INDEPENDENT CONTRIBUTIONS OF SUPPORTING BODY WEIGHT AND ACCELERATING BODY MASS TO METABOLIC POWER DURING WALKING ON UPHILL AND DOWNHILL SLOPES

By

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Abstract:

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Independent contributions of supporting body weight and accelerating body mass on metabolic power during walking on uphill and downhill slopes

Thesis directed by Assistant Professor, Alena Grabowski

The metabolic cost of walking is partially determined by the muscle force needed to support body weight and the muscle work needed to redirect and accelerate the center of mass (CoM). During level-ground walking, ~28% of the net metabolic power (NMP) is due to supporting body weight and ~45% is due to performing CoM work. We hypothesized that supporting body weight would incur a greater percentage of NMP for uphill and a lower percentage of NMP for downhill compared to level-ground walking. Additionally, we hypothesized that performing work incur a greater percentage of NMP for uphill and a lower percentage of NMP for downhill compared to level-ground walking. We independently varied weight and mass and measured the metabolic cost of walking at 1.25 m/s on slopes of 0°, ±3°, and ±6° in 10 subjects (5 F). By calculating the metabolic power per newton of reduced body weight for each slope, we found that the percentage of NMP to support body weight was 63.1% and 70.7% on +3° and +6°, respectively, which were greater than 15% for level-ground walking ($p<0.025$). We found that the percentage of NMP to redirect the CoM was 18.9% and 23.1% on slopes of +3° and +6°, respectively, which were lower than 35% for level-ground walking ($p<0.025$). Our findings partially support our hypotheses, and elucidate that the percentage of NMP attributed to body weight support and mass redirection are different for slopes; compared to level ground, which informs biomimetic assistive device designs aimed at reducing metabolic demand.

Table of Contents

I. **Introduction**……………………………………………………………….........1

II. **Methods**…………………………………………………………………….........4

III. **Results**……..……………………………………………………………..…..…9

IV. **Discussion**………………………………………………………………..........15

V. **References**….………………………………………………………………........22

List of Tables

Tables

1. Net Metabolic Power (NMP) by Slope and Condition.......…………………………..11

2. R-squared values of Regression lines……………..………………………………...13

3. Average Percentage of NMP for Mass Redirection and BW Support…………..…..13

4. Absolute NMP for Mass Redirection, BW Support and Unexplained……………...18

5. NMP accounted for by Reduced Gravity Apparatus Vector Components………..……….20

List of Figures

Figures

1. Simulated Reduced Gravity Apparatus… ………………………………………………6

2. Net metabolic Power on 5 Slopes……………………………………..…...…...….12

3. Average Percentage of Net Metabolic Power on Slopes ………………..…………..….14

4. Parallel and Perpendicular forces of Simulated Reduced Gravity on Slopes…..……..…20

I. Introduction

There are two main biomechanical tasks that contribute to the metabolic cost of level ground walking. The generation of force to support body weight (BW) during the stance phase accounts for ~28% of the net metabolic power (NMP) to walk on level ground (17) and the redirection and acceleration of the center of mass (CoM) during the step-to-step transition accounts for ~45-47% of the NMP to walk on level ground (15, 17). Previously, the metabolic costs attributable to biomechanical tasks have been utilized to inform biomimetic powered ankle-foot prostheses (20). Thus, determining the percentage of NMP attributed to BW support and mass redirection during uphill and downhill sloped walking would further inform the design and function of biomimetic devices such as powered prostheses and orthoses, aimed at reducing the metabolic energy required for typical activities of daily living.

Previous studies have modeled the stance phase CoM dynamics of walking using an inverted pendulum, where the leg is represented as a rigid, massless strut that supports the CoM, which is represented as a point mass (6). During single limb support, CoM energy is conserved through the phasic exchange of kinetic and gravitational potential energies, which are approximately equal in magnitude and ~180° out of phase (6). This phasic energy exchange suggests that there is essentially zero net mechanical work done on the CoM during single limb support at a constant walking velocity on level ground (6, 16). However, metabolic energy is required by the leg muscles to generate force to support BW (17). To maintain constant average CoM velocity across multiple steps, simultaneous positive and negative mechanical work are required from the trailing and leading legs, respectively, to redirect and accelerate the CoM between steps (7, 8). Thus, the leg muscles consume metabolic energy to generate the mechanical work needed to redirect and accelerate the CoM during the step-to-step transition

(17). The remainder of the NMP required to walk on level ground is likely due to maintaining balance/stability and limb swing (9, 14).

