# The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking

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## ABSTRACT

BROWNING, R. C., J. R. MODICA, R. KRAM, and A. GOSWAMI. The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking. *Med. Sci. Sports Exerc.*, Vol. 39, No. 3, pp. 515–525, 2007. **Purpose:** The metabolic cost of walking increases when mass is added to the legs, but the effects of load magnitude and location on the energetics and biomechanics of walking are unclear. We hypothesized that with leg loading 1) net metabolic rate would be related to the moment of inertia of the leg ($I_{leg}$), 2) kinematics would be conserved, except for heavy foot loads, and 3) swing-phase sagittal-plane net muscle moments and swing-phase leg-muscle electromyography (EMG) would increase. **Methods:** Five adult males walked on a force-measuring treadmill at 1.25 $m \cdot s^{-1}$ with no load and with loads of 2 and 4 kg per foot and shank, 4 and 8 kg per thigh, and 4, 8, and 16 kg on the waist. We recorded metabolic rate and sagittal-plane kinematics and net muscle moments about the hip, knee, and ankle during the single-stance and swing phases, and EMG of key leg muscles. **Results:** Net metabolic rate during walking increased with load mass and more distal location and was correlated with $I_{leg}$ ($r^2 = 0.43$). Thigh loading was relatively inexpensive, helping to explain why the metabolic rate during walking is not strongly affected by body mass distribution. Kinematics, single-stance and swing-phase muscle moments, and EMG were similar while walking with no load or with waist, thigh, or shank loads. The increase in net metabolic rate with foot loading was associated with greater EMG of muscles that initiate leg swing and greater swing-phase muscle moments. **Conclusions:** Distal leg loads increase the metabolic rate required for swinging the leg. The increase in metabolic rate with more proximal loads may be attributable to a combination of supporting (via hip abduction muscles) and propagating the swing leg. **Key Words:** LOCOMOTION, LEG LOADING, ELECTROMYOGRAPHY, METABOLIC RATE, LOAD CARRIAGE

R ecent studies suggest that the primary determinants of the net metabolic rate during walking are performing work to propel the center of mass forward, supporting body weight, and swinging the legs (10–12). Gottschall and Kram (10) estimate that leg swing can account for only about 10% of the net metabolic rate during human walking. Yet, when a modest mass is added to the shank or foot, the metabolic rate during walking increases dramatically (29). A biomechanical explanation for this increase in net metabolic rate during walking has not been clearly established. A better understanding of the relationship between lower-extremity loading and the energetics and biomechanics of walking has practical importance for the study of obesity as well as the design of lower-extremity body armor, prosthetic legs, and powered leg exoskeleton devices. For example, we may gain insight into the effects of increased leg mass, via obesity or body armor, on the net metabolic rate of walking. Also, designers of lower-extremity prosthetic and assistive devices (e.g., powered orthosis) may be able to better estimate how the mass of these devices will affect the energetics and biomechanics of walking. We may also improve our understanding of neuromuscular control strategies that are used during gait.

Walking with an external load generally increases metabolic rate. At normal walking speeds, adding a moderate load via a backpack or around the body's center of mass increases gross metabolic rate in direct proportion to the added mass (i.e., a 20% body weight load results in a 20% increase in metabolic rate) (13). When expressed as net metabolic rate (gross − standing) per kilogram of body mass, the increase is greater than proportional to the load. For example, when walking at 1.5 $m \cdot s^{-1}$, Griffin et al. (13) report an approximately 30% increase in net metabolic rate with a load of 20% of body mass, whereas Grabowski et al. (12) report a 98% increase in net metabolic rate with a load of 50% of body mass.

When mass is added to the extremities, metabolic rate increases disproportionately with load and is greater with more distal locations of the load (29). Rose et al. (26) have shown that adding just 2 kg to each foot increased the gross metabolic rate by 30%. Royer and Martin (28) report that

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net metabolic rate increased 8% when 2.8 kg was added to each proximal shank compared to walking with a similar load at the waist. Although the studies mentioned above have measured the effects of shank and foot loading on walking metabolic rate, the effects of thigh loading, large load mass, or load location (thigh, shank, or foot) are unknown. Thus, there is a need for a systematic, integrative (i.e., metabolic rate, kinematics, kinetics, and EMG) study using a range of loads on the thigh, shank, and foot.

Changes in lower-extremity kinematics during walking may depend on the location of the added load. Compared with normal walking, lower-extremity kinematics are similar when a load is carried on the waist (13). Although the kinematics of walking with thigh loads have not been reported, lower-extremity kinematics do not change with thigh loading during running (18). Walking with heavy ankle loads increased stride length compared with normal walking (15). However, Royer et al. have reported recently that stride length increased only slightly (1%) with the addition of a 2-kg mass to the proximal and distal shanks of each leg (28). The kinematic conservation observed with thigh and shank loads may be a strategy to control the position of the foot, a phenomenon that has been demonstrated in normal walking (1).

Any increase in net metabolic rate during walking with an external load would, presumably, be accompanied by changes in the biomechanical determinants of metabolic rate. The stance-phase determinants of net metabolic rate (body-weight support and center-of-mass work) increase in proportion when load is carried at the waist (13). If the stance phase of walking is modeled as a point-mass inverted pendulum, then leg loads should not change stance-phase mechanics. Thus, any increase in net metabolic rate when loads are added to the legs compared with the waist should be associated primarily with the energy required for swinging the legs. Lower-extremity loads increase the moment of inertia of the swing leg about the hip joint ($I_{leg}$). This greater $I_{leg}$ has been shown to independently increase the swing-phase net muscle moments at the hip and knee compared with normal walking (28).

Electromyographic (EMG) activity can provide additional insights into the relationship between loading and metabolic rate. If the increase in net metabolic rate for walking with leg loading is caused by swinging the leg, the EMG of the muscles responsible for leg swing should increase. There is some debate as to the muscles associated with initiating and propagating leg swing. Some studies conclude that the ankle plantarflexors, which are active during mid- to late stance, are important swing initiators (14,23). Other studies discount the plantarflexors and suggest that the hip flexors are primarily responsible for leg swing initiation (10,22). At the end of swing, the lower leg must be decelerated before heel strike, a task that is performed by the knee flexor muscles (6). Although studies on the effect of leg loading on the EMG of the leg muscles are limited (28), one would expect that the muscles responsible for leg swing initiation, propagation, and termination would be more active with leg loading compared with waist loading.

