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Effects of adding weight to the torso on roll-over characteristics of walking

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Abstract—Ten participants without physical impairment walked with 0 kg, 11.5 kg, and 23.0 kg of added weight equally distributed about the torso in a harness. At each weight level, the participants walked at slow, normal, and fast self-selected walking speeds. We examined the roll-over characteristics by determining the ankle-foot and knee-ankle-foot roll-over shapes. These shapes, which are the effective rockers created by the respective lower-limb systems between heel contact and opposite heel contact of walking, are found if one transforms the center of pressure of ground reaction force into body coordinate systems. The roll-over shapes of the ankle-foot and knee-ankle-foot systems did not change appreciably with added weight at any of the three walking speeds. The invariance of these biologic systems to added weight should be considered when prostheses and orthoses are designed that intend to replace and augment their function in walking.

Key words: ankle, foot, gait, human movement, knee, orthotics, prosthetics, rockers, roll-over shapes, weighted walking.

INTRODUCTION

Lower-limb prosthetic devices (e.g., prosthetic feet) are generally chosen on the basis of weight and activity level of the intended users. However, apart from anecdotal clinical observation, no information is present to suggest how a given prosthetic foot should function for persons of different body mass. Furthermore, no data exist suggesting whether or not a prosthetic foot should adapt when body weight changes or when the user carries different loads. This paper examines the roll-over characteristics of nondisabled persons' lower-limb systems when they walk with added weight. The findings may prove useful for the design of rehabilitation devices such as prostheses and orthoses and may suggest properties inherent in some of the present devices that help to achieve biomimetic responses to added weights.

Muybridge was an early investigator of persons walking with added weights [1]. He published a series of photographs of a man carrying a 75 lb boulder on his head and a woman carrying a basket on her head. No gross differences appear to be present in the lower-limb kinematics of their gait with this weight compared with normal walking. Other studies have examined the energy cost

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Abbreviations: ANOVA = analysis of variance, BCOM = body center of mass, KAF = knee-ankle-foot, SD = standard deviation, VA = Department of Veterans Affairs, VACMARL = VA Chicago Motion Analysis Research Laboratory.

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of carrying weights and how this energy cost changes when the weights are carried at different locations on the body [2–7]. Lind and McNicol showed that persons use more energy and fatigue faster when carrying weights by hand as opposed to carrying the weights using a shoulder harness [2].

Harman et al. examined gait kinematics and gait kinetics of 16 nondisabled male research participants who carried various loads in a backpack [8]. The ankle, knee, and hip kinematic curves from Harman et al.'s report indicate that as load was increased, the gait kinematics were not appreciably altered. However, the ankle, knee, and hip torques did change appreciably as the load was increased. Frigo et al. also found increased lower-limb joint torques when loads were increased for children but did not report kinematic changes of the ankle, knee, or hip joints [9]. The results of these studies are consistent with the results of Selles, who found that a kinematic invariance strategy was likely used by persons using transtibial prostheses when weights were added to their prostheses (as opposed to a kinetic invariance strategy) [10].

Rocker models of walking have the potential to provide a more thorough understanding of nondisabled as well as disabled walking [11–14]. The recent measurement of roll-over shapes, which are the effective rockers created by various lower-limb systems during walking, may help to show how nondisabled persons adapt to changes in various walking conditions such as speed, carrying added weights, and walking with shoes of different heel heights [15–19]. This information could be useful in the design of rehabilitation devices such as lower-limb prostheses and orthoses that can adapt in ways that more closely mimic biology. The utility of the roll-over shape stems from its simplicity and its direct connection to body-based reference frames. The roll-over shape of the able-bodied ankle-foot system, for example, can be considered a goal for mechanical properties of an ankle-foot prosthesis during walking. The roll-over shape, by means of one simple geometry, indicates a history of end points of deflection of the prosthesis at various angles of loading throughout the roll-over phase of walking (**Figure 1**). The use of roll-over shape as a possible tool for aligning prostheses has also been demonstrated [20].