The NMP attributed to body weight support and CoM redirection likely change during walking on different uphill and downhill slopes, but these contributions are unknown. The total NMP required to walk changes with slope, but does not exhibit a linear relationship across different slopes (28). Metabolic power is minimized at -6° and increases with more positive and negative slopes during walking (16, 26). The metabolic cost of walking on slopes can partially be attributed to the muscular force generated by the legs to raise or lower the CoM relative to the slope (27). Greater force must be generated by the muscles to walk uphill to overcome gravity and this exacts a higher metabolic cost than level-ground walking. Additionally, during uphill walking there is a net increase in gravitational potential energy (GPE) and thus an increase in the CoM height, which results in asymmetrical CoM energy fluctuations relative to the walking surface (13, 16). For downhill walking, there is a net decrease in GPE and thus a decrease in the CoM height. The unequal exchange of kinetic and gravitational potential CoM energies change the metabolic power requirements for sloped walking.

During the double support phase of sloped walking, individual leg net mechanical work increases by ~276% from 0° to +9° and decreases by ~84% from 0° to -9° (13, 23). Both legs simultaneously produce more positive power and absorb less negative power during uphill walking and both legs simultaneously absorb more negative power and produce less positive power during downhill walking (13). In addition step length increases with uphill walking (25) and decreases with downhill walking (24, 31). Further, NMP increases with slope (7) and it is likely that the percentage of NMP due to mass redirection during the step-to-step transitions increases during uphill compared to level-ground walking. The negative mechanical work performed by the legs during downhill walking, primarily eccentric muscle actions, has a relatively lower metabolic cost compared to the positive mechanical work performed by both legs during uphill walking, primarily concentric muscle actions (1). Based on previous studies, because the mechanical power and NMP change across slopes, the percentage of NMP attributed to generating force to support BW and producing work to redirect the CoM likely change.

The purpose of the current study is to determine the percentages of NMP attributed to weight and mass for different slopes. We independently varied weight and mass by using simulated reduced gravity and added loading, and determined the independent contributions of generating force to support body weight and performing work to redirect and accelerate the center of mass on the NMP required for walking on a range of slopes. We hypothesized that: 1) generating force to support BW would incur a greater percentage of the NMP during uphill walking and a lower percentage of the NMP during downhill walking compared to level ground. We also hypothesized that: 2) performing mechanical work would require a greater percentage of NMP during uphill walking and a lower percentage of NMP during downhill walking compared to level ground.

II. Methods

Ten healthy adults volunteered to participate (5 females, mean ± SD: body mass 67.8 ± 11.2 kg; age 24.9 ± 1.8 years; height 172.4 ± 8.5cm). All subjects gave informed written consent in accordance with the University of Colorado Boulder Institutional Review Board approved protocol.

Each subject performed a standing metabolic trial on level ground each day of the five-day protocol. Then on each day, subjects walked at 1.25 m/s on one of five slopes, which included 0°, ±3° and ±6°. For each slope, subjects completed a total of seven walking trials comprised of the following conditions: 1.0 Body Weight (BW) with 1.0 Body Mass (M), 0.75 BW with 1.0 M, 0.50 BW with 1.0 M, 1.25 BW with 1.25 M, 1.0 BW with 1.25 M, 1.50 BW with 1.50 M, and 1.0 BW with 1.50M. All of the conditions were in 0.25 increments of body weight and mass so that our data could be compared to those of previous studies (17). All subjects walked at +6° for their first day of trials to ensure they could complete the study while primarily utilizing aerobic metabolism (RER<1.1) (3). All subsequent days and trials were randomized, and completed on a dual-belt force-measuring treadmill (Bertec, Columbus, OH). Forces were measured at 1000 Hz and we used a vertical force threshold of 20 N to determine stride kinematics. Contact time was defined as the heel strike of one foot to the toe-off of the same foot.

To reduce each subject's BW, we applied an upward force on the body with a modified climbing harness attached to a frictionless pulley system in-series with pieces of rubber tubing that were stretched using a hand-cranked winch (Figure 1). This design is similar to previous studies that have used simulated reduced gravity (17, 19, 30). We used the winch to stretch the rubber tubing to at least 2.5 times its resting length and applied an upward force on each subject through an inextensible nylon cord. The stretched rubber tubing ensured that we applied a constant force throughout a walking stride as previously described by Griffin et al. (18) and Grabowski et al.(17). We positioned the pulley system directly above the subject on a frictionless trolley that rolled along an I-beam and moved with the subject, and thus only provided a vertical force against gravity. We measured the force that the pulley system exerted on the person with a force transducer (Omegadyne, Norwalk, CT) at 1000 Hz positioned at the end of the nylon cord, opposite the winch and verified this force with the ground reaction forces. We calibrated the force transducer to calculate the voltage associated with each force using a linear trend line. We used measurements from the force transducer and a custom Matlab code (Mathworks, Natick, MA), and verified forces with the ground reaction forces and a second custom Matlab code to determine the average force applied to each subject. The trolley was only capable of fore-aft