The purpose of this study was to measure the effects of added leg-load magnitude and location on net metabolic rate during walking and to quantify the kinematic, kinetic, and muscular factors associated with leg loading. We hypothesized the following: 1) Net metabolic rate during walking would increase with leg loading, and the increase would be greater for more distal load locations. Net metabolic rate during walking would, therefore, be related to the moment of inertia of the leg about the hip joint. 2) Lower-extremity kinematics would be unchanged with loading, except for the extreme foot loads, which would result in increased stride lengths. 3) Any increase in net metabolic rate with leg loading would be associated with greater net muscle moments during late stance and swing and increased EMG of the leg muscles responsible for leg swing initiation, propagation, and termination.

## METHODS

**Experimental design.** Five healthy male subjects, (74.16 ± 5.18 kg, 1.82 ± 0.05 m, mean ± SD) volunteered to participate in this experiment. Subjects gave written informed consent, and the study was approved by the University of Colorado human research institutional review board. Subjects walked on a motorized force-measuring treadmill (17) at 1.25 m·s$^{-1}$. Experiments began with a no-load standing trial, followed by a no-load walking trial, then, in random order: sets of foot-, shank-, thigh-, and waist-load trials. A set consisted of loaded trials (in increasing mass order) for each location (e.g., foot set was two trials: a trial with 2 kg on each foot, and a trial with 4 kg on each foot). An additional no-load trial was inserted between each set of trials for each loading location. All trials lasted 7 min to allow for steady-state metabolic measurements. We collected kinematic and EMG data during the fifth minute of each trial. Subjects rested at least 5 min between trials. Because of the duration of the experiment (~6 h per subject), subjects drank water and ate small amounts of mixed diet snacks during rest periods.

**Leg and waist loading.** Load magnitudes at each segment were selected on the basis of practical relevance (lighter loads) and the maximum load that could be tolerated by the subjects (heavier loads). By using two loads at each segment, we were able to determine the effect of load magnitude on our energetic and biomechanical measures. In addition, we elected to use maximum tolerable loads to investigate neuromuscular control strategies. We used 2- and 4-kg loads per limb for the foot and shank. The 2-kg foot/shank load is similar to the mass of lower-leg prosthetics (19) and powered ankle-foot orthosis (9). We used 4- and 8-kg loads per limb for the

thigh. The 4-kg thigh load is similar to the extra thigh mass of obese versus normal-weight individuals (3). To allow equal-mass control comparisons, subjects also walked with 4-, 8-, and 16-kg loads wrapped around the waist over a well-padded hip belt.

Leg loads consisted of 0.125-inch-thick, rectangular strips of lead attached to either the foot, shank, or thigh segments. Duct tape covered the lead strips to prevent exposure of the skin to the lead. For the foot loads, we wrapped lead around the heel and over the dorsal (top) surface of the subject's shoes and secured them with duct tape (Fig. 1). We positioned the lead strips so that the location of the center of mass of the foot (5) was not altered. Heel and dorsal loads were placed an equal distance behind and in front of the foot center of mass, respectively. For shank loading, we protected the skin by covering the upper shank with athletic prewrap tape, and then we taped the lead strips to the prewrap around the circumference of the shank. We estimated the position of the center of mass of the shank for each subject using standard anthropometric equations (5), and we positioned the center of the lead strips at that proximal–distal location. Thigh loads were positioned and attached to the thighs in the same manner as the shank loads. For the remainder of the manuscript, we refer to the total load at each location, rather than the load per leg.

**Energetic measurements.** To quantify the effect of added mass on the net metabolic rate during walking, we measured the rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) using an open-circuit respirometry system (Physio-Dyne Instrument Corporation, Quogue, NY). Before beginning the experimental trials, we measured standing metabolic rate. For all trials, subjects walked 5 min to reach steady state, and then we calculated the average $\dot{V}O_2$ (mL $O_2 \cdot s^{-1}$) and $\dot{V}CO_2$ (mL $CO_2 \cdot s^{-1}$) for the subsequent 2 min. We calculated metabolic rate ($W \cdot kg^{-1}$ body mass) using a standard equation (2), and we derived net metabolic rate by subtracting the standing value from the experimental values.

**Kinematics.** We used foot switches and high-speed video techniques to record the lower-extremity kinematics.



FIGURE 1—Subject with 4-kg load attached to the foot. Lead strips were wrapped around the heel and over the dorsal aspect of the foot and secured with duct tape.

Piezoresistive foot switches (B&L Engineering, Tustin, CA) were placed in the subject's shoes and connected to a DC voltage circuit. The foot-switch voltage sampling rate was 1000 Hz, and the output signal was low-pass filtered at 25 Hz using a recursive fourth-order digital Butterworth filter. We identified heel strike by the rise in foot-switch voltage above 0 V and toe-off by foot-switch voltage returning to 0 V.

We placed lightweight reflective markers on the right leg at the following anatomic locations: fifth metatarsal–phalangeal joint, lateral malleolus, a point midway between the lateral epicondyle of the femur and the head of the fibula, greater trochanter, and the anterior superior border of the iliac spine. We recorded sagittal-plane marker trajectories at 200 Hz using a high-speed video camera (JC Labs), and we digitized the video data using the Peak Motus Measuring System (Version 5.0). Raw coordinate data were low-pass filtered at 5 Hz using a fourth-order zero-lag digital Butterworth filter. We used the filtered marker coordinates to determine the absolute elevation angles of the thigh, shank, and foot. Segment elevation angles were defined counterclockwise relative to the right-hand horizontal axis. Segment linear and rotational velocities and accelerations were calculated using finite difference equations within the Peak Motus software.

**Kinetics.** Using the force-measuring treadmill, we collected vertical and horizontal ground reaction forces for 10 s at 1000 Hz. The ground reaction force data were filtered using a recursive fourth-order Butterworth zero-lag low-pass filter with a cutoff frequency of 12 Hz. To calculate net muscle moments about the ankle, knee, and hip, we used an inverse dynamics solution. We calculated net muscle moments only during the right-leg-stance and swing phases because it was not possible to isolate the right-leg ground reaction forces during double support. Kinematic and force data were synchronized to the right heel strike using a synchronization signal and the foot-switch data. To determine the mass, center of mass location, and moment of inertia of the leg segments, we used standard anthropometric prediction equations (5). The increased moment of inertia attributable to the added mass was estimated by modeling the added mass as a hollow cylinder wrapped around the center of mass of the segment. Net muscle moments were determined at equally spaced points throughout the single-stance and swing phases using a cubic spline. We calculated the mean net muscle-moment patterns for five consecutive strides for each subject and the mean across subjects for each loading condition.

