Energetics and Biomechanics of Inclined Treadmill Walking in Obese Adults

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ABSTRACT

EHLEN, K. A., R. F. REISER II, and R. C. BROWNING. Energetics and Biomechanics of Inclined Treadmill Walking in Obese Adults. *Med. Sci. Sports Exerc.*, Vol. 43, No. 7, pp. 1251–1259, 2011. Brisk walking is a recommended form of exercise for obese individuals. However, lower-extremity joint loads and the associated risk of musculoskeletal injury or pathological disease increase with walking speed. Walking uphill at a slower speed is an alternative form of moderate intensity exercise that may reduce joint loading. **Purpose**: The purpose of this study was to quantify the energetics and biomechanics of level and uphill walking in obese adults. We hypothesized that compared to brisk level walking, walking slower up a moderate incline would reduce lower-extremity net muscle moments while providing appropriate cardiovascular stimulus. **Methods**: Twelve obese adult volunteers, with mass of 100.5 ± 15.7 kg and body mass index of 33.4 ± 2.6 kg·m^{-2} (mean ± SD), participated in this study. We measured oxygen consumption, ground reaction forces, and three-dimensional lower-extremity kinematics while subjects walked on a dual-belt force-measuring treadmill at several speed (0.50–1.75 m·s^{-1}) and grade (0°–9°) combinations. We calculated metabolic rate, loading rates, and net muscle moments at the hip, knee, and ankle for each condition. **Results**: Metabolic rates were similar across trials and were of moderate intensity (48.5%–59.8% of $\dot{V}O_{2max}$). Walking slower uphill significantly reduced loading rates and lower-extremity net muscle moments compared with faster level walking. Peak knee extension and adduction moments were reduced by ~19% and 26%, respectively, when subjects walked up a 6° incline at 0.75 m·s^{-1} versus level walking at 1.50 m·s^{-1}. **Conclusions**: These results suggest that walking at a relatively slow speed up a moderate incline is a potential exercise strategy that may reduce the risk of musculoskeletal injury/pathological disease while providing proper cardiovascular stimulus in obese adults.
Key Words: PHYSICAL ACTIVITY, ENERGY EXPENDITURE, EXERCISE INTERVENTION, MECHANICS, JOINT LOADS

The prevalence of obesity in America continues to exceed 30% across most age groups (10). Obesity is associated with many diseases including heart disease, certain cancers, and diabetes and is the main preventable risk factor for large-joint osteoarthritis (OA) (9,24,32). Obesity is due, in part, to an imbalance in energy intake and expenditure, and both diet and physical activity are considered essential tools in weight management (11).

Obese individuals are advised to participate in moderate-intensity physical activity (40%–60% of $\dot{V}O_{2max}$) for a minimum of 30 min at least five times a week (13). Walking is often the recommended form of physical activity for obese persons because it is convenient and there is a low injury rate among lean individuals (15). However, when walking at their preferred speed (~1.2 m·s^{-1}), the intensity may not meet the moderate threshold (3,14). As a result, moderately obese individuals must walk faster (i.e., "brisk" pace, ≥1.5 m·s^{-1}) to meet physical activity guidelines, achieve the physiological benefits of exercise, and increase energy expenditure.

A strong positive relationship exists between level walking speed and lower-extremity joint loading (estimated via net muscle moments (NMM), joint reaction forces, and joint loading rates) (23,39). During level walking, the magnitude and mediolateral distribution of knee joint loading can be estimated via internal extension and abduction (external adduction) NMM, respectively. According to Lelas et al. (23), peak sagittal-plane knee extensor NMM during early stance increase nearly 40% as walking speed increases from 1.2 to 1.5 m·s^{-1}. Knee abduction moments also increase with walking speed, although the increase is more modest (20). In addition, recent research suggests that absolute NMM (N·m) may be greater in moderately obese versus nonobese persons (4,19). Therefore, the combination of brisk level walking and obesity is likely to result in relatively large loads across the lower-extremity joints, the medial compartment of knee in particular. Faster walking speeds also increase the rate of loading during early stance because peak vertical ground reaction force (GRF) increases while early stance duration decreases. These greater and more

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rapidly applied loads may increase the risk of acute musculoskeletal injury or the development of OA (28,32). In fact, ~25% of habitual walkers who are overweight experience a significant musculoskeletal injury in any 12-month period, and one-quarter of these injured women and one-third of injured men permanently abandon their exercise program (15). Walking slower would presumably reduce these risks but would also reduce the cardiovascular benefits because of the relatively low aerobic demand.

A potential strategy to maintain adequate exercise intensity would be to have obese individuals walk uphill at a relatively slow speed. Walking up a moderate incline (<10°) increases metabolic rate compared to walking on a level surface at the same speed (1). Equations of the American College of Sports Medicine (ACSM) are commonly used to estimate oxygen consumption and can therefore be used to predict speed–grade combinations that elicit similar metabolic responses (33). Using standard prediction equations, walking at 0.75 $m \cdot s^{-1}$ up a 6° incline would require approximately the same oxygen consumption as level walking at 2.1 $m \cdot s^{-1}$ (33).