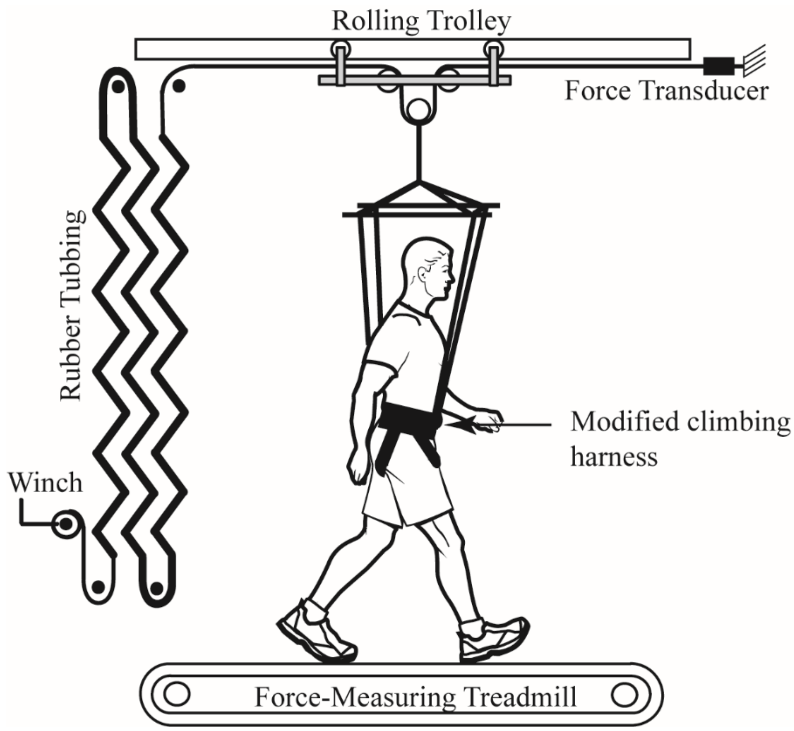

Figure 1. Simulated reduced gravity apparatus applied a constant upward force on the body's center of mass via a modified climbing harness (pictured in black). We adjusted the upward force by stretching the rubber tubing with a hand-cranked winch, thus applying a vertical force through the low-friction pulley system to the subject. The frictionless pulley system had the ability to move with the subject in the fore-aft direction as they walked on the treadmill. The applied vertical force was measured with a force transducer placed in-series with the rubber tubing.

motion and thus mitigated fore-aft forces and reduced gravity on the CoM, but not on the swinging limbs.

We added weight to each subject by wrapping lead strips tightly around their waist via a modified weight belt as close to their CoM as possible. To increase mass only, we added load using the weight belt and used simulated reduced gravity to apply an upward force equal to the added weight. This allowed us to determine the independent contribution of performing work to redirect and accelerate the center of mass on the NMP required for walking on a range of slopes.

We asked subjects to fast for at least two hours prior to each experimental session and each session was at the same time of day to control for day-to-day variability in metabolic rates. We measured subject's rates of oxygen consumption and carbon dioxide production throughout each 5-minute trial using an open-circuit respirometry system (Parvo, Sandy, UT). We averaged these metabolic rates during the last 2 minutes of each trial and then calculated metabolic power using a standard equation (5). Previous studies have determined that standing metabolic rate is not influenced by reduced gravity (10) or added load (18), so we subtracted standing metabolic power from walking metabolic power to determine NMP for each trial. To ensure that all subjects were primarily utilizing oxidative metabolism, we confirmed that their respiratory exchange ratios were less than 1.1 (3) during the duration of all trials. Between each trial, we required subjects to rest for a minimum of five minutes to mitigate any potential effects of fatigue.

To calculate the percentage of NMP allocated to support body weight, we calculated the slope of the regression line for the NMP of walking at 1.0 BW for each subject, to the NMP of walking at 0.5 BW of each subject. We used the y-intercept of the line (0G) and subtracted this from the NMP of unaltered walking at 1.0 BW with 1.0 M (UW). Then we divided 0G by the

NMP of UW to determine the percentage of NMP that is required to support body weight (Eq. 1). Finally, we averaged the percentage of NMP due to body weight support across all subjects.

$$\frac{UW - 0G}{UW}\ x\ 100 = \%\ of\ NMP\ to\ support\ BW \tag{1}$$

We determined the percentage of NMP due to mass redirection and acceleration by calculating the ratio of the added mass (AM) alone to the added weight (AW) condition, multiplied by 100, and averaged the ratio for 1.25 and 1.5 added conditions at each slope for each subject (Eq. 2). Then, we averaged the percentage of NMP due to mass redirection and acceleration across all subjects.