**EMG.** To quantify the effects of leg and waist loading on muscle activity, we measured surface EMG signals using a telemetered amplifier system (Noraxon, Scottsdale, AZ) and International Society for Electrophysiology and Kinesiology standard procedures (20). Before electrode placement, we prepared the shaved skin of the right leg with fine sandpaper and alcohol. We placed bipolar

silver-chloride surface electrodes (1-cm diameter discs) over the medial gastrocnemius (MG), soleus (SOL), anterior tibialis (TA), vastus medialis (VM), rectus femoris (RF), and semitendinosus (ST) muscles. The interelectrode distance was 2 cm. The EMG amplifier gain was 1700. We used a series of contractions suggested by Winter et al. (31) to verify that the cross-talk between the muscles was negligible.

EMG sampling rate was 1000 Hz, and the data were filtered and rectified. Specifically, we digitally filtered the signals with a zero-lag, fourth-order Butterworth filter with an effective band pass of 16–500 Hz, and then we full-wave-rectified the signals. To determine when each muscle was active, we digitally filtered the rectified signals with a low-pass filter with a cutoff frequency of 7 Hz to generate a linear envelope. We defined a muscle as active when the linear envelope exceeded a threshold of three standard deviations above the baseline mean for at least 100 ms. The baseline mean was determined from the EMG linear envelope during a period in the stride when muscle activity is normally absent (10). We calculated the mean EMG amplitude (mEMG) of the rectified signals during specific portions of the stride when the muscle of interest was active and averaged five consecutive strides for each loading condition. For each subject, we compared the relative muscle activity between loading conditions using the mEMG normalized to the no-load condition.

**Statistical analysis.** To determine an adequate sample size, we calculated the statistical power (http://calculators.stat.ucla.edu/powercalc) to detect changes in metabolic rate and lower-extremity kinematics using an alpha level of 0.05. The power analysis assumed paired *t*-tests or equivalently one-sided tests of the differences between the loading conditions. Using the walking metabolic rate data of Ralston (25) and kinematic data of Winter (30), we estimated that we would have 80% power to detect a 9% change in gross walking metabolic rate and a 2.7° change in segment elevation angles in our sample of five adults. We considered the ability to detect changes of 9% in walking metabolic rate acceptable, because the lower-extremity loads used in this study would increase walking metabolic rate by more than 10%.

We hypothesized *a priori* that increasing load magnitudes (e.g., 0-, 4- and 8-kg load on the feet) and more distal load locations (e.g., 4 kg at the waist, shanks, and feet) would increase metabolic rate, net muscle moments, and EMG. We compared the means of the metabolic rate, kinematics, net muscle moments, and EMG when load was increased at a specific location (e.g., waist). We also compared the means of the dependent variables mentioned above when a given load (e.g., 8 kg) was carried at different locations. We used paired *t*-tests with levels of significance adjusted using the Dunn (Bonferroni) method of multiple comparisons of means. Thus, significance was defined as $P < 0.008$ for waist-load comparisons, $P < 0.017$ for thigh-, shank-, and foot-load comparisons, $P < 0.017$

for 4-kg-load comparisons, $P < 0.008$ for 8-kg-load comparisons, and $P < 0.05$ for 16-kg-load comparisons.

## RESULTS

**Energetics.** In general, net metabolic rate during walking increased with load magnitude and more distal location (Fig. 2A). Although the 4-kg waist load did not result in a significant increase in net metabolic rate compared with no-load walking ($P = 0.078$), the 8- and 16-kg waist loads increased net metabolic rate by 14 and 32%, respectively ($P = 0.002$). Walking with 8-kg thigh and foot loads (total load) increased net metabolic rate above the 8-kg waist-load values by 14 ($P = 0.005$) and 48% ($P < 0.001$), respectively. Although average net metabolic rate numerically increased 9% when walking
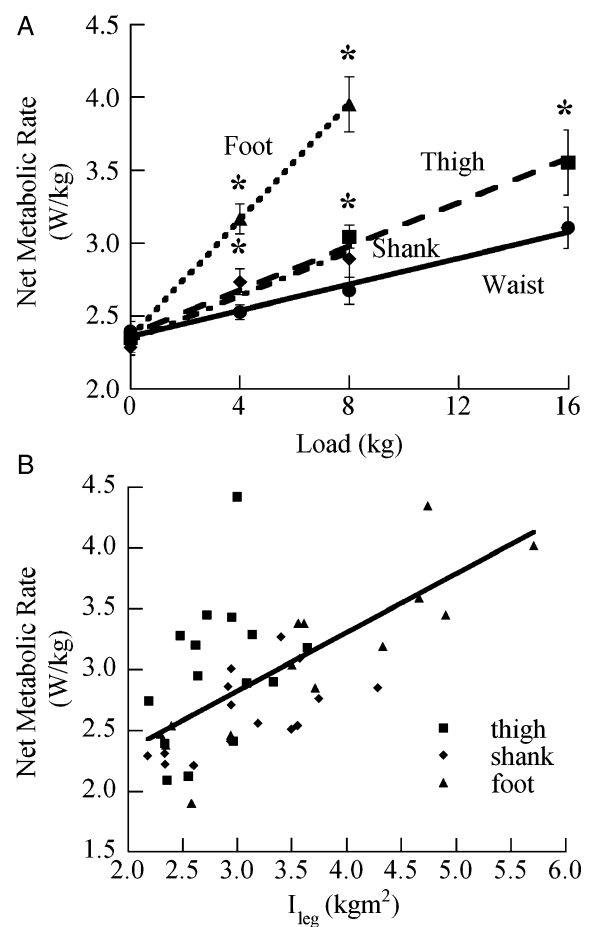


FIGURE 2— *A*, Mean net metabolic rate (W·kg$^{-1}$) vs load mass (*m*) and location. *B*, Net metabolic rate vs moment of inertia of the leg ($I_{leg}$) for each subject. The increase in net metabolic rate was greatest with foot loads, similar for thigh and shank loads, and least for waist loads. Linear regression equations for panel A: Waist watts per kilogram = 2.36 + 0.045*m*, $r^2 = 0.65$. Thigh watts per kilogram = 2.38 + 0.075*m*, $r^2 = 0.72$. Shank watts per kilogram = 2.34 + 0.076*m*, $r^2 = 0.61$. Foot watts per kilogram = 2.36 + 0.20*m*, $r^2 = 0.85$. Linear regression equations for panel B: watts per kilogram = 1.3774 + 0.48312$I_{leg}$, $r^2 = 0.43$. * Significant difference, load location vs waist load.

with 8-kg shank versus waist load, it was not a statistically significant difference ($P = 0.068$).

