

The ground reaction force is the net force acting upon the center of mass of the body during locomotion, and reflects the acceleration of the center of mass of the body (Munro, Miller & Fuglevand, 1987). The primary purpose of this investigation was to determine if the addition of mass to the body resulted in an increase in ground reaction force parameters. The increases in mass were hypothesized to increase ground reaction forces because a greater total mass would be vertically accelerated and decelerated during the gait cycle. The hypothesis was reasonable because inertial mass was not altered during body weight suspension. Although body weight was maintained, greater mass would need to be propelled upward and decelerated downwards, resulting in greater ground reaction forces.

Walking

For walking, the added mass led to an increase in vertical impact forces and loading rates. However, vertical propulsive forces decreased, and impulse was unaffected. The increase in impact forces and loading rates followed a linear dose-response relationship up to 30% of additional mass. There was a slight decrease in impact forces and loading rates from 30% additional mass to 40% additional mass, which suggests that a threshold effect may occur at additional masses greater than 30% of normal mass.

Stride time was significantly greater during the 40% AM condition than the 20% AM condition, and tended to be longer during the 40% AM condition than all other conditions. It is possible that there was a critical force level that was obtained at 30% AM, and that when more mass was added, motor patterns were modified to eliminate any additional increases in impact force. It is not clear, however, why this threshold effect occurred.

The increase in vertical impact force, coupled with the decrease in vertical propulsive force, suggests that any adaptations made to additional mass occurred during the latter phases of contact. The increase in mass resulted in greater forces necessary to decelerate the body as the foot contacted the ground. The lack of adaptation to increased mass during the initial phase of stance resulted in a dose-response relationship between impact force and additional mass. However, the decrease in ground reaction force during the propulsive phase of ground contact suggests, however, that some adaptation had occurred that affected the latter phases of stance. It is also possible that the increase in mass caused a difficulty in controlling the trunk. Therefore, subjects decreased propulsive forces in order to increase the control of the trunk.

Running

During running, peak vertical impact forces and loading rates decreased as mass was added. Peak vertical propulsive forces did not change. Vertical impulse increased as mass was added. The acceleration of the center of mass is directly related to ground reaction force magnitude and the amount of time for the forces to act. It is possible that the adaptation of increased contact time as mass was added occurred to allow peak impact forces to decrease. The increased contact time allowed vertical forces to act for a longer period of time. Chang et al. (2000) demonstrated that impact peaks occur during the initial 20% of foot-ground contact time. Although peak impact forces decreased as mass was added, it is over a relatively small proportion of the entire ground reaction force trajectory; therefore, it is possible that total change in ground reaction force was not enough to cause a decrease in impulse.

We hypothesized that increasing mass would result in an increase in ground reaction forces. During running, peak vertical propulsive forces did not increase. Chang (2000) found the same phenomenon. It was anticipated that vertical propulsive forces would be increased at the highest added mass level because Chang only investigated up to 30% of additional mass. The fact that no differences were found even at 40% of additional mass suggests that the adaptations that occur during running with increased mass are geared towards maintaining consistent ground reaction force magnitudes regardless of AM level.

JOINT KINETICS

Walking

The shapes of the joint torque curves for the hip and ankle found during the stance phase of walking with added mass are in general agreement with those reported in past literature during overground walking (DeVita, Torry, Glover & Speroni, 1996; DeVita & Hortobagyi, 2000; Koopman, Grootenboer & de Jough, 1995; Eng & Winter, 1995). For the hip, the joint torque is generally a sinusoidal shape, with peak extensor moments occurring during the first half of stance and peak flexor moments occurring during the latter half of stance. The ankle is dominated by plantarflexor torques throughout the entire stance phase.

The joint torque found at the knee during this investigation, however, was different than that reported in the literature during overground walking. We found flexor and extensor torques present during middle and latter stance, respectively. However, the initial portions of stance were dominated by knee flexor torque. The general pattern of knee torque during stance that is reported in the literature is similar to that reported at the hip (DeVita et al., 1996; DeVita & Hortobagyi, 2000). An extensor moment dominates the initial portion of stance, followed by a flexor moment during midstance. As the toe off approaches, extensor torque dominates as the knee extends during pushoff.