$$\frac{AM}{AW} x\ 100 = \%\ of\ NMP\ for\ mass\ redirection \tag{2}$$

Statistics We compared the percentage of NMP due to weight and mass for each slope to level ground using paired t-tests (RStudio, Boston, MA) and a custom statistical code (32). We adjusted for multiple comparisons at each slope with Bonferroni corrections (P<0.025). We used three least squares regressions to calculate the percentage of NMP due to weight and mass for each subject at each slope: 1) percentage decrease in NMP due to simulated reduced gravity, 2) percentage increase in NMP due to added weight and mass 3) and percentage increase in NMP due to added mass.

III. Results

Average total NMP increased with steeper uphill slopes and decreased with steeper downhill slopes during UW (Table 1). We averaged vertical ground reaction forces across 10 strides and determined that BW was reduced by 0.2899 BW ± 0.0103 and 0.5373BW ± 0.0196 (Avg ± SD) but all comparisons assumed 0.25 BW and 0.5 BW. We applied a constant upward force to each subject (force varied less than ±0.04 g across a stride, g=9.81) due to the large stretch of the rubber tubing.

Reducing BW by a greater percentage using simulated reduced gravity indicated there was no change NMP compared to the unaltered walking on level ground (Figure 2; Table 1). When we added weight, NMP was greater than UW at 1.25 BW ($p<0.0001$) and 1.5 BW ($p<0.0001$) (Figure 2; Table 1). When we added mass alone, NMP increased compared to UW at 1.25 M ($p>0.001$) and 1.5M ($p>0.0001$) (Figure 2; Table 1). The least squares regressions describing added weight and added mass were best fit by second-degree polynomials (Table 2), while the regressions describing simulated reduced gravity were best fit by linear relationship (Table 2).

Percentage of NMP

The average percentage of NMP due to body weight support increased for uphill slopes compared to level ground (Table 3; Figure 2, 3). The NMP attributed to body weight support increased from 14.5 ± 15.6% (Avg ± SE) on level ground to 63.0 ± 19.7% ($p=0.001$) at +3º, and 75.7 ± 24.1% ($p=0.0001$) at +6º (Figure 3). For the same slopes, we found that the average NMP due to mass redirection and acceleration decreased from 36.1 ± 9.3% on level ground to 19.6 ± 8.5% ($p=0.011$) at +3º, and 23.4 ± 12.2% ($p=0.015$) at +6º (Figure 3). The percentage of NMP allocated to BW support or mass redirection for downhill walking was not different than level-ground walking ($p>0.025$).

Table 1. Average ± SD net metabolic power (NMP) for all slopes and conditions.

		Average NMP (W/kg)						
Condition		**0.50BW 1.0M**	**0.75BW 1.0M**	**1.0BW 1.0M**	**1.0BW 1.25M**	**1.25BW 1.25M**	**1.0BW 1.5M**	**1.5BW 1.5M**
Slope	**+6°**	3.97±0.68	5.29±0.98	6.38±1.43	6.89±1.43	8.58±1.86	7.65±1.57	11.78±2.65
	+3°	2.85±0.27	3.50±0.28	4.16±0.26	4.43±0.32	5.55±0.41	4.94±0.57	8.04±0.88
	0°	2.34±0.22	2.48±0.18	2.50±0.26	2.96±0.41	3.58±0.38	3.23±0.31	4.99±0.62
	-3°	1.61±0.27	1.60±0.32	1.70±0.29	1.96±0.29	2.33±0.34	2.31±0.28	3.71±0.55
	-6°	1.27±0.32	1.20±0.34	1.37±0.38	1.50±0.39	2.07±0.50	1.92±0.54	3.25±0.73