Walking on an incline at a relatively slow speed is not a common exercise recommendation for obese adults and this may be due to the lack of research on the effects of slow walking speeds and inclines on the biomechanics of walking in this population. In fact, there has only been limited research on the biomechanics of incline walking in nonobese adults, and no studies have examined the combined effects of speed and grade. Both McIntosh et al. (25) and Lay et al. (21) found that, in nonobese persons, lower extremity hip extensor and ankle plantarflexor NMM increased ~75% and ~20%, respectively, with incline. However, Lay et al. (21) reported that peak knee extensor NMM during early stance were not significantly different when subjects walked up an 8.5° ramp compared with level walking at ~1.25 $m \cdot s^{-1}$, whereas McIntosh et al. (25) reported that peak knee extensor NMM increased ~50% during incline (8°) versus level walking at ~1.65 $m \cdot s^{-1}$. Thus, walking uphill compared with level walking at typical speeds (~1.25–1.65 $m \cdot s^{-1}$) results in much smaller increases in peak hip and knee extensor NMM than would be observed when level walking speed is increased from 0.75 to 1.75 $m \cdot s^{-1}$ (~3.5-fold increase at both joints [23]). In addition, loading rates (slope of vertical GRF during early stance) are reduced at slower speeds because of the smaller peak vertical GRF and longer early stance times (4). The combination of adequate cardiovascular stimulus and reduced hip and knee NMM (extension and abduction) and lower loading rates could suggest that slow incline walking on a treadmill is a relatively safe and effective form of exercise for obese adults.

The purpose of this study was to quantify the energetics and biomechanics of uphill versus level walking in moderately obese adults. We hypothesized that slower walking up moderate inclines compared with faster level walking would 1) provide similar physiologic stimulus and 2) reduce lower extremity NMM and peak vertical loading rates for moderately obese persons.

METHODS

Subjects

A total of 30 individuals were recruited/screened and 12 obese adult volunteers (7 females and 5 males) met the inclusion criteria and participated in this experiment. Subjects were in good health (no known acute/chronic disease or physical activity limitations), sedentary to lightly active (<3 h of physical activity per week), not taking any medications known to alter metabolism, and with stable body mass (<2.5-kg net change during the previous 3 months). Physical characteristics of the subjects are shown in Table 1. Subjects gave written informed consent that followed the guidelines and was approved by the Colorado State University's Human Research Institutional Review Board.

Experimental Protocol

Each subject completed three experimental sessions. During the first session, which followed a 12-h fast, subjects underwent a physical examination and body composition was measured. We also recorded anthropometric characteristics required to determine lower-extremity body segment parameters (38). Finally, subjects completed a standard graded exercise stress test to determine maximal oxygen uptake ($\dot{V}O_{2max}$). During the second and third sessions, which each followed a 4-h fast, we collected metabolic and biomechanics data as subjects stood and walked (with shoes) at 16 speed–grade combinations (eight per visit). Treadmill speeds ranged from 0.50 to 1.75 $m \cdot s^{-1}$ in increments of 0.25 $m \cdot s^{-1}$ (six total) and grades were −3°, 0°, 3°, 6°, and 9°. Because the purpose of this study was to compare speed–grade combinations that met the criteria of a moderate-intensity exercise (40%–60% $\dot{V}O_{2max}$) on the basis of ACSM prediction equations and an estimated $\dot{V}O_{2max}$ of ~30 $mL \cdot kg^{-1} \cdot min^{-1}$ (3), we analyzed and report results from five of the speed–grade trials. The trials included 0.50 (9°), 0.75 (6°), 1.25 (3°), 1.50 (0°), and 1.75 $m \cdot s^{-1}$ (0°). Trials were 6 min in duration, and subjects were allowed a 5-min rest between trials. We familiarized the subjects to the treadmill at the beginning of the second session (after collecting standing metabolic data) by having them walk (0° grade) at a self-selected speed for ~10 min.

TABLE 1. Physical characteristics of participants.

Subject Characteristics	
Age (yr)	27 ± 5.5
Height (m)	1.73 ± 0.13
Body mass (kg)	100.5 ± 15.7
Body mass index ($kg \cdot m^{-2}$)	33.4 ± 2.4
% body fat	38.0 ± 7.5
Lean body mass (kg)	63.2 ± 16.7
$\dot{V}O_{2max}$ ($mL \cdot kg^{-1} \cdot min^{-1}$)	29.6 ± 5.4
Standing metabolic rate ($W \cdot kg^{-1}$)	1.21 ± 0.13

Values are mean ± SD.

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Assessments

Physical health and activity. During the first visit, each subject completed a health history form and was interviewed and assessed by a physician. Blood was drawn to test for normal metabolic function. Resting levels of thyroid-stimulating hormone and blood cell count were measured and confirmed to be within reference ranges. Physical activity levels were assessed via a questionnaire, and only individuals with <3 h of moderate–vigorous physical activity a week were invited to participate.

Body composition. We measured each subject's body composition using dual-energy x-ray absorptiometry (DEXA; Hologic Discovery, Bedford, MA). We determined percent body fat and percent lean mass for the entire body and three regions of interest: thigh, shank, and foot. Regions of interest were manually identified using the DEXA software. The thigh segment proximal end was defined as a line between the superior border of the iliac crest and the inferior border of the coccyx, excluding the pelvis. The thigh segment distal end and shank proximal end was a line between the femoral condyles and the tibial plateau. The shank segment distal end was a line between the inferior aspects of the medial and lateral malleolus. The foot segment was the remainder of the leg below the distal end point of the shank.

