

Effects of Obesity on the Biomechanics of Walking at Different Speeds

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ABSTRACT

BROWNING, R. C. and R. KRAM. Effects of Obesity on the Biomechanics of Walking at Different Speeds. *Med. Sci. Sports Exerc.*, Vol. 39, No. 9, pp. 1632–1641, 2007. **Purpose:** Walking is a recommended form of exercise for the treatment of obesity, but walking may be a critical source of biomechanical loads that link obesity and musculoskeletal pathology, particularly knee osteoarthritis. We hypothesized that compared with normal-weight adults 1) obese adults would have greater absolute ground-reaction forces (GRF) during walking, but their GRF would be reduced at slower walking speeds; and 2) obese adults would have greater sagittal-plane absolute leg-joint moments at a given walking speed, but these moments would be reduced at slower walking speeds. **Methods:** We measured GRF and recorded sagittal-plane kinematics of 20 adults (10 obese and 10 normal weight) as they walked on a level, force-measuring treadmill at six speeds (0.5–1.75 m·s⁻¹). We calculated sagittal-plane net muscle moments at the hip, knee, and ankle. **Results:** Compared with their normal-weight peers, obese adults had much greater absolute GRF (N), stance-phase sagittal-plane net muscle moments (N·m) and step width (m). **Conclusions:** Greater sagittal-plane knee moments in the obese subjects suggest that they walked with greater knee-joint loads than normal-weight adults. Walking slower reduced GRF and net muscle moments and may be a risk-lowering strategy for obese adults who wish to walk for exercise. When obese subjects walked at 1.0 versus 1.5 m·s⁻¹, peak sagittal-plane knee moments were 45% less. Obese subjects walking at approximately 1.1 m·s⁻¹ would have the same absolute peak sagittal-plane knee net muscle moment as normal-weight subjects when they walk at their typical preferred speed of 1.4 m·s⁻¹. **Key Words:** LOCOMOTION, NET MUSCLE MOMENTS, BODY MASS, KNEE, OSTEOARTHRITIS

Obesity is the main preventable risk factor for large-joint (e.g. knee) osteoarthritis (OA) (30). Exercise combined with dietary intervention probably holds the best promise for combating the obesity epidemic (17). Walking is a recommended and popular form of exercise (18) but weight-bearing exercise (i.e., walking) may be a critical source of the biomechanical loads that link obesity and OA (15). Intuitively, it would seem likely that obesity greatly increases the biomechanical loads involved in walking and that these loads increase with walking speed. If so, the common prescription of brisk walking as exercise for treating obesity (28) may inadvertently increase the risk of musculoskeletal pathology.

Remarkably, few studies have been conducted on how obesity affects the biomechanical loads involved in walking (30). Spyropoulos et al. (35) compared stride and joint angle differences between obese and normal-weight men. They reported that the obese males walked slower (1.09 m·s⁻¹) with wider steps and similar knee flexion at midstance compared with their faster-walking (1.64 m·s⁻¹) normal-weight counterparts, but they could not discern the obesity effects from the speed effects. However, according to Lelas et al. (21), peak knee flexion during the loading response phase of stance increases with increased walking speed. Combining the two studies suggests that at matched speeds, obese men might adopt a more flexed knee during stance than their normal-weight peers.

Messier et al. (25) provide the only published reports of ground-reaction forces (GRF) for obese adults walking. Absolute peak vertical GRF increased in almost direct proportion with body weight. The peak vertical GRF were approximately 1.0 × body weight, which is consistent with the slow walking speeds selected by the subjects (1.0 m·s⁻¹). Absolute peak anteroposterior and mediolateral GRF also increased proportionally with body weight. By design, Messier et al. compared arthritic obese with arthritic normal-weight subjects. Thus, although insightful, this

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study provided no information about the GRF of non-arthritic obese adults walking at a normal walking speed.

To date, the most intriguing and thorough study of the biomechanics of walking by obese subjects is that of DeVita and Hortobagyi (8). They compared the sagittal-plane net muscle moments generated at the ankle, knee, and hip joints of nonarthritic class II and III obese adults (mean BMI $\approx 42 \text{ kg}\cdot\text{m}^{-2}$) and normal-weight controls. The knee is the primary site of obesity-related OA (13), yet, surprisingly, DeVita and Hortobagyi (8) found that the absolute peak knee moments when walking at $1.5 \text{ m}\cdot\text{s}^{-1}$ were identical: $64 \text{ N}\cdot\text{m}$ at $1.5 \text{ m}\cdot\text{s}^{-1}$ for obese (123 kg) versus normal-weight (63 kg) subjects. Although the obese subjects presumably had greater absolute GRF, they maintained their knee-joint angles in more extended positions that aligned the resultant GRF more closely to the center of the knee joint. As a result, the net muscle moment required and, by proxy, sagittal-plane knee-joint forces, were likely not greater. Although these data suggest that obese adults do not have greater knee-joint loads, it is unclear how the wide range of adiposity (BMI $30\text{--}60 \text{ kg}\cdot\text{m}^{-2}$) influenced the findings and whether their findings would be similar in a group of moderately obese adults (BMI $30\text{--}40 \text{ kg}\cdot\text{m}^{-2}$). Also, the kinematic results reported by DeVita and Hortobagyi are not consistent with those of Spyropoulos et al., so a clear consensus of the effects of obesity on lower-extremity sagittal-plane biomechanics has yet to be achieved.

Walking at slower speeds may be an effective way to reduce knee-joint loads in obese adults. Previous studies have shown that in normal-weight subjects, GRF and lower-extremity net muscle moments are smaller at slower walking speeds (21,40). In most studies, obese adults prefer to walk slower than their normal-weight peers (8,23). DeVita and Hortobagyi (8) reported that at a just slightly slower speed (1.29 vs $1.50 \text{ m}\cdot\text{s}^{-1}$), peak knee sagittal-plane net muscle moment in the obese subjects was reduced by 25%. However, no study has comprehensively examined the relationship between walking speed and lower-extremity joint loads in obese versus normal-weight adults.