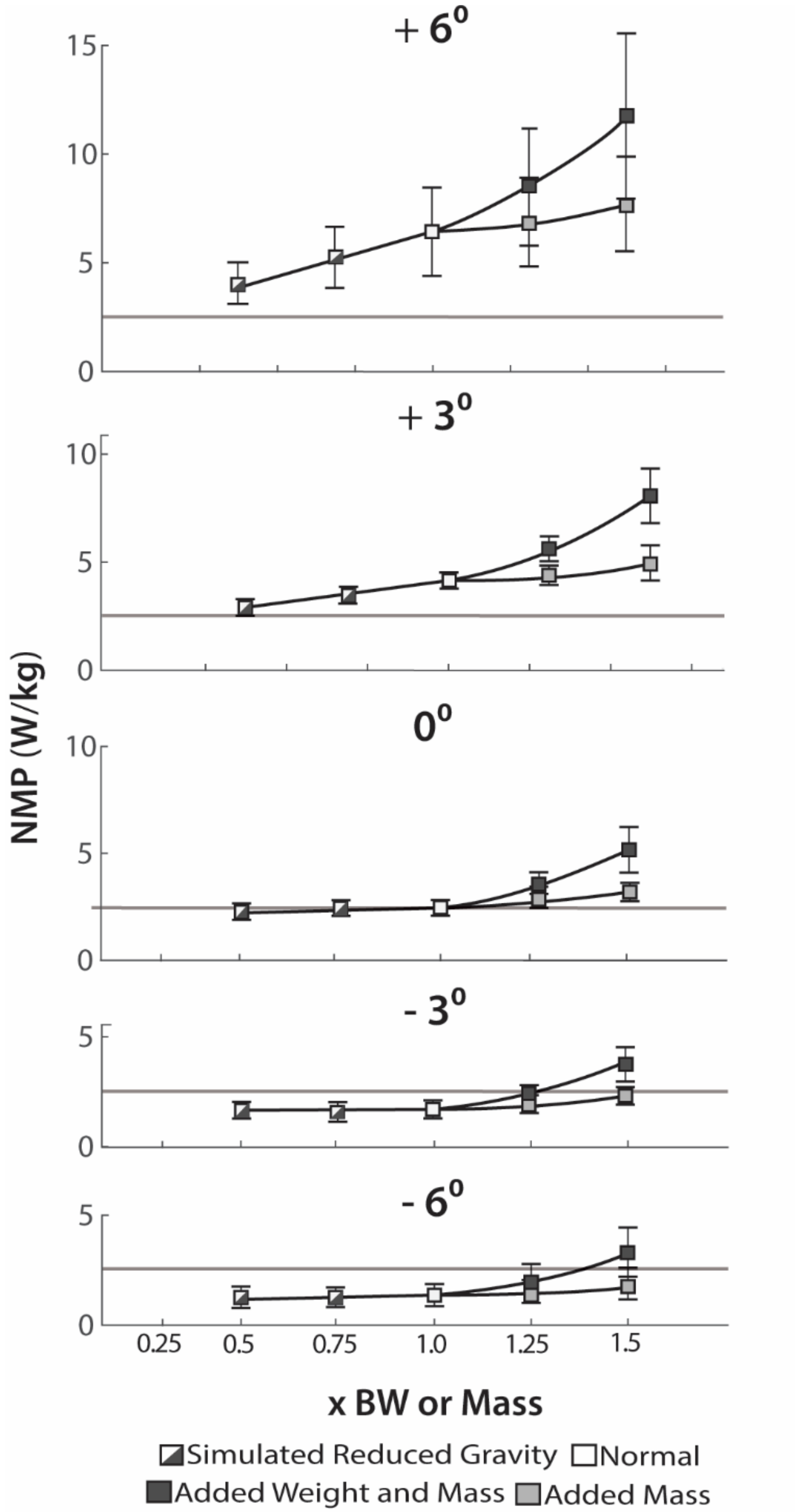
+ 6°
15
10
5
0
+ 3°
10
5
0
0°
10
5
0
- 3°
5
0
- 6°
5
0
0.25
0.5
0.75
1.0
1.25
1.5
NMP (W/kg)
x BW or Mass
Simulated Reduced Gravity
Normal
Added Weight and Mass
Added Mass

Figure 2. Average ± SD Net Metabolic Power (NMP) compared to fraction of body weight (BW or Mass for unaltered walking (open square), walking with simulated reduced gravity (half solid), walking with added mass (light gray), and walking with added weight (dark grey) across all slopes. Trend lines for reduced gravity walking are linear least squares regressions, and the added mass and added weight lines are second-order polynomial least squares regressions (R-squared values are listed in Table 2).

Table 2. R^2 Values of the NMP least squares regression lines organized by condition at each slope.

Condition	***Slope***	R^2
Added weight	***-6⁰***	0.6334
	-3⁰	0.8053
	0⁰	0.8275
	+3⁰	0.8842
	+6⁰	0.5397
Added mass	***-6⁰***	0.2637
	-3⁰	0.4493
	0⁰	0.4875
	+3⁰	0.4089
	+6⁰	0.1394
Simulated Reduced Gravity	***-6⁰***	0.3661
	-3⁰	0.5903
	0⁰	0.8635
	+3⁰	0.9999
	+6⁰	0.997

Table 3. Average percentage of net metabolic power (NMP) ± SE for BW support and mass redirection and body weight support at each slope. * indicates $p < 0.025$ compared to level ground walking.