One major difference in our study is the absence of anteroposterior ground reaction force measurement. It is probable that inclusion of this force would alter the computed torques at the joints. A simulated anteroposterior ground reaction force was added to a trial to determine if the absence of this force component significantly affected the joint torques. The general pattern and magnitude of the simulated anteroposterior ground reaction force was similar to that reported by Munro et al. (1987). Although knee joint torque trajectory changed to match that reported by DeVita & Hortobagyi (2000), the hip joint torque trajectories also changed. For the hip, flexor torques dominated the initial portion of stance, and extensor torque was present during the latter portion of stance. These findings suggest that the joint torque patterns that occur during walking on a treadmill differ from those during overground walking. It is possible that the rearward motion of the treadmill belt causes adaptations in the joint torques at the knee and hip.

During the stance phase, hip extensor angular impulse increased as mass was added to the subject. Specifically, hip extensor angular impulse was greater during the 30% AM condition than during the control condition. None of the other kinetic variables were affected during the stance phase of walking. The positive work at the hip increased and negative work decreased as mass was added. In addition, negative work at the ankle decreased as mass increased. It is possible that a threshold was obtained at 40% AM that required a change in adaptation strategy since positive work at the hip was less than the 30% AM condition.

These results suggest that the primary adaptation to additional mass during walking was to increase the torques of the hip extensors. DeVita & Hortobagyi (2000) found that healthy elders used more hip extensor torque than a younger population during walking. However, they also found less knee extensor and ankle plantarflexor torques in the older group than the younger cohort. They concluded that aging causes a redistribution of joint torques among the joints of the lower leg. It does not appear that the increases in hip torques found during this investigation accompanied decreases at the other joints, suggesting an adaptation strategy that differed from that used during aging.

An increase in hip extensor torque and positive work occurred during the initial phase of stance. The increase may be a result of the hip musculature producing greater torque to reverse the downward trajectory of the center of mass during impact. Hip extensor torque early in the stance phase should cause a movement of the thigh underneath the body, thus reducing the downward motion. It is possible that co-contractions occur at the knee and ankle to aid in stability. However, a limitation of inverse dynamics is the inability to detect muscular activity increases due to contraction where net motion does not occur. Future investigations utilizing electromyography may help answer this question.

During the swing phase, the only effect of added mass was a decrease in knee extensor impulse. Knee extensor torques generate forward motion of the shank with respect to the thigh. This probably occurs during the swing to cause knee extension. It is not clear from the present data if there is a specific phase of the swing where knee extensor torque decreased. However, the only AM condition where a difference occurred was at the 40% AM condition. Knee extensor torque at 40% AM was less than torque at

0%, 10% and 20% AM. The condition exhibiting decreased knee extensor torque was the same condition in which stride time increased. It is possible that the increase in stride time was due to the decreased knee extensor torque.

Running

The shapes of the joint torque curves for the hip and ankle during running were similar to those reported in the literature (DeVita et al., 1996; DeVita & Hortobagyi, 2000). However, unlike walking, joint torque curves computed at the knee were also similar to those reported by these authors. The absence of the anteroposterior component of the ground reaction force may not affect the kinetic computations during running as much as it does during walking. This may be due to the shorter contact time during running, and that the leg extends less during running, resulting in less anteroposterior braking and propulsion.

Running on a treadmill may be more similar to running overground than walking on a treadmill is to overground walking. Nigg, DeBoer and Fisher (1995) reported that systematic kinematic differences exist between overground running and treadmill running, but that subject dependent differences were more substantial. The systematic differences were related to a flatter foot placement on the ground during treadmill running. The differences in kinetics during stance may also not be as affected by treadmill locomotion as the kinematics of landing.

During running, hip extensor angular impulse and knee flexor angular impulse magnitudes increased as mass was added. The adaptation in hip extensor impulse appears to occur late in the stance phase. In contrast, the increase in knee flexor impulse appears to occur early in the stance phase. The increase in knee flexor angular impulse may be an active mechanism to reduce impact force. Increasing knee flexor activity could cushion the body by allowing the leg to absorb some of the impact. The increase in hip extensor torque late in the stance phase may be a response to the requirement to propel the greater mass. Compensations in the joint torque may have been used rather than modifications in the ground reaction forces, which explains why peak propulsive forces did not increase.