Previous work of others strongly suggests invariance of roll-over shape when persons carry added weights during walking. The rocker model of Gard and Childress suggests that the vertical excursion of the body center of mass (BCOM) can be described by a single and constant

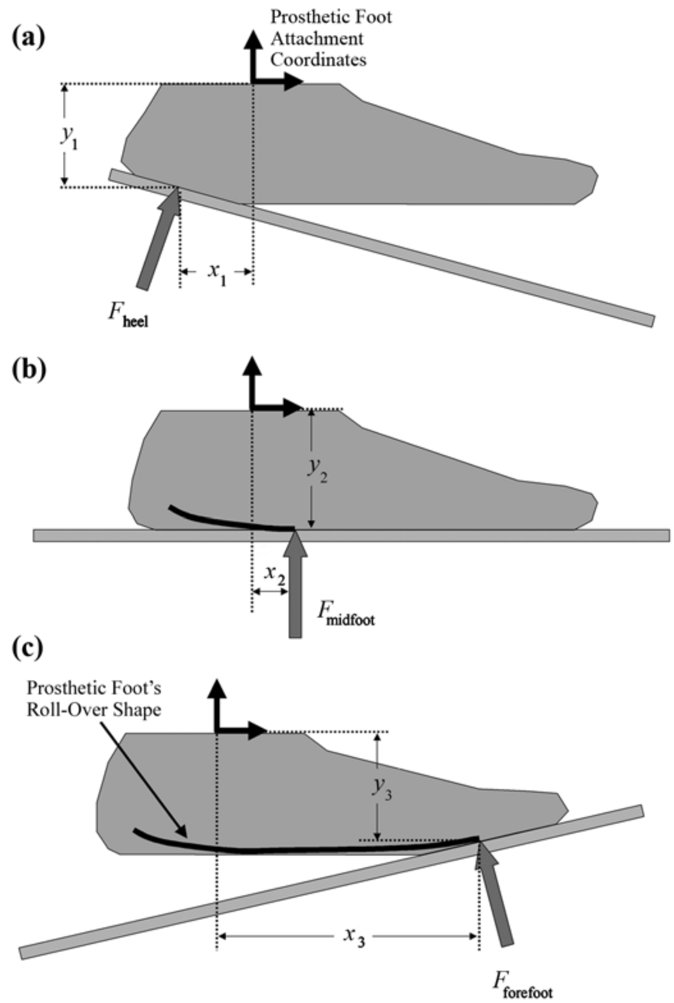

Figure 1.
Three drawings that illustrate roll-over shape concept for prosthetic feet. Loading force **(a)** just after heel contact, **(b)** near midstance, and **(c)** just before the time of opposite heel contact. Roll-over shape (defined by *x*- and *y*-points) maps history of center of pressure (COP) location in sagittal plane, indicating end points of deflection of foot when it is loaded at various angles of roll-over. Characteristics of this shape, such as radius and arc length, can be used to evaluate function of feet for walking. For example, a radius that is too small could indicate a foot that deflects excessively during walking. Also, feet with short arc lengths may lead to abrupt ends of roll-over and could cause drop-off effects [*Source*: Hansen A, Sam M, Childress D. The effective foot length ratio (EFLR): A potential tool for characterization and evaluation of prosthetic feet. J Prosthet Orthot. 2004;16(2):41–45.]. Because of fixed ankles in most prostheses, we believe feet should be designed so that their deformations result in roll-over shapes that mimic that of the able-bodied ankle-foot complex. F = forces during stance phase; x_1, x_2, and x_3 and y_1, y_2, and y_3 = horizontal and vertical positions, respectively, of COP in prosthetic foot attachment coordinates at three times during stance phase (points on roll-over shape of prosthetic foot).

radius rocker-based inverted-pendulum model with increases in BCOM excursion during walking mainly due to increases in step length [14]. Also, the work of Holt et al. has shown that the BCOM vertical excursion is not affected by added weight [21], suggesting that the rocker in Gard and Childress' model should not change with increases in weight. Thus, roll-over shape should be constant when persons walk with added weight because the roll-over shape is closely related to the rocker in Gard and Childress' model. Although evidence strongly suggests invariance of roll-over shape to added weight, we aimed to examine this hypothesis directly by experiment. In addition, the parameters of the roll-over shapes can be used to assist in the design of prosthetic devices; i.e., prosthetic devices can be designed to conform to a certain radius under walking loads [22].

In this study, we examined the effects of carrying symmetric torso loads on the roll-over characteristics, i.e., roll-over shapes of the ankle-foot and knee-ankle-foot systems, of 10 nondisabled research participants. The hypothesis of the study was that these roll-over shapes would not change appreciably with increased loads.