	NMP Allocation (%)	
Slope	***BW support***	***Mass Redirection***
+6°	70.7 ± 7.6%*	23.1 ± 3.8%*
+3°	63.1 ± 2.43%*	18.9 ± 2.7% *
0°	14.5 ± 4.9%	34.7 ± 3.0%
-3°	8.4 ± 6.3%	38.3 ± 2.7%
-6°	7.3 ± 16.1%	22.5 ± 6.1%

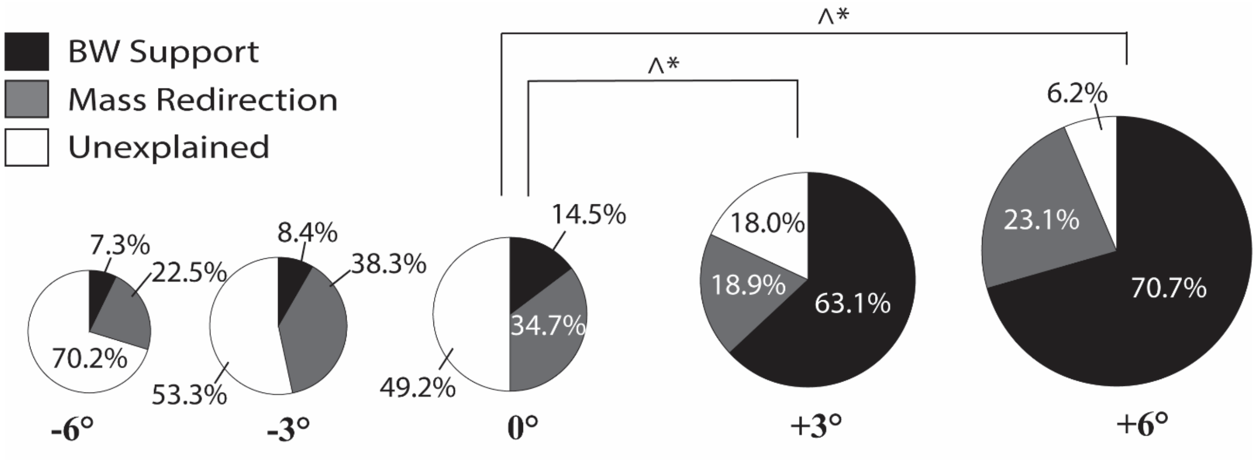

Figure 3. Pie charts of the average percentage of Net Metabolic Power (NMP) due to BW support (black), mass redirection (gray), and unexplained costs (white) across slopes. The total area of each pie chart symbolizes the total NMP for each slope. *indicates a significant difference from level ground ($p < 0.025$) for mass redirection costs. ^ indicates a significant difference from level ground ($p < 0.025$) for BW support costs.

Discussion

We determined the percentage of NMP allocated to supporting body weight and redirecting and accelerating the center of mass by comparing NMP while subjects walked with simulated reduced gravity, added mass, and added weight on uphill and downhill slopes. In partial support of our hypothesis that supporting body weight would incur a greater percentage of NMP during uphill and a lesser percentage of NMP during downhill compared to level-ground walking, we found that the percentage of NMP for body weight support increased for uphill slopes, but was the same on downhill slopes compared to level ground (Figure 3). The changes in the percentage of NMP allocated to body weight support across slopes may be due to the differences in muscle forces at different slopes (2). During uphill walking, the knee exhibits greater flexion during the stance phase compared to level ground walking (11), and requires greater knee torque. This decreases the effective mechanical advantage of the stance limb extensor muscles (4) and requires greater activation from extensor muscles at the hip (biceps femoris and gluteus maximus), knee (rectus femoris and vastus medialis), and ankle (medial gastrocnemius and soleus) (12) to raise the center of mass. Greater concentric activation of the extensor muscles implies a higher metabolic cost of walking to overcome gravity on uphill slopes compared to level ground. This infers that muscular force generation to support body weight and overcome gravity is the underlying reason for the greater percentage of NMP and greater absolute NMP with increasing slopes.

We reject our second hypothesis that redirecting and accelerating the center of mass would require a greater percentage of NMP for uphill, and a lower percentage of NMP for downhill compared to level ground walking. The percentage of NMP due to mass redirection at +3° and +6° was lower and at -3° and -6° was not different than that of level ground (Figure 3). Redirecting the center of mass requires mechanical work done by both the leading and trailing legs (7, 8). On uphill slopes, step length increases compared to level ground, which presumably increases step-to-step transition work (7). Greater mechanical work during the step-to-step transition (7) requires greater concentric and eccentric muscle action and thus would require a greater absolute NMP (1), but this does not explain the decrease in the percentage of NMP allocated to mass redirection on uphill slopes. Even though mass redirection required a lower percentage of the overall NMP for uphill slopes compared to level ground, the absolute NMP required for mass redirection was 0.55 W/kg greater at +6°. Thus, greater absolute mechanical work is needed during the step-to-step transition phase with increasing slopes even though the percentage of NMP is lower (Table 3).