Maximal oxygen uptake. We used a modified Balke treadmill protocol to determine each subject's $\dot{V}O_{2max}$. Subjects were familiarized with the treadmill and the Borg RPE scale of 6–20. A 12-lead ECG was used to monitor heart function. Each subject's HR and blood pressure was measured in the supine, sitting, and standing positions to test for orthostatic intolerance. Subjects warmed up for ~5 min after which we slowly increased the speed of the treadmill until subjects reported an RPE indicative of moderate-intensity exercise (~11). Treadmill speed was then held constant, and the grade was increased by 1% every minute. The subjects were encouraged to continue to exhaustion. During the test, physiological responses to exercise were measured by recording HR, blood pressure, and RPE every 3 min. Heart function was monitored by a physician. We determined oxygen consumption via an open-circuit respirometry (Oxycon Mobile, Yorba Linda, CA), with expired gas data averaged every 30 s.

Energetic Measurements

Energetics and biomechanics measurements were collected during sessions 2 and 3 while subjects walked on a treadmill over a range of speeds and grades. To determine metabolic rate during standing and walking, we measured the rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) using a portable open-circuit respirometry system (Oxycon Mobile). Before the experimental trials, we calibrated the system and measured standing metabolic rate for 6 min. For each trial, we allowed 4 min for subjects to reach steady state (no significant increase in $\dot{V}O_2$ during the final 2 min and a RER <1.0) and calculated the average $\dot{V}O_2$ and $\dot{V}CO_2$ ($mL{\cdot}s^{-1}$) for the final 2 min of each trial. We calculated net metabolic rate ($W{\cdot}kg^{-1}$) from $\dot{V}O_2$ and $\dot{V}CO_2$ using a standard equation (2). We then subtracted standing metabolic rate from the walking values to derive the net metabolic rate.

Biomechanics Measurements

To record biomechanics data, we used a three-dimensional motion capture system (Motus 9.2; Vicon, Centennial, CO) and a dual-belt, inclinable, force-measuring treadmill (Fully Instrumented Treadmill; Bertec, Columbus, OH). We placed lightweight retroreflective spheres in accordance with the modified Helen Hayes marker set to identify anatomical landmarks and delineate lower extremity segments (17). Markers were placed on the sacrum (S1) and anterior superior iliac spine, midthigh (femoral wand), femoral epicondyle, midshank (tibial wand), lateral malleolus, second metatarsal head, and calcaneous of each leg. Marker trajectories were recorded at 60 Hz using eight optoelectronic cameras. GRFs and moments were recorded at 1200 Hz by force platforms embedded under each treadmill belt. Kinematic and kinetic systems were synchronized through the motion capture system. Data was collected for 30 s during the final minute of each trial.

Raw coordinate and kinetic data were smoothed using a fourth-order, zero-lag digital Butterworth filter with a cutoff frequency of 5 and 12 Hz, respectively. After exporting the data, we used vertical GRF data and a threshold of 15 N (on the basis of the SD of the vertical GRF signal during swing) (26) to determine heel strike and toe-off for each leg and computed temporal characteristics of each trial using custom software (Matlab, v12.0; Mathworks, Natick, MA). To determine the thigh and shank body segment parameters, we used the DEXA data to estimate thigh and shank mass and used the regression equations provided by Durkin and Dowling (8) to determine the radius of gyration (3). Segment mass and radius of gyration were used to calculate frontal plane moment of inertia. We used frontal plane values to represent sagittal-plane moment of inertia of the thigh and shank (6). Foot segment parameters were estimated using the Motus gait software (37). Three-dimensional lower extremity kinematic and kinetic variables (joint angles and NMM) were also computed using the Motus software, which uses the anthropometric data, estimated joint centers, segment velocities and accelerations, and the GRFs in a full inverse dynamics model (37). All variables were normalized to represent a percentage of stance. Positive and negative work at each joint were determined by numerically integrating the instantaneous positive and negative joint power (product of joint NMM and joint angular velocity) over the stance phase. Step width was determined as the distance between the midstance center of pressure of the right and left leg during consecutive steps. We also quantified the maximum vertical loading rate by determining the maximum slope of the normal GRF during the first 20% of the stance phase. We calculated

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the mean of each variable of interest of the right leg during 5–25 strides at each speed–grade combination for each subject and the mean across subjects for each speed–grade combination.

Statistical Analysis

A repeated-measures ANOVA was used to determine the effects of various speed–grade combinations on metabolic rate, temporal gait characteristics, midstance joint angles, peak NMM, and peak loading rate. If a significant main effect was observed, *post hoc* comparisons using the Holm–Sidak method were performed. When data failed the normality test, a multiple-comparison Tukey test on ranks was used. A criterion of $P < 0.05$ defined significance. SigmaPlot version 11.0 (Systat Software, Inc., San Jose, CA) was used to perform statistical analysis.