The purpose of this study was to measure how obesity affects walking biomechanics, particularly knee-joint loads, by determining GRF and lower-extremity sagittal-plane joint moments across a range of walking speeds. We hypothesized that compared with normal-weight adults, 1) obese adults have greater absolute GRF during walking, but their GRF are reduced by walking slower; and 2) obese adults have greater absolute lower-extremity joint moments at a given walking speed, but these moments are reduced at slower walking speeds.

METHODS

Subjects. Two groups of young adults volunteered for this study: obese ($N = 10$; five females and five males) and normal-weight ($N = 10$; five females and five males). BMI

was used to classify the participants; obese subjects had BMI values of $30\text{--}43 \text{ kg}\cdot\text{m}^{-2}$, and normal-weight subjects had BMI values of $20\text{--}25 \text{ kg}\cdot\text{m}^{-2}$. All subjects were body-mass stable ($< 2.5\text{-kg}$ net change during the previous 3 months) and healthy (other than obesity). Subjects gave written informed consent that followed the guidelines of the University of Colorado Human Research Committee. The obese subjects were part of an earlier physiological study (5) that included a whole-body dual-energy x-ray absorptiometry (DEXA) scan. The physical characteristics of the groups are shown in Table 1.

Experimental protocol. Each subject performed six level walking trials on a dual-belt force-measuring treadmill. First, we familiarized the subjects to the treadmill by having them walk for at least 10 min at a comfortable walking speed. The familiarization was followed by the experimental trials. The treadmill speeds were $0.50, 0.75, 1.00, 1.25, 1.50,$ and $1.75 \text{ m}\cdot\text{s}^{-1}$, and the trial order was randomized for each subject. During each trial, subjects walked with the right leg on the right treadmill and the left leg on the left treadmill for 2 min, followed by 2 min of walking with both feet on the right treadmill.

GRF. We measured GRF using a dual-belt force-measuring treadmill. The treadmill is a hybrid of the designs developed by Kram et al. (20) and Belli et al. (3) with the force plate under the right treadmill. The right and left treadmill belts were separated by less than 1 cm. We collected right leg vertical, anteroposterior (AP) and mediolateral (ML) GRF and moments about the AP and ML axes for 10 s at 1000 Hz. The GRF data were filtered using a recursive fourth-order Butterworth low-pass filter with a cutoff frequency of 12 Hz. We used the vertical GRF data and a threshold force of 15 N to determine right foot heel strike and toe-off.

Kinematics. To record the kinematics, we used both footswitches and high-speed video techniques. Piezo-resistive footswitches (B&L Engineering, Tustin, CA) were placed in the subjects' shoes and connected to a DC voltage circuit. Footswitch 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 left foot heel strike by the rise in footswitch voltage above 0.25 V, and we identified left foot toe-off by footswitch voltage returning to below 0.25 V.

We placed lightweight reflective spheres on the right leg at the following anatomical locations: fifth metatarsal-phalangeal joint, lateral malleolus, a point midway between the lateral epicondyle of the femur and the head of the

TABLE 1. Physical characteristics of obese and normal-weight females and males.

	Female		Male	
	Obese (5)	Normal (5)	Obese (5)	Normal (5)
Body mass (kg)	106.2 (25.1)	58.5 (2.1)	115.0 (16.4)	76.8 (4.5)
Height (m)	1.69 (0.08)	1.67 (0.04)	1.83 (0.07)	1.82 (0.03)
BMI ($\text{kg}\cdot\text{m}^{-2}$)	37.0 (6.0)	21.0 (0.8)	34.1 (3.7)	23.1 (1.6)
Age (yr)	31.0 (7.1)	31.8 (5.7)	26.8 (7.6)	25.6 (9.8)

Values are mean (SD).

fibula, greater trochanter, and the anterior superior border of the iliac spine. To maximize the accuracy of marker placement on the obese subjects, we measured the DEXA scan images to determine the segmental distances between the lateral malleolus, knee joint, greater trochanter, and anterior border of the iliac spine. We then placed the reflective markers on the skin so that the segmental distances matched the DEXA measurements.

We recorded sagittal-plane marker trajectories at 200 Hz using a high-speed video camera (JC Labs) and digitized the video data using a 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 segment elevation angles of the thigh, shank, and foot and the hip-, knee-, and ankle-joint angles in the sagittal plane. Segment elevation angles were defined counterclockwise relative to the righthand horizontal axis. Segment linear and rotational velocities and accelerations were calculated using finite difference equations. Because of the physical space constraints of the laboratory, we were unable to record frontal-plane kinematics.

Net muscle moments. To calculate sagittal-plane net muscle moments about the hip, knee, and ankle, we used an inverse dynamics solution. Kinematic and force data were synchronized to the right heel-strike using a synchronization signal and the GRF data. Force data were down-sampled to 200 Hz to match the video sampling rate. To determine the thigh and shank body-segment parameters of the obese subjects, we digitized the DEXA scans to determine the thigh and shank segment lengths and cross-sectional areas, as we have described in detail previously (5). Briefly, segment endpoints of the thigh and shank were the superior border of the greater trochanter, knee-joint axis, and lateral malleolus. Polygons defined the thigh and shank cross-sectional area, and the DEXA software calculated segment mass. To determine thigh and shank radius of gyration, we used the regression equations provided by Durkin and Dowling (12) and calculated frontal-plane moment of inertia (I_{com}) using the segment mass and radius of gyration. Differences between frontal- and sagittal-plane segment parameters have been shown to be small (7), so we used the frontal-plane values to represent the sagittal-plane moments of inertia. Lower-extremity segment parameters

for the normal-weight subjects and the feet of the obese subjects were estimated using the equations of De Leva (7). Net muscle moments were normalized using a spline technique, to represent a percentage of the stride. We calculated the mean instantaneous net muscle moment for 4–10 consecutive strides at each speed for each subject, and the mean across subjects for each walking speed.

Step width. Step width was determined by calculating the distance between the midstance ML center of pressure location of successive steps while subjects walked with both feet on the right treadmill (11).