METHODS

We recruited 10 persons without physical impairment (five male and five female) and, through a process approved by the institutional review board, obtained their informed consent to participate in the added-weight experiment (see subject-specific data in **Table 1**). We measured and recorded each subject's height, weight, and foot length.

Table 1.
Subject-specific data for participants in added-weight study.

Subject Identifier	Gender	Weight (kg)	Height (cm)	Age (yr)	Foot Length (cm)
1	F	51.1	159.0	29	25.3
2	F	62.0	162.0	33	25.0
3	F	66.8	168.5	27	27.0
4	F	65.0	168.5	34	26.2
5	F	61.5	160.0	25	26.0
6	M	78.4	177.0	26	28.7
7	M	82.5	181.5	22	32.0
8	M	122.5	183.0	32	32.0
9	M	93.5	184.0	35	30.5
10	M	100.9	196.0	23	32.0

F = female, M = male

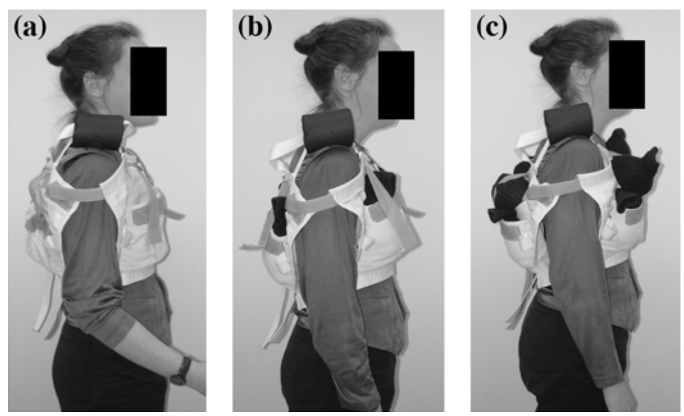

Figure 2.
Harness used in experiment. Harness was used **(a)** first without weights, **(b)** then with 11.5 kg of added weight, and **(c)** last with 23.0 kg of added weight. Weights were symmetrically distributed about body with equal-weight bags in four pouches around trunk.

To avoid subjects' fatigue during the walking trials, we fitted them with an adjustable shoulder harness that was designed to hold the added weights (**Figure 2**) as opposed to having subjects carry weights by their arms. Each subject first walked with no weights in the harness, then with 11.5 kg (25 lb) of added weight in the harness, and last with 23.0 kg (50 lb) of added weight in the harness. We added the weights using eight bags filled with lead shot, each at an approximate weight of 2.9 kg (6.25 lb). The bags were positioned inside four pockets of the harness that were symmetrically placed around the trunk of the body. To become accustomed to the added load, subjects were allowed 1 to 2 min with each condition. At each weight level, the subject walked at three different self-selected walking speeds: slow, normal, and fast. At each added-weight/walking-speed combination, trials were taken until five "clean" force-platform hits were achieved on the left and on the right feet. "Clean" force-platform hits were ones in which only one foot contacted a force platform, without stepping over the edges of the platform. Trials were not randomized for reasons of convenience in the protocol.

Gait-analysis measurements were made at the Department of Veterans Affairs (VA) Chicago Motion Analysis Research Laboratory (VACMARL). VACMARL is equipped with an eight-camera motion analysis system* and six force platforms.† We captured motion at 120 Hz and sampled the force platforms at 960 Hz. Later, we resampled the force data at 120 Hz (i.e., every eighth sample was used) to synchronize them with the motion capture data. A Helen Hayes marker set was used for the gait data collection [23], although the only necessary markers for finding the three types of roll-over shapes were the markers on the lateral malleoli (ankle), the femoral condyles (knee), the right and left anterior superior iliac spines, and the sacrum. Virtual hip markers (hip) were estimated with Vaughan et al.'s method [24].

Following the data collection, ankle-foot and knee-ankle-foot roll-over shapes were computed and fitted with circular parameters as described previously [17–18]. We found the ankle-foot roll-over shape by transforming the center of pressure in the direction of forward progression from a laboratory coordinate system into an ankle-knee coordinate system. We found the *y*-axis of the ankle-knee coordinate system by drawing a vector from the ankle marker to the knee marker. The *x*-axis of the ankle-knee coordinate system was found as the perpendicular vector to the *y*-axis that was in the plane of forward progression (laboratory "sagittal" coordinates) and that pointed "forward." We found the knee-ankle-foot roll-over shape by transforming the center of pressure in the direction of forward progression from the laboratory coordinates into ankle-hip coordinates. The ankle-hip coordinate system was found exactly as the ankle-knee coordinate system, except that the hip marker was used instead of the knee marker.