An increase in moment of inertia of the leg ($I_{\text{leg}}$) attributable to leg loading corresponded to an increase in net metabolic rate (Fig. 2B). $I_{\text{leg}}$ was $2.52 \pm 0.11$, $3.10 \pm 0.12$, and $3.74 \pm 0.15$ kg·m$^{-2}$ (mean ± SE) when the subjects walked with a 4-kg load on the waist (no load), shank, and foot, respectively. $I_{\text{leg}}$ was $2.52 \pm 0.11$, $2.82 \pm 0.19$, $3.70 \pm 0.16$, and $4.95 \pm 0.16$ kg·m$^{-2}$ with an 8-kg load on the waist (no load), thigh, shank, and foot, respectively. Thus, the 4- and 8-kg foot loads increased $I_{\text{leg}}$ by 49% (3.74 vs 2.52 kg·m$^{-2}$) and 97% (4.95 vs 2.52 kg·m$^{-2}$), and in response, the net metabolic rate increased by 25 and 48% compared with the waist-load condition.

**Kinematics.** Temporal stride kinematics did not change with loading, except for the foot loads. Stance represented approximately 62% of the gait cycle time, and double support was approximately 43% of the stance time in the no-load and waist-, thigh-, and shank-load conditions. With foot loading, stride length increased (Table 1), stride frequency decreased, and swing time increased compared with the other trials. Compared with the other load conditions, the 4-kg foot load increased stride length by 6% ($P = 0.003$) and swing time by 9% ($P = 0.003$), whereas the 8-kg foot load increased stride length by 11% ($P < 0.001$) and swing time by 10% ($P < 0.001$).

The thigh-, shank-, and foot-segment elevation angles (angle of segment relative to right horizontal axis) generally decreased during stance and increased during swing (Fig. 3), and angular leg kinematics did not change during the no-load or the waist-, thigh-, or shank-load trials. The segment elevation angles at toe-off decreased slightly, indicating an increased leg extension during the foot-load conditions compared with the other conditions. Thus, the greater stride length observed with foot loading was a result of the collective combination of decreases in the thigh-, shank-, and foot-segment elevations.

**Kinetics.** During the single-stance phase, the sagittal-plane net muscle-moment patterns across the hip, knee, and ankle joints were similar across all loading conditions (Fig. 4). The hip net muscle moments were extensor during early single stance and flexor during late single stance. The 8-kg foot load resulted in a significantly greater peak hip extensor (0.75 vs 0.43 N·m·kg$^{-1}$, $P = 0.005$) and flexor (0.86 vs 0.44 N·m·kg$^{-1}$, $P = 0.003$) moments compared

with the no-load condition. The peak hip moments during the other loading conditions were not statistically different compared with the no-load walking trial. At the knee, the net muscle moment was extensor throughout most of single stance and tended to be greater for the 16-kg thigh- and waist-loading trails. Peak single-stance knee moments were not significantly different across loading conditions. Ankle net muscle moments were plantarflexor throughout single stance, and peak moments were similar across all loading conditions.

Net muscle moments across the hip, knee, and ankle joints during swing were relatively small in all loading conditions, but these increased with foot loading (Fig. 5). During no-load walking, the hip net muscle moment was flexor during the first part of swing, extensor during midswing, and flexor again in late swing. Foot loads increased the flexor moment during early swing and, in the case of the 8-kg foot load, resulted in a hip extensor moment in late swing. The peak hip flexor moment increased 34% ($P = 0.003$) and 58% ($P < 0.001$) with 4- and 8-kg foot loads, respectively. At the knee, the net muscle moment was small during the first half of swing, and there was a flexor moment during late swing. To arrest knee extension at the end of the swing phase, the knee flexor moment greatly increased with foot loading, with peak flexor moments that were 54% ($P < 0.001$) and 100% ($P < 0.001$) greater with the 4- and 8-kg loads compared with the no-load condition. Although ankle moments were very small, the foot loads resulted in an increased dorsiflexor moment during early swing, presumably to prevent toe drag.

**EMG.** The temporal characteristics of the muscle activity patterns did not change across loading conditions. Because we were primarily interested in the muscles that initiate, propagate, and stop the swing leg, we analyzed the SOL, MG, TA, RF, and ST magnitudes during particular periods of the stride associated with leg swing. Specifically, we analyzed SOL and MG during mid- to late stance (32–52% of stride), RF during late stance (57–62% of stride) and early swing (63–67% of stride), and TA and ST during late swing (90–100% of stride). We confirmed that each muscle was active during the phase of interest.

Muscle activity generally increased with loading and distal load location (Fig. 6). Midstance SOL and MG

TABLE 1. Temporal stride kinematics vs added load location and mass for walking at 1.25 m·s$^{-1}$.

| Load Location | Mass (kg) | Stride Length (m) | Stride Rate (Hz) | Stance Time (s) | Swing Time (s) | Double Support Time (s) |
|---|---|---|---|---|---|---|
| No load | | 1.41 (0.02) | 0.88 (0.01) | 0.70 (0.01) | 0.43 (0.01) | 0.30 (0.02) |
| Waist | 16 | 1.42 (0.02) | 0.88 (0.01) | 0.72 (0.01) | 0.42 (0.01) | 0.33* (0.02) |
| Thigh | 16 | 1.41 (0.01) | 0.89 (0.01) | 0.71 (0.01) | 0.42 (0.01) | 0.32 (0.02) |
| Shank | 8 | 1.44 (0.03) | 0.87 (0.02) | 0.70 (0.01) | 0.44 (0.01) | 0.30 (0.02) |
| Foot | 4 | 1.49* (0.02) | 0.84* (0.00) | 0.73 (0.01) | 0.47* (0.01) | 0.30 (0.02) |
| Foot | 8 | 1.57*† (0.03) | 0.80*† (0.00) | 0.75*† (0.01) | 0.50*† (0.01) | 0.30 (0.02) |