RESULTS

Energetics. Mean $\dot{V}O_{2max}$ was 9.9 $W{\cdot}kg^{-1}$ (29.6 $mL{\cdot}kg^{-1}{\cdot}min^{-1}$), and mean standing metabolic rate was 1.2 $W{\cdot}kg^{-1}$. Net metabolic rate was not different between uphill at 0.50 (9°) and 0.75 $m{\cdot}s^{-1}$ (6°) compared with level walking at 1.50 $m{\cdot}s^{-1}$ but was 29.8% and 14.4% greater during level walking at 1.75 and 1.25 $m{\cdot}s^{-1}$ (3°), respectively, compared with level walking at 1.50 $m{\cdot}s^{-1}$ (Table 2). All trials were of moderate intensity and required between 49% and 60% $\dot{V}O_{2max}$. The mean relative aerobic effort for 0.50 (9°), 0.75 (6°), 1.25 (3°), 1.50 (0°), and 1.75 $m{\cdot}s^{-1}$ (0°) was 48.3%, 52.1%, 58.0%, 51.1%, and 61.0% $\dot{V}O_{2max}$, respectively.

Kinematics. Temporal stride kinematics was significantly different across the speed–grade combinations (Table 2). As walking speed increased and grade decreased, stride frequency and stride length increased, whereas duty factor (percent of stride during stance) and double support time decreased ($P < 0.005$ for all dependent variables). Step width was significantly greater than level walking only at the slowest speed ($P < 0.001$).

Mean stance phase hip, knee, and ankle flexion/extension joint angles are shown in Figure 1. Knee and ankle joint angles differed across all speed–grade combinations but hip joint angles did not. Increasing the walking speed and decreasing the incline resulted in less flexion at the knee during early stance. Mean peak knee flexion in early stance was ~70% greater (39° vs 23°) at 0.50 (9°) versus 1.75 $m{\cdot}s^{-1}$ (0°), respectively. A steeper incline resulted in a more dorsiflexed ankle throughout the gait cycle.

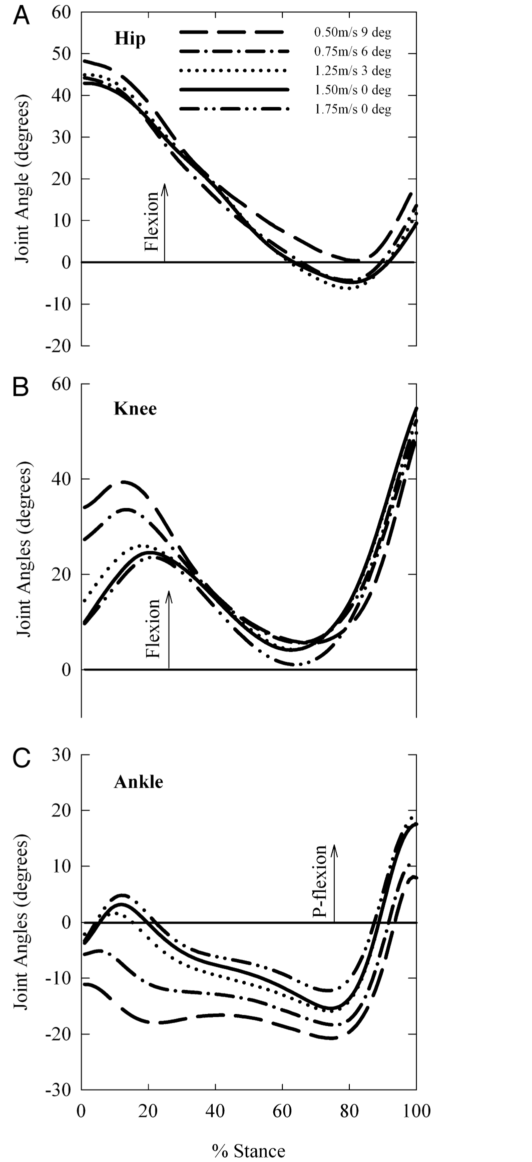

FIGURE 1—Mean hip (A), knee (B), and ankle (C) joint angles for each speed–grade. Percent stride is from right heel strike to right heel strike. Joint angles were similar across all trials in the hip. In early stance, flexion and dorsiflexion increased with grade in the knee and ankle, respectively.

Kinetics. Mean GRFs are shown in Figure 2. During early and late stance, peak normal GRFs were greater in trials with faster speeds and lower grades (Fig. 2A). Peak

TABLE 2. Net metabolic rate and temporal stride kinematics.

Speed ($m{\cdot}s^{-1}$)	Grade (°)	Net metabolic rate ($W{\cdot}kg^{-1}$)	Stride Frequency (Hz)	Stride Length (m)	Stance (% Cycle)	Double Support (% Cycle)	Step Width (m)
0.50	9	3.67 ± 0.10	0.58[a] ± 0.02	0.87[a] ± 0.03	72.6[a] ± 0.6	44.0[a] ± 1.2	0.188[a] ± 0.014
0.75	6	3.88 ± 0.09	0.71[a] ± 0.02	1.08[a] ± 0.04	70.3[a] ± 0.5	40.0[a] ± 0.9	0.170 ± 0.01
1.25	3	4.42[a] ± 0.09	0.89[a] ± 0.02	1.42[a] ± 0.03	66.9[a] ± 0.7	34.7[a] ± 0.8	0.168[a] ± 0.011
1.50	0	3.80[a] ± 0.10	0.99[a] ± 0.03	1.54[a] ± 0.05	64.5[a] ± 0.5	30.2[a] ± 0.6	0.161[a] ± 0.011
1.75	0	5.02[a] ± 0.13	1.07[a] ± 0.02	1.64[a] ± 0.03	63.6[a] ± 0.4	27.4[a] ± 0.7	0.167[a] ± 0.010

Values are mean ± SE.
[a] Significant difference between condition and 1.50 $m{\cdot}s^{-1}$ condition.