Statistical analysis. A two-factor (obesity and speed) ANOVA with repeated measures determined how obesity and walking speed affected peak GRF, temporal gait characteristics, midstance lower-extremity joint angles, peak net muscle moments, and peak external knee adduction moment. Main effects of group, speed, and the interaction effect of group and speed were calculated. Significant main effects were analyzed without the interaction if the interaction was found to be significant. A criterion of $P < 0.05$ defined significance. To distinguish difference between groups at each speed, we used Student's t -tests and applied a Bonferroni correction on the basis of the number of comparisons being made. Thus, the level of significance for the Student's t -tests was set at 0.0083.

RESULTS

GRF. Absolute GRF (N) were significantly greater for the obese versus normal-weight subjects and decreased significantly at slower walking speeds in both groups (Table 2, Fig. 1). At each walking speed, peak vertical GRF were approximately 60% greater for the obese versus normal-weight subjects ($P < 0.001$). Scaled to body weight, peak vertical GRF were significantly smaller for the obese subjects versus normal-weight subjects at 1.00 $m \cdot s^{-1}$. Slower walking speeds resulted in smaller magnitudes of the first peak of the vertical GRF, but they had less of an influence on the magnitude of the second peak of the vertical GRF. For example, in the obese subjects, the first and second peaks of the vertical GRF were 15 and 6% smaller, respectively, when walking at 1.0 versus 1.5 $m \cdot s^{-1}$.

Absolute peak AP GRF (N) were about 63% greater ($P < 0.001$) for the obese versus normal-weight subjects at the

TABLE 2. Peak ground-reaction forces (GRF) for obese vs normal-weight adults.

Speed ($m \cdot s^{-1}$)	Vertical (N)		Vertical (\times BW)		A-P (N)		M-L (N)		M-L (\times BW)	
	O	NW	O	NW	O	NW	O	NW	O	NW
0.50	1094* (64.8)	676 (32.1)	1.00 (0.01)	1.02 (0.01)	77.0* (5.9)	41.6 (1.6)	48.8* (6.7)	28.3 (2.4)	0.04 (0.004)	0.04 (0.002)
0.75	1081* (73.3)	669 (31.9)	0.99 (0.01)	1.01 (0.01)	109.5* (8.3)	63.8 (3.5)	63.7* (8.1)	33.3 (2.6)	0.06 (0.004)	0.05 (0.003)
1.00	1080* (64.6)	684 (39.3)	1.00* (0.01)	1.03 (0.01)	151.5* (9.7)	91.2 (5.4)	73.8* (10.8)	41.1 (3.5)	0.07 (0.006)	0.06 (0.003)
1.25	1158* (71.5)	743 (39.3)	1.07 (0.02)	1.12 (0.01)	201.1* (12.6)	121.5 (6.4)	83.7* (10.8)	43.2 (2.7)	0.09 (0.007)	0.07 (0.004)
1.50	1259* (76.2)	783 (44.6)	1.16 (0.02)	1.20 (0.02)	253.0* (15.4)	157.0 (10.9)	95.4* (10.5)	49.8 (3.8)	0.09 (0.007)	0.08 (0.006)
1.75	1383* (87.4)	874 (56.7)	1.27 (0.02)	1.31 (0.02)	303.1* (18.9)	183.0 (15.3)	105.8* (14.1)	58.7 (3.0)	0.10 (0.01)	0.09 (0.005)

Values are mean (SE). O, obese; NW, normal weight; peak vertical, first peak of the vertical GRF; A-P, anteroposterior braking force; M-L, mediolateral force of the first medial peak of the mediolateral GRF; BW, body weight. Absolute vertical, A-P, and M-L GRF were significantly greater in the obese vs normal-weight subjects at each speed. Scaled to body weight, vertical GRF was less in obese vs normal-weight subjects at 1.00 $m \cdot s^{-1}$.

* $P < 0.0083$, obese vs normal weight.