Values are mean (SE). * $P < 0.017$, load vs no-load condition. † $P < 0.017$, 8- vs 4-kg foot-load condition. The temporal kinematics of all other load conditions were not significantly different from the no-load condition.
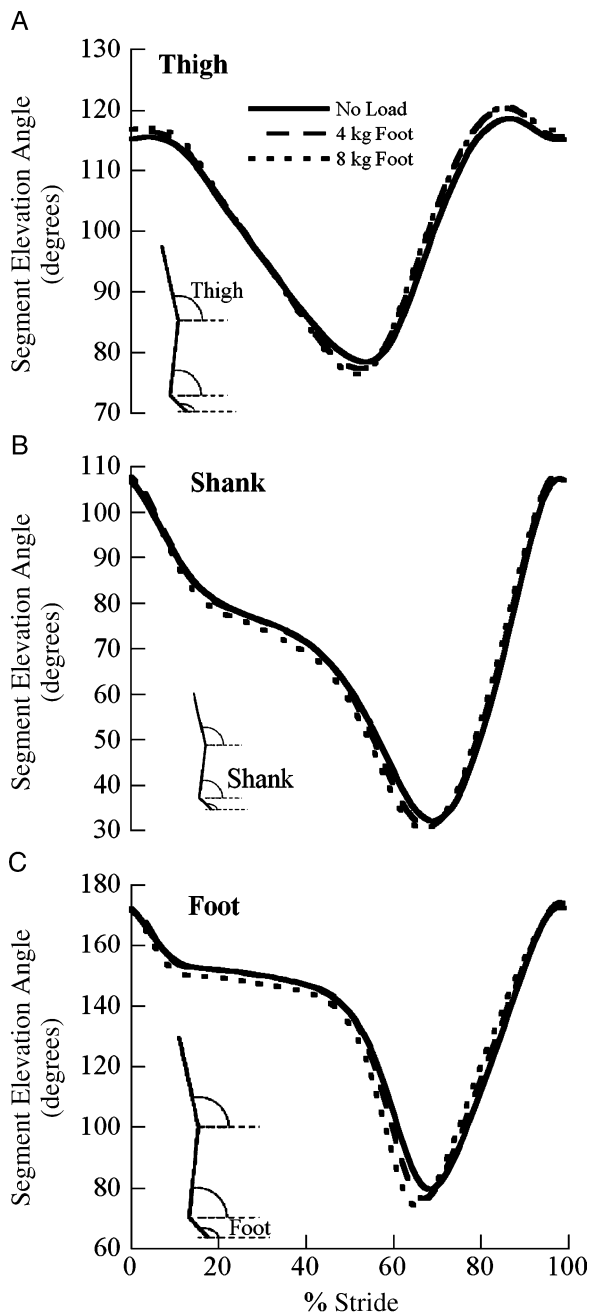
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FIGURE 3—Elevation angles vs percent stride for the thigh (*A*), shank (*B*), and foot (*C*) for the no-load and foot-loading conditions. Stride begins at heel strike (i.e., 0% stride = heel strike). Values are means for all subjects. Elevation angles are relative to the right horizontal axis. In general, foot loading resulted in small, non-significant decreases in the segment elevation angles at toe-off. The 8-kg foot load resulted in a significantly smaller shank elevation angle at toe-off compared with the no-load trial. The other loading conditions were even more similar to the no-load condition and are not shown.

mEMG were significantly greater with the maximum loads on the waist, thigh, and foot compared with the no-load condition. In addition, MG mEMG increased with 4 kg on the shank compared with the no-load condition ($P = 0.01$). SOL mEMG increased with 4 kg on the foot compared

with a similar waist load ($P = 0.002$), whereas MG mEMG did not change significantly ($P = 0.09$). There were no significant differences in SOL or MG mEMG between 8-kg waist versus foot trials ($P = 0.028$ and 0.01,
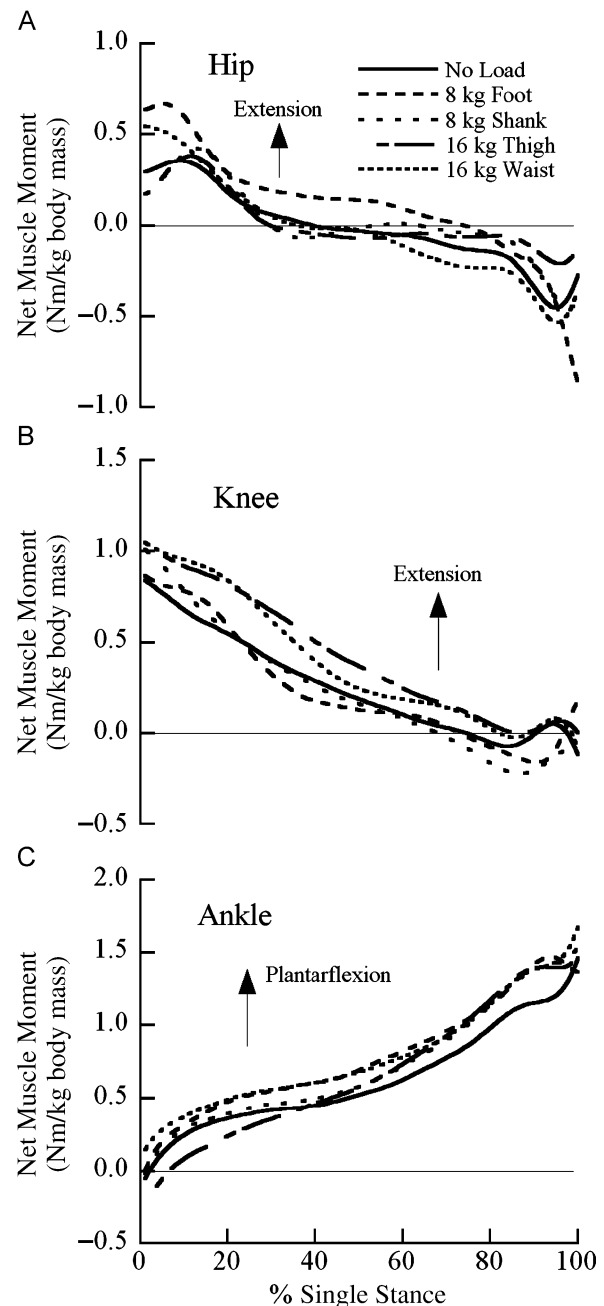