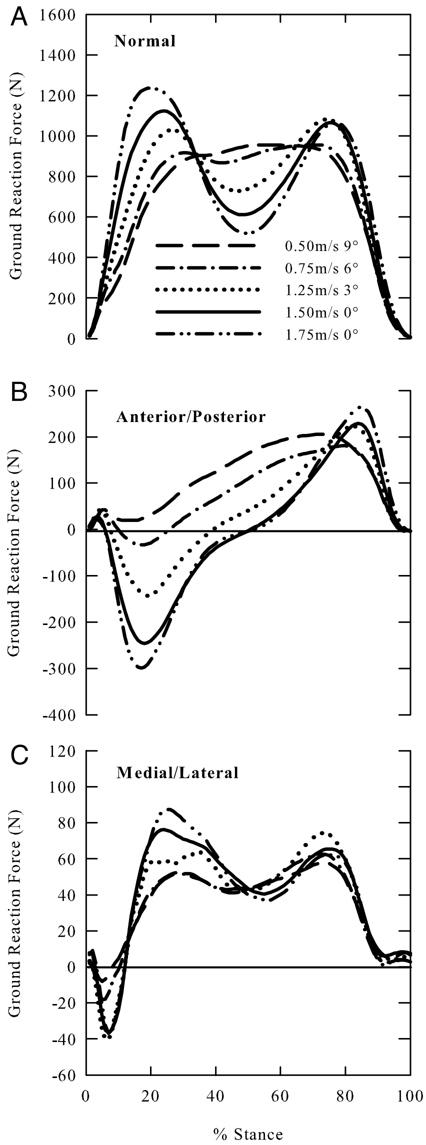

FIGURE 2—Mean normal (A), AP (B), and mediolateral (C) GRFs while walking at all speed–grade combinations. Faster walking speeds resulted in greater peak normal GRF during early and late stance and greater peak AP GRF in early stance. Mediolateral GRFs were slightly greater during faster walking speeds.

anterior–posterior (AP) GRFs in early stance were greater in the faster, level speeds, and there was no braking force at the steepest incline (Fig. 2B). Faster speeds and lower grades had slightly greater medial–lateral (ML) GRFs compared with slower speeds and moderate inclines (Fig. 2C). Maximum normal force loading rate increased as the walking speed increased. Maximum normal force loading rates were 3758, 5288, 8691, 12,997, and 19,237 N·s^{-1} when walking at 0.50 (9°), 0.75 (6°), 1.25 (3°), 1.50 (0°), and 1.75 m·s^{-1} (0°), respectively.

There was a significant main effect of speed–grade on peak NMM (flexion and extension) about the hip ($P < 0.001$), knee ($P = 0.002$), and ankle ($P < 0.001$). Peak NMM were smaller during the slower speed/moderate incline trials and increased as the speed increased and incline decreased (Table 3 and Fig. 3). Peak extensor NMM about the hip during early stance and the ankle during late stance were significantly lower when walking at 0.75 m·s^{-1} (6°) versus both 1.50 (0°) and 1.75 m·s^{-1} (0°). Peak extensor NMM about the knee during early stance were significantly lower when walking at 0.75 m·s^{-1} (6°) versus both 1.50 (0°) and 1.75 m·s^{-1} (0°) ($P = 0.002$). There was a 19% and 28% reduction in peak knee extensor NMM between 0.75 m·s^{-1} (6°) compared with 1.50 (0°) and 1.75 m·s^{-1} (0°), respectively (Fig. 3B). The peak knee abduction/adduction muscle moments increased with speed (Fig. 3D) ($P = 0.005$). The peak knee abduction moment was reduced ~54% and ~26% when walking at 0.50 (9°) and 0.75 m·s^{-1} (6°) versus 1.50 m·s^{-1} (0°), respectively.

Total positive joint work during stance (sum of hip, knee, and ankle positive work) was similar, but total negative joint work was smaller during slower incline walking compared with faster, level walking. Total positive joint work was 62, 78, 80, 71, and 85 J at 0.50 (9°), 0.75 (6°), 1.25 (3°), 1.50 (0°), and 1.75 m·s^{-1} (0°), respectively. Total negative joint work was −11, −20, −32, −45, and −47 J at 0.50 (9°), 0.75 (6°), 1.25 (3°), 1.50 (0°), and 1.75 m·s^{-1} (0°), respectively. As a result, total net joint work (sum of positive and negative work at hip, knee, and ankle) during stance was greater when walking uphill (58 J at 0.75 m·s^{-1} (6°)) versus faster, level walking (27 J at 1.50 m·s^{-1}). Hip and knee positive work was similar across the trials, and ankle positive work increased with faster/level speeds, whereas less negative work was done at all joints during slower speed/incline versus faster/level walking.