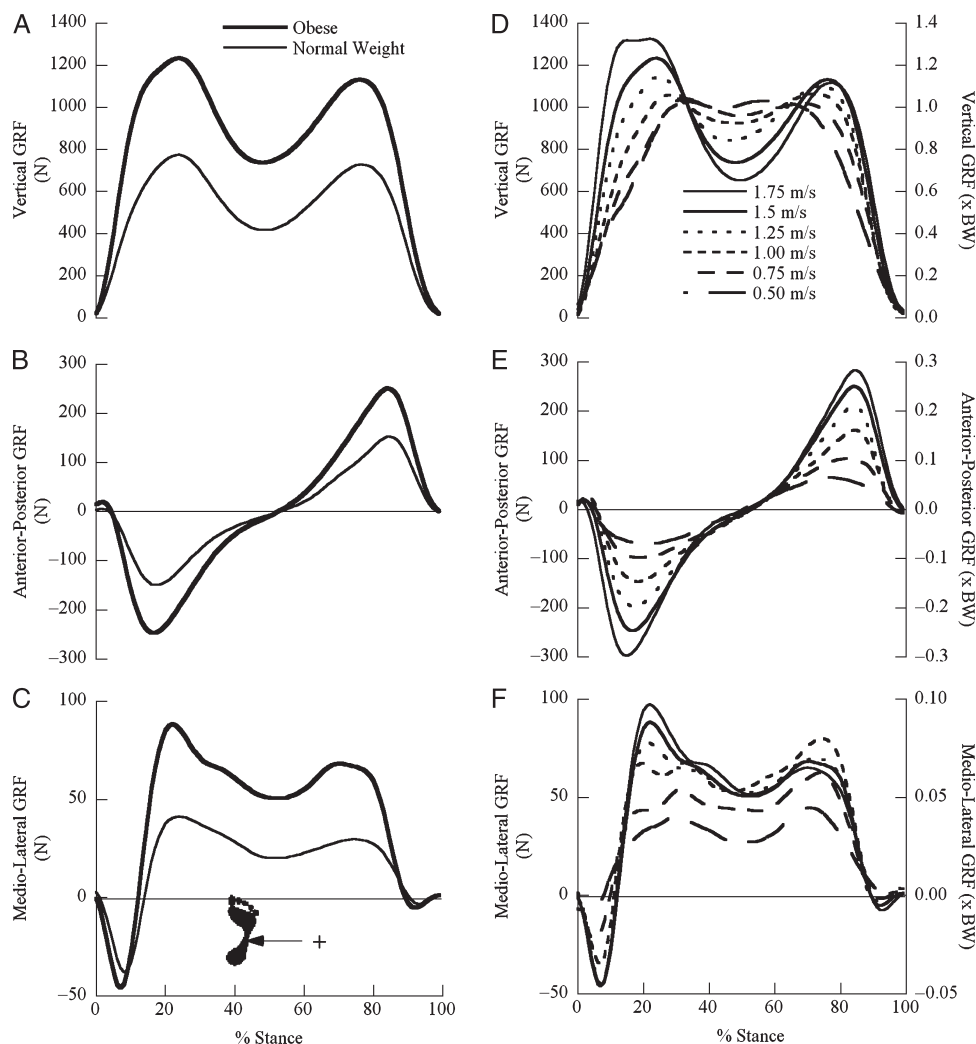


FIGURE 1—Mean vertical (*A*), anteroposterior (*B*), and mediolateral (*C*) ground-reaction forces (GRF) while walking at 1.5 m⁻¹ for the obese vs normal-weight subjects, and vertical (*D*), anteroposterior (*E*), and mediolateral (*F*) GRF for the obese subjects walking at each speed. The cartoon in *panel C* shows the orientation of the mediolateral GRF relative to the foot. Compared with normal-weight subjects, the obese subjects had greater GRF, but in both groups GRF were smaller at the slower walking speeds.

faster walking speeds (1.0–1.75 m⁻¹) and were 71 and 85% greater at 0.75 and 0.50 m⁻¹, respectively. Scaled to body weight, peak AP GRF were similar ($P = 0.80$) between the obese and normal-weight groups at each speed. In both groups, peak AP GRF were approximately 40% smaller at 1.0 versus 1.5 m⁻¹.

Peak ML GRF were about 85% greater ($P < 0.001$) for the obese versus normal-weight subjects across the range of walking speeds. Scaled to body weight, peak ML GRF were

similar ($P = 0.35$) between the obese and normal-weight groups at each speed. Slower walking speeds reduced the first, but not the second, peak of the ML GRF. For example, in the obese subjects, the first and second peaks of the ML GRF were 28% smaller and 6% greater, respectively, when walking at 1.00 versus 1.5 m⁻¹. Peak ML GRF were dramatically smaller at slower walking speeds. For the obese subjects, the first peak of the ML GRF decreased by 23% as walking speed decreased from 1.5 to 1.0 m⁻¹.

TABLE 3. Temporal stride kinematics for obese and normal-weight adults.

Speed (m·s ⁻¹)	Stride Length (m)		Stride Frequency (Hz)		Stance (% Cycle)		Swing (% Cycle)		Double Support (% Cycle)	
	O	NW	O	NW	O	NW	O	NW	O	NW
0.50	0.90 (0.02)	0.87 (0.02)	0.56 (0.01)	0.58 (0.01)	72.0* (0.6)	67.6 (0.9)	28.0* (0.6)	32.4 (0.9)	39.0* (1.2)	32.8 (1.2)
0.75	1.05 (0.02)	1.03 (0.01)	0.72 (0.01)	0.73 (0.01)	69.3* (0.6)	64.8 (0.6)	30.7* (0.6)	35.2 (0.7)	34.5* (1.1)	27.2 (0.7)
1.00	1.21 (0.02)	1.18 (0.01)	0.83 (0.02)	0.85 (0.01)	67.5* (0.5)	63.7 (0.8)	32.5* (0.5)	36.3 (0.8)	32.3* (1.0)	25.3 (0.9)
1.25	1.37 (0.03)	1.35 (0.01)	0.92 (0.02)	0.92 (0.01)	65.6* (0.4)	62.4 (0.6)	34.4* (0.4)	37.6 (0.6)	29.5* (0.8)	23.4 (0.6)
1.50	1.51 (0.03)	1.49 (0.01)	1.00 (0.02)	1.01 (0.01)	64.1* (0.5)	61.5 (0.4)	35.9* (0.5)	38.5 (0.4)	27.3* (0.7)	21.8 (0.5)
1.75	1.63 (0.03)	1.63 (0.02)	1.08 (0.02)	1.07 (0.01)	63.1* (0.6)	60.4 (0.3)	36.9* (0.6)	39.6 (0.3)	25.5* (0.6)	20.0 (0.3)

Values are mean (SE). O, obese; NW, normal weight.

* $P < 0.0083$ obese vs normal weight.

TABLE 4. Midstance joint angles for obese vs normal-weight adults.

Speed (m·s ⁻¹)	Hip Angle (°)		Knee Angle (°)		Ankle Angle (°)	
	O	NW	O	NW	O	NW
0.50	147 (2.9)	150 (3.7)	168 (1.9)	167 (1.9)	107 (2.2)	107 (1.4)
0.75	146 (3.1)	153 (5.0)	167 (1.6)	167 (2.9)	108 (2.2)	107 (1.6)
1.00	145 (3.0)	147 (3.0)	166 (1.7)	162 (1.8)	107 (2.1)	107 (1.7)
1.25	145 (3.2)	147 (3.6)	164 (1.8)	162 (1.5)	107 (2.1)	107 (1.7)
1.50	143 (2.9)	145 (3.4)	162 (1.5)	163 (1.5)	107 (2.0)	107 (2.2)
1.75	141 (3.2)	147 (4.5)	159 (1.5)	160 (1.2)	106 (2.2)	107 (2.2)

Values are mean (SE). O, obese; NW, normal weight. There were no significant differences between groups at any walking speed.

Temporal and angular kinematics. The temporal stride characteristics were similar between the obese and normal-weight groups (Table 3). Specifically, stride length and stride frequency were not different between the groups at any walking speed ($P = 0.50$ and 0.57 , respectively). Compared with the normal-weight adults, the obese adults spent significantly more time in stance and less time in swing, and they had a longer period of double support at all walking speeds. Walking speed had a significant effect on temporal stride variables ($P < 0.001$). For the obese subjects, the duty factor (% stride in stance) increased from 63% at $1.75 \text{ m}\cdot\text{s}^{-1}$ to 72% at $0.50 \text{ m}\cdot\text{s}^{-1}$.