Our results are similar to those of previous studies (10, 17). Grabowski et al. found that ~28% of the NMP was due to BW support and ~45% of the NMP was due to mass redirection for level ground walking at 1.25 m/s, whereas we found that the percentage of NMP due to BW support was 14.5 ± 15.6 % (Figure 3) and due to mass redirection was 34.7 ± 9.3 % (Figure 3). We calculated the percentages based on each subject and then averaged these percentages; whereas Grabowski et al. averaged all values of NMP and then calculated the percentages, which may account for ~5% difference in calculations. We did not ask subjects to walk at 0.25 of normal BW due to time constraints of the study. If we include the NMP for 0.25 BW from Grabowski et al., the percentage of NMP due to BW support would be ~19%. Additionally, the percentage of NMP due to BW support in Grabowski et al. (~28%) was within 1 SE of our calculated percentage of NMP due to BW support (14.5 ± 15.6%). The large variability implies that subjects may have used different walking strategies when BW was reduced.

With each successively steeper uphill slope, the combined percentage of NMP due to body weight support and mass redirection accounted for a larger portion of the NMP (Figure 3).

The remaining metabolic power can likely be attributed to costs of balance/stabilization and limb swing during walking (9, 14). For these parameters, the percentage of NMP that is unexplained is not different for level ground, -3°, and -6° ($p>0.05$). The percentage of NMP that is unexplained is significantly lower for +3° and +6° compared to level ground ($p<0.001$). These data indicate that during uphill walking, a smaller percentage of NMP and a larger absolute NMP (Table 4) is allocated to balance/stability and limb swing, which may be due in part to increased stride length and thus increased braking forces (7) during uphill compared to level ground walking. During downhill walking the absolute NMP is not different than level ground walking, indicating that the costs of stability and balance may also be the same (29).

Our results can be utilized to inform the development and design of assistive devices such as lower extremity powered prostheses, which could normalize the metabolic cost of walking for people with leg amputations when walking on slopes. During level ground walking, people with a transtibial amputation using a passive prosthesis have a 10-30% higher NMP compared to non-amputees (22, 33, 34) and use of a powered ankle-foot prosthesis that provides power at the prosthetic ankle normalizes NMP compared to non-amputees (20). When walking on slopes the percentage of NMP allocated to mass redirection decreases compared to level ground and thus the mechanical power provided will change with slope. The increased percentage of NMP for BW support during uphill compared to level ground walking implies that more force is needed to overcome gravity and thus biomimetic prostheses should be designed to provide greater force, presumably at the prosthetic ankle (23) during uphill walking to reduce the metabolic cost.

Table 4. Absolute net metabolic power (NMP) allocated to BW support, mass redirection and unexplained expenditure. We calculated the unexplained NMP by subtracting the percentage of NMP allocated to BW support and mass redirection from 100%. We calculated absolute NMP by multiplying the respective percentage of NMP by the absolute NMP for each condition and slope.

	NMP (W/kg)		
Slope	***NMP BW Support***	***NMP Mass Redirection***	***NMP Unexplained***
-6°	4.51	1.47	0.40
-3°	2.62	0.79	0.75
0°	0.36	0.87	1.23
+3°	0.14	0.65	0.91
+6°	0.10	0.31	0.96

The decreased percentage of NMP for mass redirection on uphill slopes implies that a lower percentage of mechanical work is needed for the step-to-step transition phase on uphill slopes, but greater absolute mechanical power is likely needed during the step-to-step transition compared to level ground. The percentage of NMP for BW support during downhill walking was similar to uphill walking, but downhill walking had greater variability in NMP (Table 3). This may indicate that there are multiple mechanical strategies used for downhill walking and thus individually tuning powered prostheses and orthoses for downhill walking might be more important than for uphill walking.

The use of simulated reduced gravity on a slope produces a small parallel assistive or resistive force to the subject. The vertical force applied to the subject can be broken down into two different component vectors, one that acts parallel to the walking surface and one that acts perpendicular to it (Figure 4). We calculated the magnitude of the vector that pulls parallel to the slope because this presumably affects mass redirection and NMP. The assistive or impeding parallel force is at most ~5% of body weight (Figure 4). Previous studies have shown that on level ground an aiding force of 5% BW decreases NMP by approximately 1 W/kg and an impeding force of 5% BW increases NMP by approximately 1.7 W/kg (15) (Table 5). Thus, on uphill slopes we applied an aiding force that may account for a 0.65W/kg decrease in NMP and on downhill slopes we applied a braking force that may account for a 0.35 W/kg increase in NMP (Table 5) (Extrapolated from Table 1 Gottschall Kram 2003 (15)).