DISCUSSION

Energetics. We accept our hypothesis that slower walking up moderate inclines provides similar physiological stimulus and energy expenditure versus faster, level walking for moderately obese persons. Our results are in agreement with other studies that found that inclines increase the metabolic rate of walking (18,27), whereas slower speeds decrease

TABLE 3. Peak flexion/extension and abduction/adduction NMM.

Speed (m·s^{-1})	Grade (°)	Hip Extension (N·m)	Knee Extension (N·m)	Knee Abduction (N·m)	Ankle Plantarflexion (N·m)
0.50	9	44.8 ± 3.7[a]	52.8 ± 4.2	38.7 ± 5.3[a]	125.4 ± 15.1[a]
0.75	6	59.8 ± 3.5[a]	49.0 ± 5.8[a]	47.3 ± 3.9[a]	140.0 ± 11.8[a]
1.25	3	101.0 ± 8.0	55.2 ± 6.1	50.3 ± 6.1	175.8 ± 17.4
1.50	0	100.3 ± 8.4	58.1 ± 6.8	59.4 ± 7.2	172.6 ± 18.6
1.75	0	131.0 ± 6.9[a]	62.8 ± 7.4	63 ± 5.7	179.4 ± 15.0

Values are mean ± SE.
[a] Significant difference between condition and 1.50 m·s^{-1} condition.

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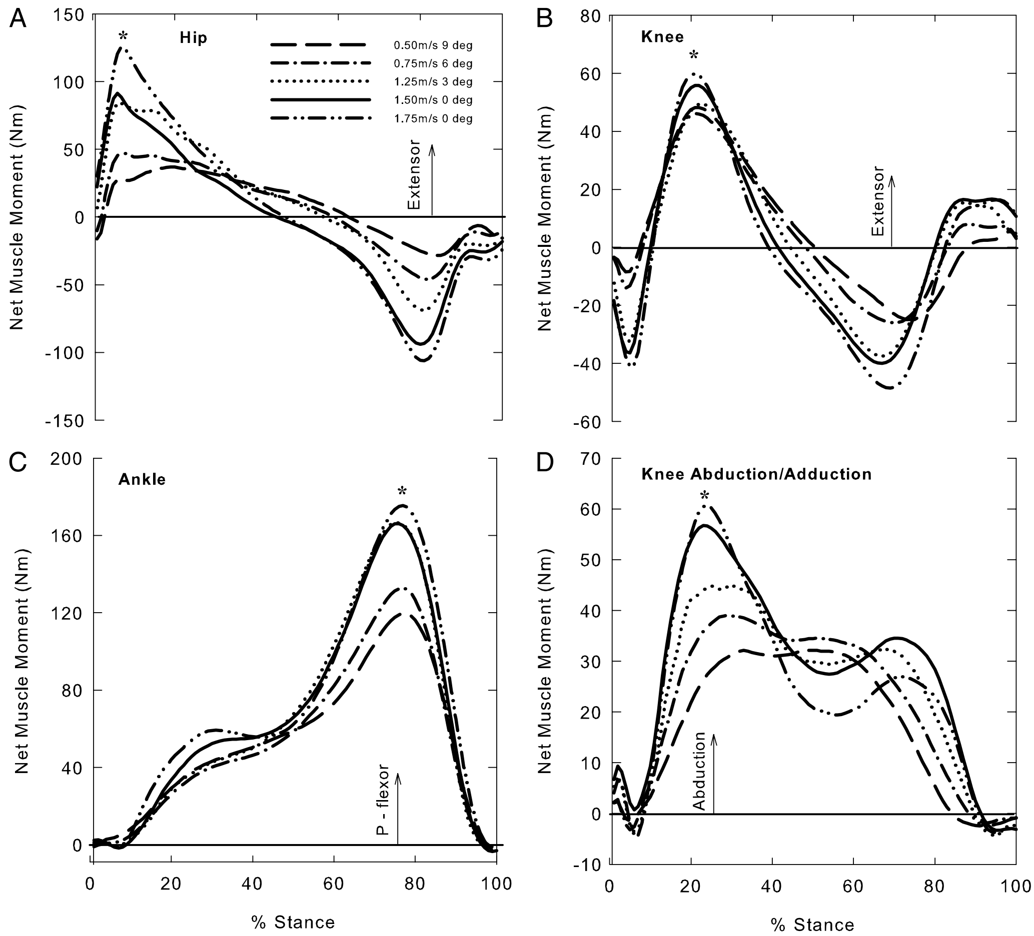

FIGURE 3—Mean hip (A), knee (B), and ankle (C) NMM during stance while walking at all speed–grade combinations. Positive moments are extensor. Stance begins at right heel strike. Peak NMM at the hip and ankle were greater at faster speeds. The peak extensor moment at the knee during early stance was significantly lower walking at 0.75 m·s^{-1} (6°) versus 1.75 (0°) and 1.50 m·s^{-1} (0°). Mean abduction/adduction moments at the knee (D) for all speed–grade combinations. Positive moments are abduction. Faster speeds resulted in greater peak adduction moments about the knee. *Significant main effect of speed–grade on peak NMM.

the metabolic rate (3,18). We used the ACSM metabolic rate prediction equation to aid in selecting the speed–grade combinations and found that this equation did not accurately predict metabolic rate in our moderately obese subjects. The prediction equation overestimated metabolic rate during slower/uphill walking by ~10% and underestimated metabolic rate during faster, level walking by up to 28% (1.75 m·s^{-1}). These results suggest that caution must be used when estimating intensity and associated energy expenditure via ACSM prediction equations in this population.