Angular joint kinematics were similar between the obese and normal-weight groups (Table 4, Fig. 2). Midstance hip-, knee-, and ankle-joint angles were not different ($P = 0.41$, 0.61 , and 0.86 , respectively) between the groups across all walking speeds. The hip and knee were generally more extended at the slower walking speeds, whereas the ankle was more plantarflexed during stance and more dorsiflexed during swing.

Sagittal-plane net muscle moments. Absolute net muscle moments (N·m) were significantly greater in the obese versus normal-weight subjects and decreased significantly at slower walking speeds in both groups (Table 5, Fig. 3). Peak hip extensor, knee extensor, and ankle plantarflexor muscle moments were greater ($P = 0.003$, 0.025 , and 0.026 , respectively) in the obese versus normal-weight subjects. *Post hoc* comparisons revealed that the peak hip extensor moment was significantly greater at all speeds except $0.75 \text{ m}\cdot\text{s}^{-1}$, whereas peak extensor knee moments were significantly greater in the obese subjects at $1.75 \text{ m}\cdot\text{s}^{-1}$. While walking at $1.50 \text{ m}\cdot\text{s}^{-1}$, the peak knee extensor moment was 51% greater ($P = 0.009$) for the obese versus normal-weight subjects. Comparing walking at 1.00 and $1.50 \text{ m}\cdot\text{s}^{-1}$ in the obese subjects, the peak knee extension moment was 43% smaller, whereas the peak ankle plantarflexor moment was only 9% smaller. When scaled to body mass ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$), peak hip and knee extensor moments were not different ($P = 0.394$ and 0.36 , respectively) between the groups at each walking speed. The scaled peak ankle plantarflexor moment was, however, significantly smaller ($P = 0.02$) for the obese versus normal-weight subjects at each walking speed.

Step width. Step width was about 30% greater ($P = 0.008$) in the obese versus normal-weight subjects (Fig. 4)

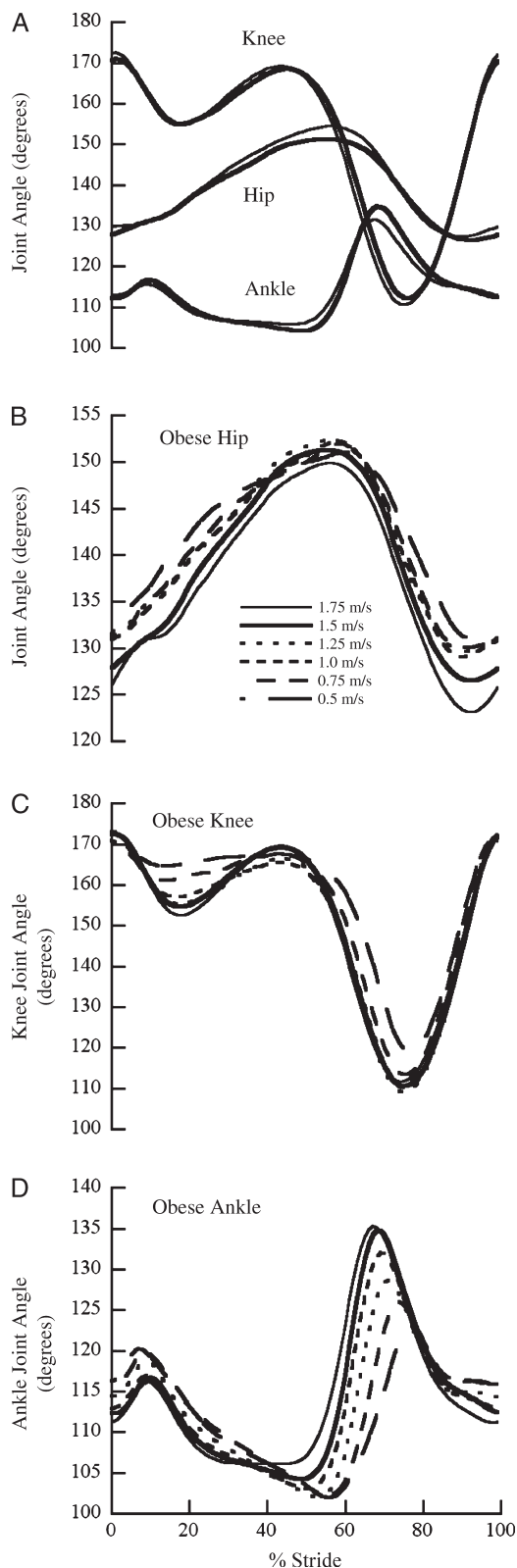


FIGURE 2—Mean joint angles for obese (thick line) vs normal-weight (thin line) subjects while walking at $1.5 \text{ m}\cdot\text{s}^{-1}$ (A), and hip- (B), knee- (C), and ankle-joint (D) angles for the obese subjects walking at each speed. Percent stride is from right heel strike to right heel strike. Joint angles were similar between obese and normal-weight subjects across walking speeds. Slower walking speeds resulted in reduced hip and knee range of motion during the stance phase of walking, whereas ankle range of motion was greater at the slower walking speeds.

TABLE 5. Peak sagittal-plane net muscle moments for obese vs normal-weight adults.

Speed (m·s ⁻¹)	Hip (N·m)		Knee (N·m)		Knee (N·m·kg ⁻¹)		Ankle (N·m)	
	O	NW	O	NW	O	NW	O	NW
0.50	39.0* (3.82)	23.3 (3.44)	42.1 (9.17)	28.1 (4.42)	0.38 (0.073)	0.43 (0.067)	86.0 (7.76)	63.9 (4.85)
0.75	60.1 (7.55)	37.3 (3.63)	47.4 (10.69)	38.7 (6.06)	0.43 (0.079)	0.57 (0.096)	103.4 (8.91)	75.2 (6.11)
1.00	83.5* (8.32)	47.3 (4.70)	64.5 (7.51)	50.1 (6.05)	0.58 (0.054)	0.73 (0.086)	119.4 (10.41)	78.2 (7.36)
1.25	113.0* (11.74)	64.8 (4.88)	84.6 (8.83)	62.6 (6.81)	0.77 (0.054)	0.90 (0.081)	125.5 (11.27)	88.2 (8.98)
1.50	139.6* (13.06)	81.3 (6.16)	114.0 (9.95)	75.7 (9.16)	1.02 (0.061)	1.08 (0.090)	130.2 (12.15)	85.8 (8.03)
1.75	175.7* (19.65)	99.6 (6.87)	143.1* (11.19)	96.3 (10.93)	1.28 (0.063)	1.34 (0.097)	129.5 (13.35)	95.0 (9.79)

Values are mean (SE). O, obese; NW, normal weight.