FIGURE 4—Net muscle moments per kilogram of body mass vs percent single stance for the hip (*A*), knee (*B*), and ankle (*C*) for the no-load and maximum segment load conditions. Stride begins at heel strike (i.e., 0% stride = heel strike). Values are means for all subjects. Positive moments are extensor. Net muscle-moment patterns during single stance were similar across loading conditions. Peak hip extensor and flexor moments during early and late single stance, respectively, were significantly greater with the 8-kg foot vs no-load trials. Peak knee and ankle net muscle moments were not significantly different across loading conditions.
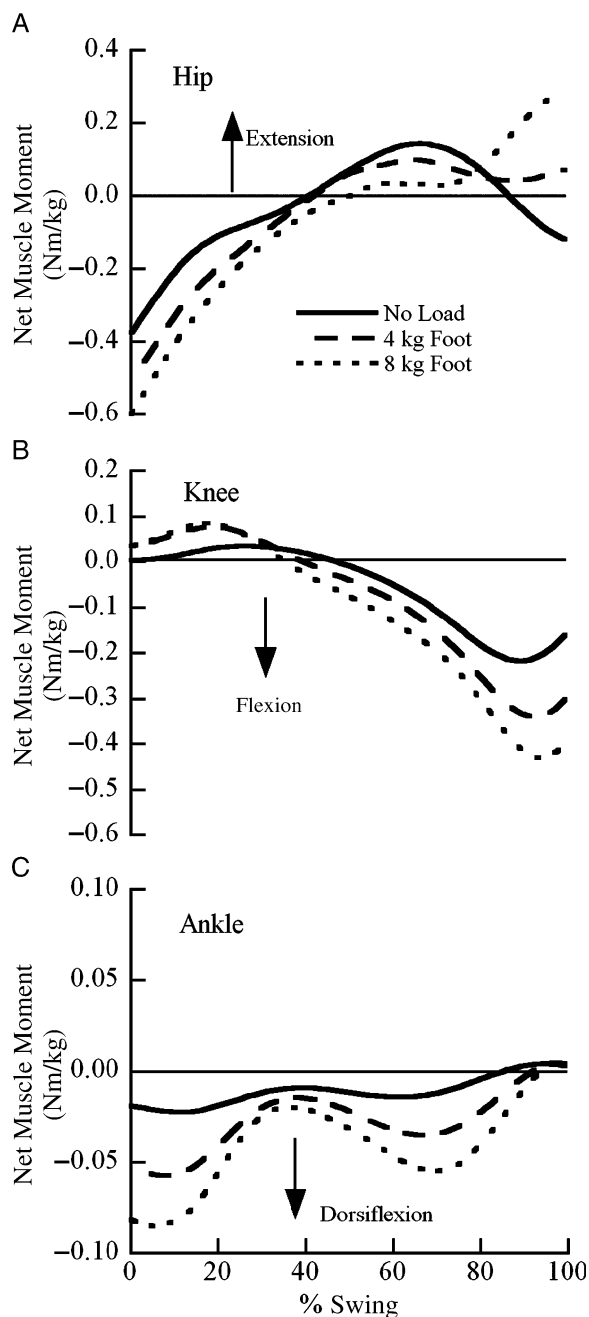
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locations. TA mEMG during late swing only increased significantly with the 8-kg foot loads ($P = 0.002$). There were no other significant differences in TA or ST mEMG during late swing compared across loads or locations.



FIGURE 5—Mean net muscle moments per kilogram of body mass vs percent swing for the hip (*A*), knee (*B*), and ankle (*C*) for the no-load and foot-loading conditions. Swing begins at toe-off. Values are means for all subjects. Positive moments are extensor. Foot loads resulted in significantly greater hip flexion and extension moments at toe-off and heel-strike, respectively. Foot loading required a greater knee extensor moment during early swing and a greater knee flexor moment to slow the foot before heel strike. Ankle moments were relatively small throughout the swing phase, but peak dorsiflexor moments were significantly greater with foot vs no-load walking. The other loading conditions are not shown because they were very similar to the no-load condition.

respectively). RF mEMG during late stance increased significantly with the 8-kg foot loads ($P = 0.005$), but there were no other significant differences in RF mEMG during late stance or early swing compared across loads or
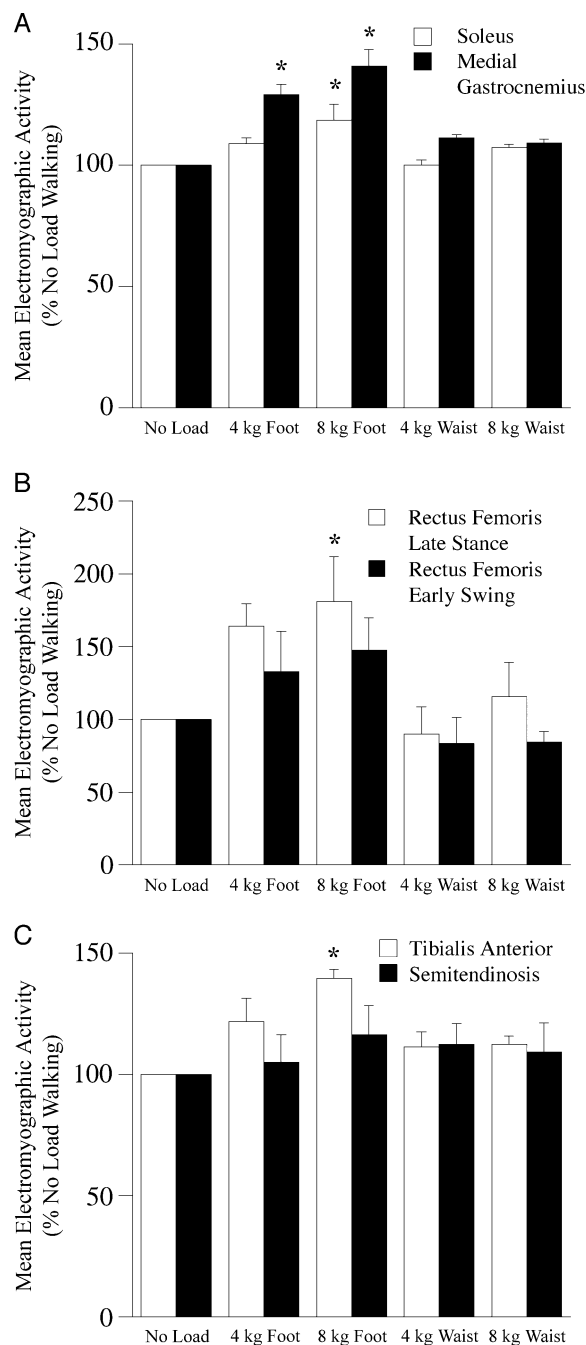
FIGURE 6—Normalized mean electromyogram vs load condition relative to unloaded walking for the soleus and medial gastrocnemius during midstance (32–52% stride) (*A*), the rectus femoris during late stance (57–62% stride) and early swing (62–67% stride) (*B*), and the tibialis anterior and semitendinosus during late swing (90–100% stride) (*C*). Values are means for all subjects. EMG magnitudes tended to increase with foot loading in the soleus, medial gastrocnemius, and rectus femoris during stance and the tibialis anterior during swing.