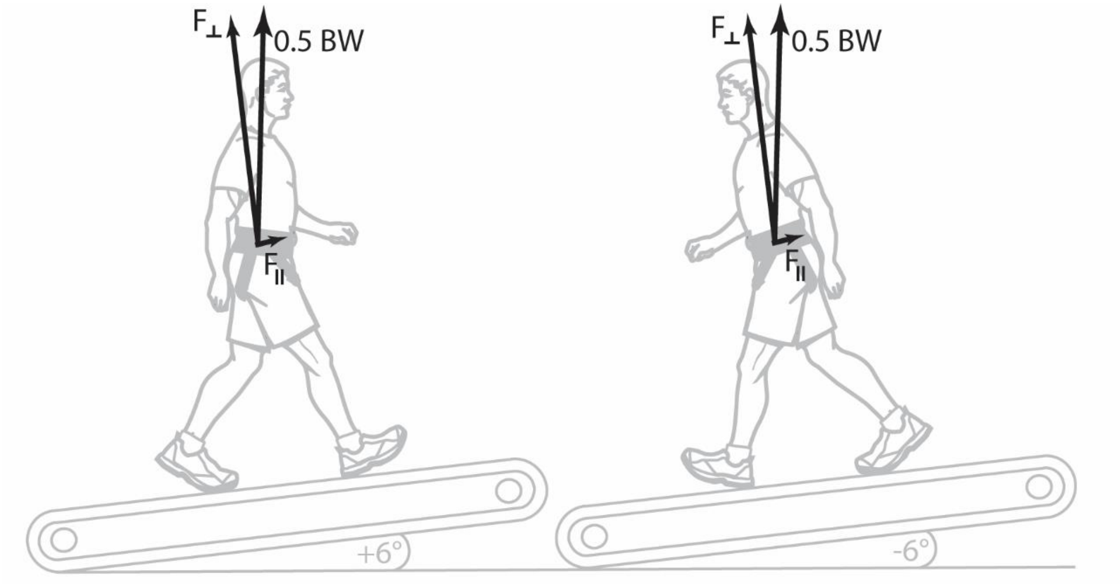

Figure 4. Simulated reduced gravity provides a force parallel ($F_{\parallel}$) and perpendicular($F_{\perp}$) to the slope, and acts as an assistive force on uphill slopes and an impeding force on downhill slopes. The parallel component of the 0.5 BW force vector is 0.5*BW*(sin6°) ≈ 5%BW and the perpendicular component of the 0.5 BW force vector is 0.5*BW*(cos6°) ≈95%BW for both uphill and downhill walking.

Table 5. The percentage of net metabolic power (NMP) accounted for by the parallel portion of the vector for uphill and downhill walking using the simulated reduced gravity apparatus.

Slope	*Vertical Force (BW)*	*Parallel Force (BW)*	*W/kg*
-6°	50%	-5.22%	0.35
	25%	-2.61%	0.20
-3°	50%	-2.62%	0.20
	25%	-1.31%	0.10
+3°	50%	1.31%	-0.15
	25%	2.62%	-0.30
+6°	50%	2.61%	-0.30
	25%	5.22%	-0.65

An aiding force during uphill walking would act to decrease the percentage of NMP allocated to mass redirection because it reduces the NMP needed for positive mechanical power done by the leading and trailing legs. An impeding force during downhill walking would decrease the percentage of NMP allocated to mass redirection because it reduces braking forces in the leading and trailing legs, which would presumably decrease NMP.

We determined the percentage of NMP required for body weight support and center of mass work during walking on slopes of -6° to +6°; however more extreme slopes would further elucidate how these biomechanical tasks affect metabolic costs. Thus, future studies should include a larger range of slopes to determine how these percentage allocations change. Additionally, we only applied simulated reduced gravity to the center of mass and not the swinging limbs and the moment of inertia of the limb might influence NMP. Additionally, future studies that reduce body weight to 0.25 BW are needed to quantify the percentage of NMP attributed to body weight support.

We found that the percentage of NMP allocated to body weight support and center of mass redirection and acceleration during walking changes with uphill and downhill slopes of -6° to +6°. During walking the percentage of NMP due to BW support increases with uphill slopes and the percentage of NMP due to mass redirection generally decreases with uphill slopes compared to level ground. Our results will inform the design and development of biomimetic assistive devices such as lower extremity powered prostheses and orthoses.

Conflict of Interest None of the authors have financial or personal conflicts of interest regarding this study.

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