* $P < 0.0083$, obese vs normal weight.

and did not change significantly with walking speed ($P = 0.48$).

DISCUSSION

We purposefully elected to emphasize the absolute rather than normalized GRF and joint moments. We believe that

although normalizing these variables to body mass allows relative comparisons across groups, it distracts attention from the actual loads placed on the joints. Support for our rationale is the recent finding of Ding et al. (9) that joint articulating surface area does not scale with body mass. In that study, body mass was 48% greater in the obese subjects versus normal-weight subjects, whereas tibial articulating

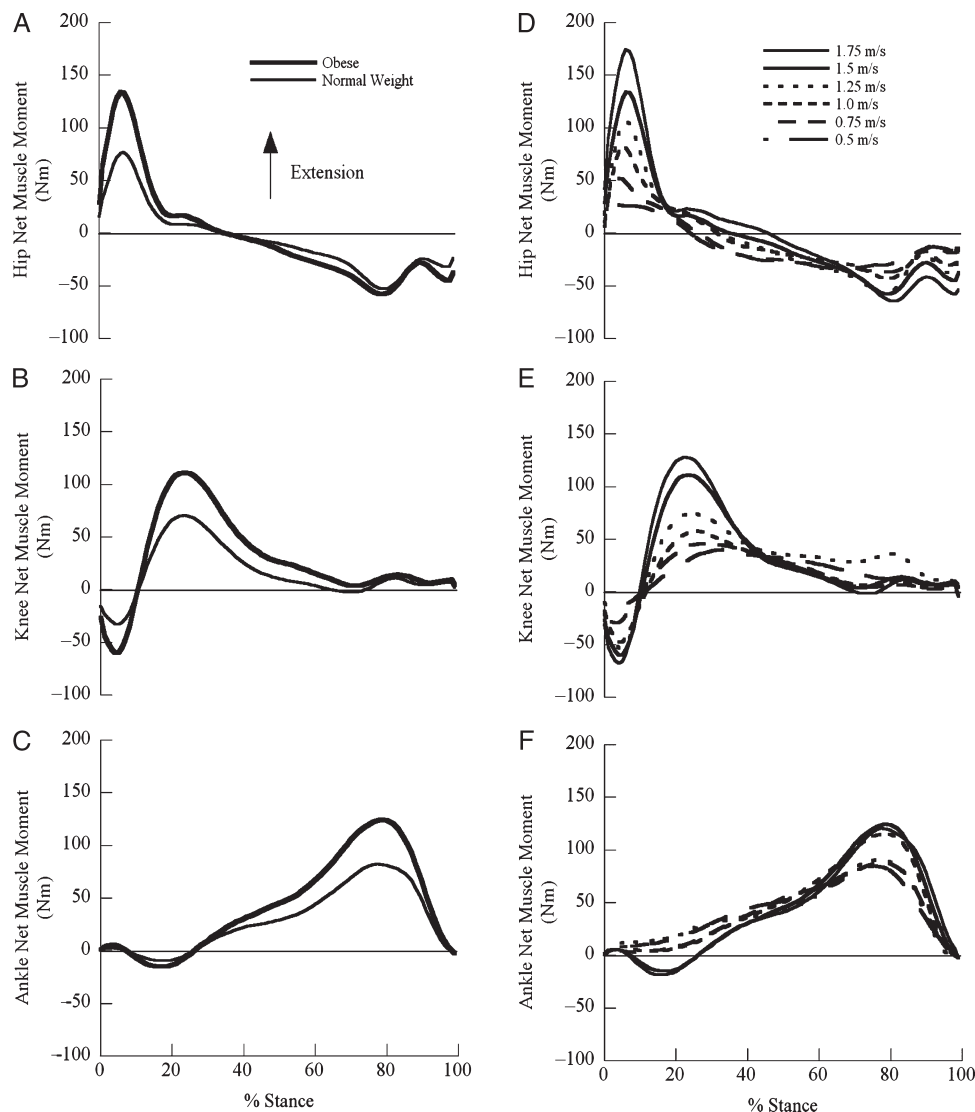


FIGURE 3—Mean hip (A), knee (B), and ankle (C) net muscle moments during stance while walking at 1.5 m·s⁻¹ for the obese vs normal-weight subjects, and hip (D), knee (E), and ankle (F) net muscle moments during stance for the obese subjects walking at each speed. Positive moments are extensor. Stance begins at right heel strike. Compared with the normal-weight subjects, the obese subjects had greater net muscle moments, but in both groups net muscle moments were smaller at the slower walking speeds.

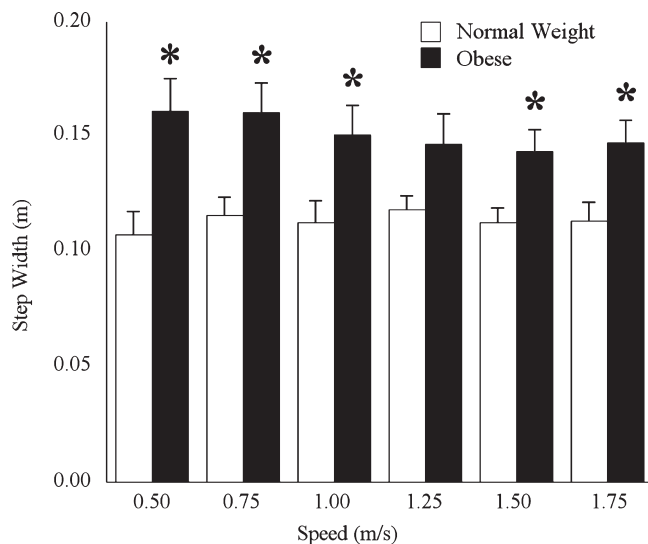


FIGURE 4—Mean step width at midstance for the obese vs normal-weight subjects. Step width was significantly greater for the obese subjects at all speeds except $1.25 \text{ m}\cdot\text{s}^{-1}$ and was similar across walking speeds. * $P < 0.0083$ for obese vs normal-weight subjects at each speed.

surface area was only 8% larger. This would suggest that obese adults would have increased mechanical loads (e.g., stress) across the tibial articulating surfaces. In addition, given the relatively small temporal differences between the groups, obese adults experience a greater rate of joint loading. Because the development of OA has been associated with increased magnitude and rate of mechanical loading (14,29,31,32), absolute force seems a logical comparison.