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## DISCUSSION

**Energetics.** We accept our hypothesis that the net metabolic rate for walking increases with leg loads and is greater with more distal load locations. As the amount of added mass increased at a particular location, the net metabolic rate increased. The 16-kg waist load represented an increase of approximately 21% of body mass and increased net metabolic rate by 32% compared with the no-load condition. This increase in net metabolic rate is similar to the approximately 27% predicted by Griffin et al. (13) for a load of 20% body mass. Foot loads were considerably more expensive than the other loading conditions. The 36% increase in net metabolic rate with a 4-kg foot load compared with the no-load condition was similar to the approximately 30% increase reported by Rose et al. when their subjects walked with a 4-kg foot load (26).

The net metabolic rate during walking was similar with the same mass on the thigh or shank (i.e., 8 kg). This unexpected result may be attributable to the way we applied the load to the thigh. We wrapped the lead strips around the circumference of the leg at the approximate center of mass of the thigh. The strips of lead on the medial of the legs may have caused step width and/or lateral leg-swing circumduction to increase. Net metabolic rate increases when step width and lateral leg circumduction are increased (7). Net metabolic rate measured with thigh loads may reflect both the addition of the load and frontal-plane alterations in gait.

The relatively small increase (6%) in net metabolic rate between the 4- and 8-kg shank loads was unanticipated. Net metabolic rate increased 8% in the 4-kg shank versus 4-kg waist trials. This increase is consistent with the data of Royer and Martin (28), who report that net metabolic rate further increased by 8% when comparing a 5.6-kg shank load (2.8 kg per shank) against a 5.6-kg load at the waist. Assuming a linear increase in net metabolic rate with shank loading, one would predict approximately 3 $W \cdot kg^{-1}$ for the 8-kg shank trial, rather than the 2.85 $W \cdot kg^{-1}$ we report here. We cannot readily explain the relatively low net metabolic rate for walking with a heavy shank load, but results from shank-loading experiments in individuals walking with prosthesis are also paradoxical. For example, Czerniecki et al. (4) have reported that the addition of 1.34 kg to a prosthetic shank did not increase the metabolic cost of walking among above-knee amputees. Conversely, Mattes et al. (19) report that metabolic rate was 6% greater when a mass of 1.7 kg was added to the prosthetic shank of transtibial amputees.

As we hypothesized, net metabolic rate during walking was related to $I_{leg}$ ($r^2 = 0.43$, $P < 0.001$). However, the strength of this correlation is strongly influenced by the 8-kg foot-loading trials. When the data from these trials are removed, the correlation between net metabolic rate and $I_{leg}$ is not as strong ($r^2 = 0.24$). Figure 7A shows the relative increase in net metabolic rate versus the relative
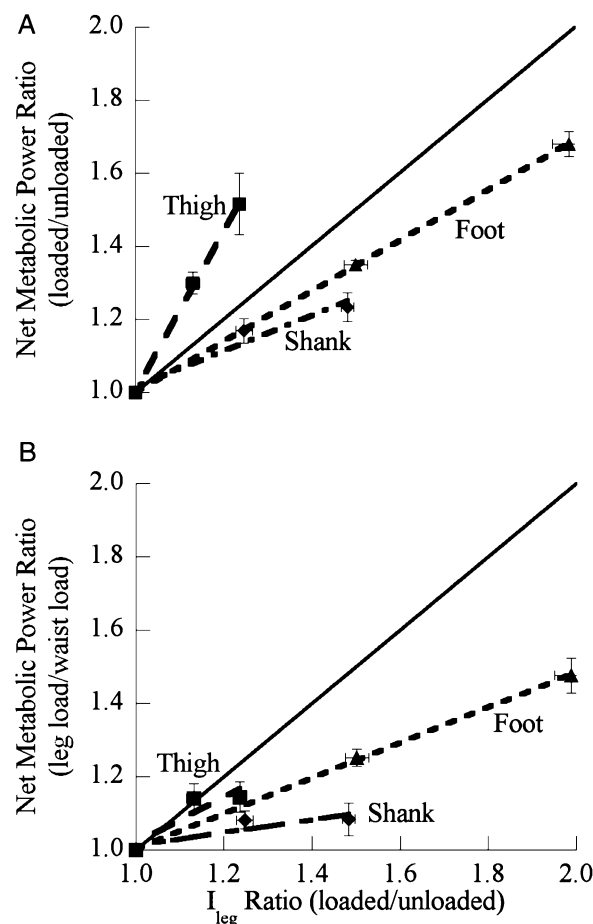


**FIGURE 7— A**, Net metabolic power ratio (load/no load) vs $I_{leg}$ ratio (load/no load). The *thin solid line* is the line of identity. Relative increases in $I_{leg}$ with thigh loading resulted in greater increases in net metabolic rate than more distal loading locations. Linear regression equations for panel A: thigh metabolic ratio = $-0.74 + 1.81 I_{ratio}$, $r^2 = 0.50$. Shank metabolic ratio = $0.63749 + 0.40916 I_{ratio}$, $r^2 = 0.46$. Foot metabolic ratio = $0.32639 + 0.68067 I_{ratio}$, $r^2 = 0.98$. **B**, Metabolic rate ratio (load/waist load) vs $I_{leg}$ ratio (load/waist load). Expressed relative to waist loads, increases in $I_{leg}$ resulted in similar increases in net metabolic rate with thigh and foot loading.

increase in $I_{leg}$. The strong correlation between the net metabolic rate ratio and $I_{leg}$ with foot loading ($r^2 = 0.98$) suggests that as $I_{leg}$ becomes large, $I_{leg}$ is a primary determinant of net metabolic rate for walking. However, the difference in the slope of the net metabolic rate ratio versus $I_{leg}$ ratio for thigh, shank, and foot loading suggests that other factors influence net metabolic rate. Mass itself is one such factor. Royer and Martin (28) have found that net metabolic rate during walking increased compared with a baseline leg-loading condition when leg mass was increased without increasing $I_{leg}$. This mass effect may have been particularly important with the heavy thigh loads used in this study. Such loads would presumably increase the activity of the hip abductors, which act to support the pelvis while the contralateral leg is swinging (24) and to laterally swing the heavy leg. As a result, net metabolic rate might increase above what would be expected on the basis of $I_{leg}$.

Further support for the effect of load mass is evident when the net metabolic rate ratio is expressed relative to an equivalent waist load (Fig. 7B). This ratio reflects the energy required for swinging the leg, which is similar for the thigh and foot, given an equivalent increase in moment of inertia of the leg. For example, with thigh and foot loads, a 25% increase in $I_{leg}$ results in an approximately 15% increase in net metabolic rate compared with walking with the same load at the waist. The fact that the slopes of the lines in Figure 7B are less than identity also demonstrates that sagittal-plane leg swing and $I_{leg}$ ratio are not the only determinants of net metabolic rate during walking.