The transformations of the center of pressure of the ground reaction force into the body-based coordinate systems give the pathways of "where the force acted," both anteriorly and vertically, for the shank and for the entire lower limb. These pathways can be thought of as effective rockers, or roll-over shapes, of the ankle-foot and the knee-ankle-foot systems [18].

We determined the best-fit circular arc for each knee-ankle-foot roll-over shape using a nonlinear least-squares algorithm (steepest descent). We used the solution of the second-order linear Taylor series expansion of the equation of the lower half of a circle to find the starting parameters for the algorithm. The circular equation was expanded about the roll-over shape's median *x* value (where "*x*" refers to the horizontal component of the roll-over shape). The parameters found in the circular-fitting algorithm included radius and the forward position of the center of the circle ("XARC") with respect to the ankle marker. The arc length was calculated as the length of the best-fit circular arc that would extend to both the minimum and maximum *x*-values of the roll-over shape data. (**Figure 3** shows the three circular arc parameters that were found.) For comparison between subjects, all parameters found from circular fitting were normalized by the body height of the subject.

After circular fitting, we normalized roll-over shapes by height and by the time needed to create the shape (time between heel contact and opposite heel contact). We then set these time-normalized shapes into equal-length arrays with a cubic spline routine to allow averaging of shapes for similar conditions (i.e., same walking speed and added-weight trials). We also determined the variability of the roll-over shapes by finding the standard deviations (SDs) for *x* and for *y* (where "*y*" refers to the vertical component of the roll-over shape) for each indexed data point in the equal-length arrays. We used the mean and SDs of *x* and *y* to extract the mean roll-over shape and SD of error ellipses at each point. The error ellipse at each point had a width and height equal to one SD in *x* and *y*, respectively.

For each condition of walking (e.g., walking fast with half the weight added), we found the medians of the roll-over shape parameters. The medians of each parameter for each walking condition and for each subject were used in a 3×3 two-way repeated-measures analysis of variance (ANOVA). We used this statistical test to determine if added weight, walking speed, and/or the interaction between added weight and walking speed were factors that would affect the various parameters of the roll-over shapes. We examined the assumption of sphericity with Mauchly's Test of Sphericity. If the assumption of sphericity was violated, we used the Greenhouse-Geisser correction factor [25]. We administered post hoc tests, using the Bonferroni adjustment for multiple comparisons, for factors that were significantly changed at the $p < 0.05$ level. Statistical tests were calculated with a statistical software package.‡

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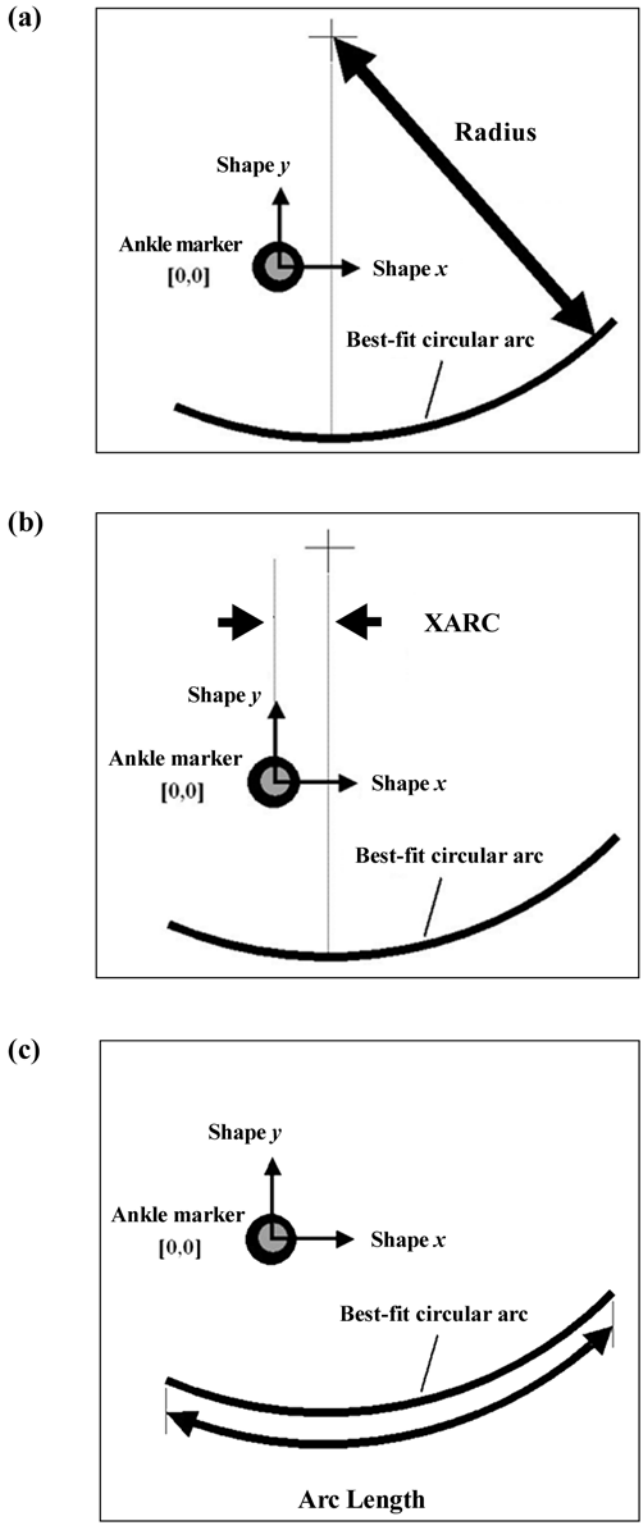