GRF. As we hypothesized, absolute GRF were much greater in the obese versus normal-weight subjects and were substantially reduced by walking slower. Our vertical and AP GRF results are similar to those of Messier et al. (25). They report that peak vertical and AP GRF were 1.03 and $0.14 \times \text{body weight (BW)}$, respectively, when arthritic obese subjects walked at $1.06 \text{ m}\cdot\text{s}^{-1}$. Interpolating our results to $1.06 \text{ m}\cdot\text{s}^{-1}$, we estimate peak vertical and AP GRF magnitudes to be approximately 1.02 and $0.15 \times \text{BW}$, respectively, in our obese subjects. Messier et al. also report peak medial GRF of $0.05 \times \text{BW}$, which is somewhat smaller than the approximately $0.07 \times \text{BW}$ recorded when our obese subjects walked at $1.00 \text{ m}\cdot\text{s}^{-1}$. Because the obese subjects in the Messier et al. study had knee OA, they may have altered their gait to reduce the medial GRF during early stance.

The larger GRF of the obese versus normal-weight subjects reflect the much greater body mass of the obese adults. Whereas the vertical and AP GRF increased in nearly direct proportion to body mass, the ML GRF were greater in the obese than would have been predicted according to body mass. The body mass of the obese subjects was 61% greater than normal-weight subjects, whereas the first medial peak of the ML GRF was 91% greater when walking at $1.5 \text{ m}\cdot\text{s}^{-1}$. The disproportionate

increase in ML GRF is likely attributable to the greater step width used by the obese subjects. There was a significant positive correlation between step width and the absolute peak ML GRF ($r^2 = 0.45$, $P < 0.001$). Spyropoulos et al. (35) reported a mean step width of 0.16 m when obese males (129 kg body mass) walked at $1.09 \text{ m}\cdot\text{s}^{-1}$, a finding similar to the step width of 0.15 m adopted by our obese subjects at a similar speed. Further, Donelan et al. (10) reported that ML GRF were greater when normal-weight adults walked using wider step widths. A greater step width may also increase the muscle forces required to control the degree of pronation during early stance. Pronation has been reported to be greater and problematic in obese versus normal-weight adults (24).

Slower walking speeds resulted in considerably smaller GRF magnitudes. In particular, the AP and first medial peak of the ML GRF were much smaller at the slower walking speeds. Yet, the second medial peak of the MG GRF was not smaller at the slower walking speeds. This may also be related to the greater step width used by the obese subjects. The second peak of the ML GRF acts to redirect the body weight onto the contralateral stance leg. We would not expect this force to be reduced at slower walking speeds, given that step width was not different at slower speeds.

Sagittal-plane net muscle moments. We accept our hypotheses that sagittal-plane net muscle moments are greater in obese versus normal-weight adults and are reduced by walking more slowly. The greater hip, knee, and ankle net muscle moments of the obese versus normal-weight subjects were attributable to the combination of greater GRF and similar joint kinematics. Our findings suggest that the obese subjects were walking with greater joint loads and that walking slower can reduce these loads.

The greater sagittal-plane knee net muscle moment of the obese versus normal-weight subjects is not consistent with the findings of DeVita and Hortobagyi (8). The discrepancy in the findings of DeVita and Hortobagyi and our study are a result of differences in the kinematics of the knee joints of obese subjects. The normal-weight subject's joint moment values of our study are consistent with other published data (40). Baliunas et al. (2) have reported a significant correlation ($r^2 = 0.66$) between the peak extensor knee moment and peak knee-joint flexion during midstance. DeVita and Hortobagyi reported that obese adults walked with less knee flexion during stance, but we detected no significant difference in knee-joint angles between obese and normal-weight subjects. Our finding that moderately obese adults walked with similar joint kinematics despite their greater body mass is also consistent with reports of unchanged lower-extremity kinematics when individuals walk with added mass on the waist or back (16,37). This discrepancy in joint kinematics and moments between our study and that of DeVita and Hortobagyi may be attributable to differences in body mass and BMI of the two populations. Approximately 50% of the obese subjects in the DeVita and Hortobagyi study had a BMI value > 40

$\text{kg}\cdot\text{m}^{-2}$, and it seems that these subjects had the lowest average absolute knee net muscle moments (N·m) during stance (see DeVita and Hortobagyi, Fig. 4A). Thus, it may be that as the degree of adiposity increases above a critical level (i.e., $\text{BMI} > 40 \text{ kg}\cdot\text{m}^{-2}$), individuals adjust their gait to reduce knee-joint loads. Additional factors might contribute to altered gait kinematics in obese adults, including the length of time an individual has been obese, lower-extremity musculoskeletal strength, and development of early-stage OA.

The sagittal-plane net muscle moments at the knee were dramatically smaller when obese subjects walked at slower speeds. Our recent study (5) reports that young moderately obese adults prefer to walk at about $1.4 \text{ m}\cdot\text{s}^{-1}$. If these individuals reduced walking speed to $1.0 \text{ m}\cdot\text{s}^{-1}$, the knee net muscle moment would be about 40% smaller. One of the primary determinants of the peak extensor net muscle moment is the AP GRF; when obese subjects walked at 1.00 versus $1.50 \text{ m}\cdot\text{s}^{-1}$, the peak AP GRF was 40% smaller. This reduction in AP force is consistent with the adoption of a more extended leg during stance at slower speeds, which would reduce the muscle forces required at the knee (4).