The increase in hip extensor angular impulse resulted in increased positive work at the hip as mass increased. As with walking, increases in mass resulted in decreased negative work at the hip. The hip provided more and absorbed less energy as mass was added to the body.

During the swing phase, hip extensor, hip flexor, knee extensor and ankle plantarflexor angular impulse decreased as mass was added. Negative work at the knee also decreased. The decrease in these torques may be an attempt to decrease ground reaction forces accepted and generated by the body. If the torques at the knee and hip were maintained, stride time may not have changed, but ground reaction forces may have increased. This suggests that the body may adapt to the increase in mass during running by altering joint torques to maintain ground reaction force.

The increase in hip extensor torque and work during walking without modifications in kinematics is consistent with the concept of non-univocality, which states that similar motions can be produced with different muscular activity patterns (Bernstein, 1967). If body weight were the main input when determining the motion patterns during locomotion, then it should be expected that no adaptations to increased mass would occur. However, since adaptations did occur during this investigation, mass must be a factor in the control process.

LIMITATIONS

A primary limitation in this study was the absence of ground reaction force measurement in the horizontal plane due to the inability to accurately measure shear forces on a treadmill belt. The absence of anteroposterior ground reaction forces during the inverse dynamics analysis may have affected the computed joint torques and power. There may also be changes in the horizontal braking and propulsive forces associated with walking and running that are affected by adaptations to added mass. These changes may be physiologically beneficial, but were undetectable in the current investigation.

It was not feasible to perform overground locomotion with added mass while maintaining body weight given current limitations in resources. The only way to complete this investigation while allowing subjects to run overground would have been to create a body weight suspension device that allowed horizontal translation.

It may be that the body weight support system caused adaptations due to the device itself rather than due solely to the addition of mass. During suspension, subjects wore a harness that connected to the lifting mechanism of the suspension device. While the manufacturer states that the vertical lift provided by the device is nearly constant, it is possible that there were small variations in the upward lift. These factors may have caused differences in adaptive patterns due to the suspension device, and not due to the increase in mass. However, all subjects completed trials under identical conditions and did not report any interference in locomotion. Therefore, these limitations were consistent across subjects and differences reported were due to the experimental conditions.

APPLICATIONS TO SPACEFLIGHT AND REHABILITATION

The results of this investigation indicate that the use of an added mass vest during treadmill exercise on the International Space Station could result in an increase in peak ground reaction forces during walking. The increase in force could enhance the osteogenic stimulus obtained during walking. During running, however, it does not appear that the use of added mass will increase ground reaction forces. Adaptations occur in joint torque and work at the hip, knee and ankle to maintain 0% AM ground reaction force magnitudes. However, adding mass may increase the work done at the hip during walking and running, which could be of benefit to the astronauts.

It is not clear if these results would occur in microgravity. In microgravity, the astronauts will not have their upper body supported; therefore, the addition of mass to the trunk may result in different adaptations due to stability requirements. In addition, forces are added to the astronaut to return them to the treadmill.

SUMMARY AND CONCLUSIONS

The specific purpose of this investigation was to determine if adding mass to astronauts during locomotive exercise in microgravity would be beneficial to increasing the forces experienced by the musculoskeletal system. It was hypothesized that the addition of mass would affect ground reaction forces and joint torque and work patterns of the lower extremities. It was found that the increase in mass increased impact forces and loading rates during walking. During running, impact forces decreased as mass was added. There were slight differences in kinematic adaptations between walking and running. These results suggest that the adaptive processes to increasing mass that occur during walking are different than running.

In both locomotive modes, hip musculature activity and positive work increased, suggesting that the hip is the primary area of adaptation of joint torque. It is possible that increasing mass during locomotive exercise in microgravity may be beneficial to increasing impact forces and hip extensor activity during walking. During running, there may be adaptations in locomotive patterns to minimize increases in ground reaction forces. Further investigations in actual microgravity should determine if similar adaptations occur in the absence of body weight support.

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