Our biomechanics results provide insights into the metabolic cost of walking. The primary determinant of the metabolic cost of walking is muscle work required to support and move the center of mass as well as swing the legs (29). Although we were not able to quantify muscle work, our joint moment data allowed the determination of joint work, which is related to metabolic cost (36). During incline walking, lower-extremity muscles must perform positive mechanical work to increase the gravitational potential energy with each step and more of this work is performed by proximal muscles/joints (hip and knee) compared with level walking (29,34). In our study, the combination of slower speeds/inclines versus faster level walking resulted in a similar metabolic rate across conditions, whereas summed (hip, knee, and ankle) net joint work (sum of positive and negative work) during stance was greater when walking uphill (58 J at 0.75 m·s^{-1} (6°)) versus faster, level walking (27 J at 1.50 m·s^{-1}). The increase in net joint work during uphill walking was due to a decrease in negative joint work (−20 J at 0.75 m·s^{-1} (6°) versus −45 J at 1.50 m·s^{-1}), suggesting that the ability to use stored elastic energy was reduced during the uphill walking trials. During level walking, considerable elastic energy is returned (~30% of total musculotendon mechanical work), primarily via the Achilles tendon (29). Elastic energy storage/return is smaller at slower speeds (30), and the contribution of this energy to the total work required is less when walking on an incline versus level walking at the same speed (34). While a reduction in utilization of stored elastic energy would presumably increase the metabolic rate during slow, uphill walking, the metabolic cost of swinging the legs may be reduced relative to that of faster, level walking. However, it is also possible that obese individuals have an altered ability to store/return elastic energy that would be affected by speed but not grade or *vice versa*. Muscle activity to control upper body posture may also play a role in the energetics/mechanics relationship. Individuals lean forward when walking uphill, and this may incur a metabolic cost without changing lower-extremity NMM. Future musculoskeletal modeling studies that are able to estimate the individual muscle contributions to the metabolic cost of walking in obese and nonobese will assist us in understanding the influence of excess body mass and whether obese individuals adopt a walking gait that conserves metabolic energy.

Kinematics. Temporal–spatial stride kinematics changed with speed–grade. The temporal stride characteristics we report are similar to those of previous studies with obese subjects (4,7). In our recent study of moderately obese subjects during level walking, stride length and frequency increased with walking speed, whereas duty factor decreased (4). A comparison of the results of this study with our earlier level walking results suggests that stride characteristics are dependent on speed and not grade, as has been reported for nonobese individuals (4). Thus, moderate obesity does not seem to elicit changes in the stride length–frequency relationship during uphill versus level walking. Obese adults spend more time in stance and double support than their nonobese counterparts (19). Although we did not have a nonobese control group, our level walking results support this finding because our values are similar to those of others (19).

The observed joint angle patterns are inconsistent with some of the previous studies reporting level walking kinematics of obese persons (4,19,35). DeVita and Hortobagyi (7) report that obese adults walk with a more erect posture (i.e., less knee flexion and greater plantarflexion during stance) compared with lean adults, although other studies have not supported this finding (19,35). We found that obese adults had $\sim$15° of early stance knee flexion during level walking at 1.50 $m{\cdot}s^{-1}$, which is similar to the range of motion reported for both nonobese and obese adults (4,21,35). The more extended posture adopted by the participants in the study of DeVita and Hortobagyi (7) may be attributed to fact that some of their participants were severely obese.

Our results show that uphill walking requires a greater degree of knee flexion during early stance and ankle dorsiflexion throughout stance compared with level walking. The combination of more flexed joints during early stance and greater extension range of motion is characteristic of uphill walking and is necessary to raise the body's center of mass and ensure foot clearance (25). To our knowledge, no study has analyzed the kinematics of uphill walking in obese persons. We observed similar hip joint angles across the speed–grade combinations. Lay et al. (21) and McIntosh et al. (25) reported that hip flexion increased $\sim$60% at heel strike between level and uphill ($\sim$9°) walking in nonobese individuals. This difference is most likely a result of walking speed. At the 9° incline, our subjects walked much slower than the participants in these studies (21,25). Slower walking speeds result in shorter stride lengths, which limit the range of motion of the hip joint. It is also possible that the obese subjects did not increase pelvic tilt during uphill walking, as has been observed in nonobese individuals (25). A smaller change in pelvic tilt would reduce the increase in hip flexion during uphill versus level walking and may reduce the muscle forces required to support the upper body. Future studies that quantify pelvic motion in obese adults during level and uphill walking are needed.