External knee adduction moment. Viewed in the frontal plane during normal walking, the resultant GRF passes medially to the center of the knee and causes an external adduction moment at the knee. This external knee adduction moment plays an important role in the distribution of the total knee-joint load across the tibiofemoral compartment (1) and is positively correlated with OA disease severity and progression (26). A greater external knee adduction moment would concentrate the compressive load on the medial compartment of the knee. To our knowledge, external knee adduction moments in obese individuals during gait have not been previously reported.

We calculated the peak external knee adduction moment attributable to the ML GRF, to gain insight into the distribution of joint loads across the ML axis of the knee. To determine this peak external knee adduction moment, we used the first peak of the ML GRF and the corresponding moment arm distance from the knee-joint marker to the treadmill belt surface. The peak external knee adduction moment increased with walking speed and was significantly greater in the obese versus normal-weight subjects (Fig. 5). Although this finding would suggest that obese adults have a greater compressive load on the medial compartment of the knee joint, we could not determine the contribution of the vertical GRF to the external adduction moment. Obese adults may walk with an externally rotated leg (35) or may have a valgus knee alignment, which acts to reduce the external knee adduction moment (19). Thus, we do not know whether medial knee-joint loads are greater in obese adults or whether these individuals adopt a gait pattern that reduces these potentially injurious loads. Future studies that include three-dimensional biomechanical analysis and estimates of knee-joint loads using musculoskeletal models during walking in obese adults are clearly needed.

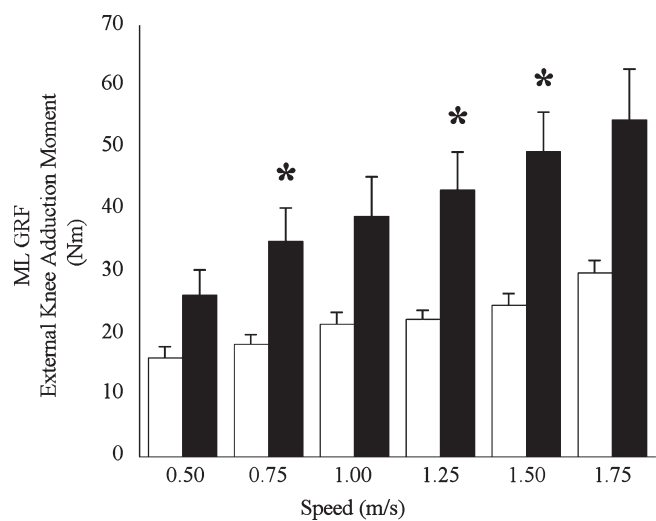


FIGURE 5—Peak external knee adduction moment attributable to the ML GRF for the obese vs normal-weight subjects. Peak knee adduction moment was significantly greater for the obese subjects. * $P < 0.0083$ for obese vs normal-weight subjects at each speed.

Relevance for exercise prescription. The results of this study can be used to estimate a “biomechanically equivalent” walking speed where the joint loads (using net muscle moments as a proxy) are similar between normal-weight and obese adults. Given that preferred walking speed in normal-weight adults is about $1.4 \text{ m}\cdot\text{s}^{-1}$ (33), our data suggest that obese adults would need to walk at about $1.1 \text{ m}\cdot\text{s}^{-1}$ to have the equivalent sagittal-plane knee-joint loads as normal-weight adults walking at $1.4 \text{ m}\cdot\text{s}^{-1}$. Adopting a slower walking speed may also have energy-expenditure benefits, because slower walking speeds expend more energy per unit distance than walking at a normal speed (6,33). Although walking at a slower speed may be impractical for some individuals who walk as a form of transportation, and it may also reduce stability, walking at $1.1 \text{ m}\cdot\text{s}^{-1}$ is not typically perceived as uncomfortably slow, and it also reduces the aerobic demands and perceived exertion of the task.

Limitations. There are several limitations to this study that should be noted. The two-dimensional nature of this study does not account for rotation of the limb. If the leg is externally rotated during walking, a two-dimensional analysis may underestimate early-stance anatomical-axis sagittal-plane moments and overestimate early-stance external knee adduction moment attributable to the M-L GRF. Although it is possible that the obese subjects walked with a more externally rotated leg during stance, the differences between the obese and normal-weight groups are likely to be small, especially in our young, otherwise healthy subjects. Spyropoulos et al. (35) report that obese males walked with essentially the same hip external rotation at heel strike compared with normal-weight males (mean difference of 0.2°). Using the center of pressure to determine step width may not account for variations in foot morphology. Although there may be differences in foot

morphology between obese and normal-weight individuals (24), they are not likely to account for the relatively large differences in step width reported here. In addition, we did not obtain radiographs to rule out the possibility of knee OA in our obese subjects. However, our subjects were young and reported no musculoskeletal problems.

Caution may be warranted when translating treadmill data to overground walking. Some studies that have compared treadmill and overground walking mechanics have reported small but statistically significant differences in temporal, kinematic, and kinetic measures (36,39). However, a recent study by Riley et al. (34) reported that although there were differences in kinematic and kinetic parameters during treadmill versus overground walking, they were small and within the range of repeatability of the measures. The degree of familiarization with treadmill locomotion can also affect the reported differences between treadmill and overground walking. After a familiarization period of at least 4 min, the kinematics of walking have been reported to be similar in treadmill versus overground walking (22,27). The subjects in our study walked for at least 10 min before data were collected. According to the first principles of physics, locomotion on a motorized treadmill with adequate momentum and power should be identical to overground walking or running (38). However, perceptual differences, trepidation, and unfamiliarity may

cause people to alter their gait. We feel that the large biomechanical differences we report here between normal-weight and obese subjects are of a much greater magnitude than the biomechanical differences sometimes observed for normal-weight subjects walking on treadmills versus overground. It is possible that, compared with normal-weight subjects, obese subjects may have much greater differences in their gait on treadmills versus overground, but we are not aware of any studies on the topic.

SUMMARY

In conclusion, obesity greatly increases GRF during walking, without changes in lower-extremity sagittal-plane kinematics. As a result, sagittal-plane net muscle moments (and, by proxy, joint loads) at the hip, knee, and ankle are also greater for obese versus normal-weight adults. At slower walking speeds, GRF and net muscle moments are all significantly smaller. These results suggest that slower walking may reduce the risk of musculoskeletal pathology in obese adults.

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