Our results demonstrate that, compared with more distal locations, loads can be carried on the thighs relatively inexpensively. The 16-kg thigh load represents an approximately 83% increase in the mass of the thighs, assuming that the each thigh is approximately 13% of total body mass (8). Expressed as a ratio of thigh mass to body mass, the 16-kg thigh load increased this ratio from 13 to 19.5%, an increase of 50%. Despite these dramatic increases in absolute and relative leg mass, net metabolic rate increased by only 14% compared with the 16-kg waist load trial.

The finding that the relative mass of the thigh has a small effect on net metabolic rate during walking helps explain our recent finding that net metabolic rate during walking was not related to body mass distribution (i.e., thigh mass/body mass) in obese and normal-weight adults (3). In that study, we found that although the thigh mass/body mass ratio was 7.5% in obese females versus obese males (14.2 vs 13.2%, respectively), this ratio did not explain the 10% greater net metabolic rate of obese females versus males. These findings are supported by the results of the present study, given that a much more substantial increase in relative thigh mass resulted in only a modest increase in net metabolic rate.

**Kinematics.** We accept our hypothesis that, except for foot loads, lower-extremity kinematics would be conserved with loading. Our results are consistent with other studies that have used normal walking (21), waist loads (13), and foot loads (15). The 5% decrease in stride rate when walking with a 4-kg foot load is similar to the 6% decrease reported by Holt et al. (15) using a 4.4-kg foot load. The decreased stride rate with leg loading has been used as evidence that an energetically optimal stride rate is selected on the basis of the pendular dynamics of the swing leg (15). Although our foot-loading trials did support this idea, the finding that the other leg-loading conditions (e.g., 4-kg shank load) did not result in a change in stride frequency or swing time compared with the no-load condition does not support this theory.

The kinematic conservation we observed with leg loading supports the theory that the position and velocity of the foot are tightly controlled during walking (1,16). Borghese et al. (1) have shown that a plot of elevation angles of the lower-limb segments during a gait cycle covary along a plane. Planar covariation has been described as evidence of a law of intersegmental coordination that reduces the variability of the foot trajectory (1,16). In our study, segment elevation angles were highly reproducible, and when plotted against each other for a gait cycle, they displayed a planar covariation that was similar for all loading conditions. This finding suggests that foot position was controlled during walking with leg loads and adds support to the theory of kinematic conservation.

It is generally agreed that under normal conditions, humans select a walking pattern that minimizes metabolic cost ($J{\cdot}kg^{-1}{\cdot}m^{-1}$) (25). In the case of acute leg loading, this may not be true. If foot position is being controlled, the net metabolic rate associated with leg loading may reflect the neuromuscular system's attempt to maintain a foot position, rather than an attempt to minimize metabolic cost. Future studies that measure the effects of leg loading across a range of speeds and over time would provide insights into the relative role of kinematic control versus metabolic cost minimization.

**Kinetics.** As we hypothesized, swing-phase net muscle moments increased when walking with foot loads. These results help explain the increase in net metabolic rate associated with foot loads. However, swing-phase net muscle moments did not change with the other leg-loading conditions, yet there were clear increases in net metabolic rate. Our single-leg-stance and swing-phase net muscle-moment patterns during normal walking were similar to those reported by Winter (30) and Mills and Barrett (21). Royer and Martin (27) have reported that swing-phase hip and knee moments were similar when subjects walked with 2.82 kg attached to each shank versus no load, a finding consistent with our results but no less puzzling.

If sagittal-plane single-leg stance and swing-phase net muscle moments do not change with thigh and shank loading, what causes the increase in net metabolic rate above that for the same waist load? One possibility is that the changes in mechanics occur primarily during the double-support phase rather than during the single-stance or swing phase. This scenario would require an increase in the metabolic energy consumed to perform work on the center of mass, which seems unlikely given the similarity in kinematics across the load conditions. The leg and body centers of mass are coupled, however, as manipulations of leg mass alter the body center of mass. This coupling could change the work done to move the center of mass forward during the double-support phase of walking. Another possibility is that the increase in metabolic rate is caused by changes in frontal-plane mechanics, as mentioned above. It seems likely that the hip abduction moment of the stance leg would be greater while supporting the pelvis and loaded swing leg. Future studies that examine changes in the biomechanics of walking with leg loading in three dimensions and the measurement of external work of the individual limbs would be insightful.

**EMG.** EMG of the muscles that initiate, propagate, and terminate leg swing generally increased with leg loading,

as hypothesized. This finding helps explain the increase in net metabolic rate that occurred with leg loading. Our data also provide insights into the relative roles of the MG and SOL during walking. The increase in MG mEMG during mid- to late single stance with foot and shank loads suggests that it does assist in initiation of the swing leg, as predicted by Neptune et al. (23). Yet, as has been shown by others (10), the MG muscle was not active past 53% of the stride, indicating that it either actively initiates swing before late stance or that it may have a limited contribution to swing initiation. The increase in RF mEMG during late stance with the 8-kg foot loads is consistent with this muscle's role as a swing initiator (10). According to Nene et al. (22) and Gottschall and Kram (10), the RF also assists in the propagation of the swing leg. The early swing mEMG tended to increase with load (although not significantly), which is consistent with the swing propagation role for the RF.

Our methodology imposed some limitations. It is possible that the use of fixed mass amounts, rather than using a percent of body mass, introduced variability that made it more difficult to detect significant differences in our kinematic, kinetic, and EMG parameters. But, subject body mass only ranged from 68 to 77 kg, so the 8-kg load represented 12 and 10.3% of body mass for the lightest and heaviest subjects, respectively. In addition, our experimental design (number of subjects) may not have had sufficient statistical power to resolve all trends in our net muscle moment and EMG data.

## Summary

In conclusion, we have found that net metabolic rate during walking increased with leg-load magnitude and more distal leg-load location. Thus, net metabolic rate was fundamentally related to the moment of inertia of the leg. The relatively small increase in net metabolic rate with proximal loading helps explain why differences in body mass distribution do not result in large differences in metabolic rate during walking in normal-weight and obese adults. The increase in net metabolic rate with foot loading was, at least in part, a result of greater sagittal-plane single-stance and swing-phase net muscle moments and EMG of the muscles associated with leg-swing initiation, propagation, and termination. Although net metabolic rate increased with the thigh- and shank-load conditions, the gait kinematics were conserved, and net muscle moments were similar to those measured during normal walking. Therefore, we lack a clear explanation of why net metabolic rate increased with shank and thigh loading. Future studies that investigate the double-support-phase and frontal-plane net muscle moments may provide further insights into the relationship between net metabolic rate and loading of the lower extremity.

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