Kinetics. We accept our hypothesis that decreased walking speeds combined with moderate inclines reduces lower-extremity NMM compared with faster, level walking. As walking speed decreased and grade increased, hip, knee, and ankle NMM decreased. During level walking, we found hip, knee, and ankle NMM patterns to be consistent with previous studies of nonobese and obese adults (4,25,39). DeVita and Hortobagyi (7) reported a peak knee extensor NMM during early stance of 0.52 $N{\cdot}m{\cdot}kg^{-1}$ when obese subjects walked at 1.50 $m{\cdot}s^{-1}$, comparable to the 0.58 $N{\cdot}m{\cdot}kg^{-1}$ reported here. Lower-extremity joint NMM decrease with speed during level walking (5) and increase with incline when speed is held nearly constant (21,25). During uphill walking, the hip joint extensor NMM increases dramatically, whereas early stance knee extensor NMM and late stance ankle plantarflexor moments have a more modest increase (21,25). For example, when nonobese adults walked up an 8.5° grade compared with level walking at $\sim$1.2 $m{\cdot}s^{-1}$, peak hip and knee extensor and ankle plantarflexor NMM increased by 104%, 46%, and 18%, respectively (21). Comparing our results with those of Lay et al. suggests that speed may influence lower-extremity muscle NMM more than grade, at least for the speed–grade combinations tested.

Net muscle moments provide a proxy measure for *in vivo* loads across the joints of the lower extremity and have been used by many investigators to estimate these loads. A limitation of this approach is that the NMM does not reflect the role of groups/individual muscles (i.e., knee extensors) that cross a joint but is a resultant (net) moment at the joint based on all muscles that cross that joint. Thus, there may be many combinations of agonist–antagonist muscle force production that produce a similar NMM at a particular joint. When walking uphill versus level at a similar speed, there is an increase in cocontraction of the knee flexors and extensors (22). Lay et al. (22) report that, when walking up a 21° slope versus level walking, vastus medialis and biceps femoris muscle activity amplitude increased 451% and 347%, and activity duration increased from $\sim$20% to 40% of stride, respectively, for both muscles. Lay et al. (22) state that the much larger hip extension moment required to walk uphill requires increased activation of the biarticular hip extensor–knee flexor muscles, which in, turn require an increase in knee extensor muscle activity to maintain an extensor moment at the knee. Importantly, these increases in muscle activity did not result in a significant increase in knee flexion/extension NMM (although the peak was 85% greater) and

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suggests that the changes in NMM are not reflective of the changes in joint loading during incline walking.

Although we acknowledge the possibility that the decrease in knee NMM we report during incline versus level walking may not reflect a decrease in knee joint loading, our subjects were walking on a more gradual incline at much slower speeds than those reported by Lay et al. (21) and muscle activity decreases with speed (31). The finding that hip extension NMM was smaller during most incline versus level walking trials suggests that the biarticular hip extensor–knee flexor muscles may not be required to assist with hip extension, which would reduce the coactivation of knee flexion–extension muscles and reduce the need for increased knee extensor torque production. An additional benefit of the slower speed/incline is that rates of loading are much slower, given the smaller peak normal GRF and longer duration of stance. We found that loading rates were ~2.5× greater during level walking at 1.50 versus 0.75 $m \cdot s^{-1}$ (6°). Clearly, future studies that quantify muscle activity and use musculoskeletal models to estimate joint forces will lead to a better understanding of the relationship between NMM and joint loading during incline walking.

Slower walking up moderate inclines reduced the peak early stance knee abduction NMM compared with faster level walking. Greater abduction NMM at the knee are associated with increased medial compartment knee loading, varus malalignment, and a greater risk of OA progression (16). Our abduction–adduction NMM pattern is consistent with that reported in the study of Foroughi et al. (12). By decreasing the speed from 1.75 to 0.50 $m \cdot s^{-1}$ and walking up an incline of 9°, the peak knee abduction–adduction moment decreased by 63%. This finding suggests that the distribution of load on the medial compartment is reduced during slower, incline walking versus faster, level walking.

Relevance for exercise prescription. Our results suggest that slower, uphill walking may be an appropriate form of exercise for moderately obese adults, particularly those who do not have lower-extremity joint discomfort/pain. For instance, a 100-kg person walking on a 6° incline at 0.75 $m \cdot s^{-1}$ for 30 min will expend ~220 kcal, which is slightly more than what that person would expend walking at 1.50 $m \cdot s^{-1}$ on a level surface for the same amount of time. Importantly, walking at slower speeds also reduces the perceived exertion of the exercise (14), which may result in increased activity time and adherence even when walking uphill. Finally, it is important to note that this would require a treadmill-specific exercise prescription because walking uphill outdoors would require downhill walking as well, which is known to increase joint loading (21,25). Thus, our findings support an exercise prescription specific to treadmill and not overground walking.

CONCLUSIONS

We found that walking at slower speeds and moderate inclines resulted in smaller NMM across the lower-extremity joints in moderately obese adults. With the exception of the fastest walking trial, the energetics of slope walking was similar across most speed–grade combinations. Although the relatively small sample size may limit the generalizability of our findings to other populations (e.g., severely obese), the results suggest that walking at slower speeds (<0.75 $m \cdot s^{-1}$) and moderate inclines (6°–9°) may reduce load, load distribution, and loading rate across the lower-extremity joints while ensuring adequate aerobic stimulus for weight management.

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