Figure 3.
Parameters of a best-fit circular arc: **(a)** radius, **(b)** forward shift of arc, and **(c)** arc length. *x* and *y* = coordinate axes for roll-over shape, [0,0] = coordinate axis origin, and XARC = forward shift of arc.

RESULTS

Mean ± SD walking speeds that subjects used in the experiment were 0.93 ± 0.23 m/s for slow walking, 1.33 ± 0.17 m/s for normal walking, and 1.76 ± 0.21 m/s for fast walking.

At normal walking speeds, the ankle-foot and knee-ankle-foot roll-over shapes do not appear to change appreciably with increases in added weight to the torso (**Figure 4**). The invariance of the roll-over shapes to added loads to the torso becomes more apparent when the roll-over shapes are plotted on the same axes (**Figure 5**). Slight changes appear to exist in the shapes as walking speed is increased, particularly with the ankle-foot roll-over shapes (**Figure 5**).

Examining the best-fit circular-arc parameters of the knee-ankle-foot roll-over shapes strengthens the hypothesis that roll-over shapes are invariant to added weight because the parameters appear constant over the range of weights carried (**Figure 6**). Each data point shown in **Figure 6** is the median value of the parameter for each subject's walking trials at each speed-weight combination.

The results of the statistical tests indicated that none of the three roll-over shape parameters were significantly changed with added weight ($p = 0.81$ for radius, $p = 0.89$ for XARC, and $p = 0.21$ for arc length) or the interaction

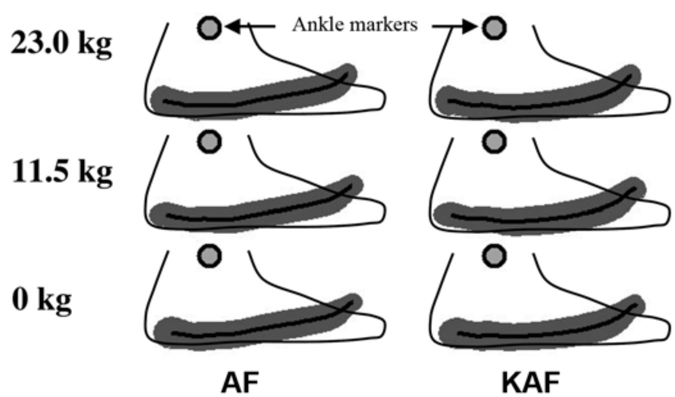

Figure 4.
Average ankle-foot (AF) and knee-ankle-foot (KAF) roll-over shapes (average of subject averages) and standard deviation envelopes associated with shapes. Gray-filled circles indicate origins of ankle-knee and ankle-hip coordinate systems for average shapes (each shape appears directly beneath corresponding origin). Shapes do not appear to change with increased weight (from bottom to top of figure). Shapes here are for normal walking speeds only (mean speed = 1.33 m/s). Sagittal outlines of foot are sketched to put roll-over shapes in physiologic context.

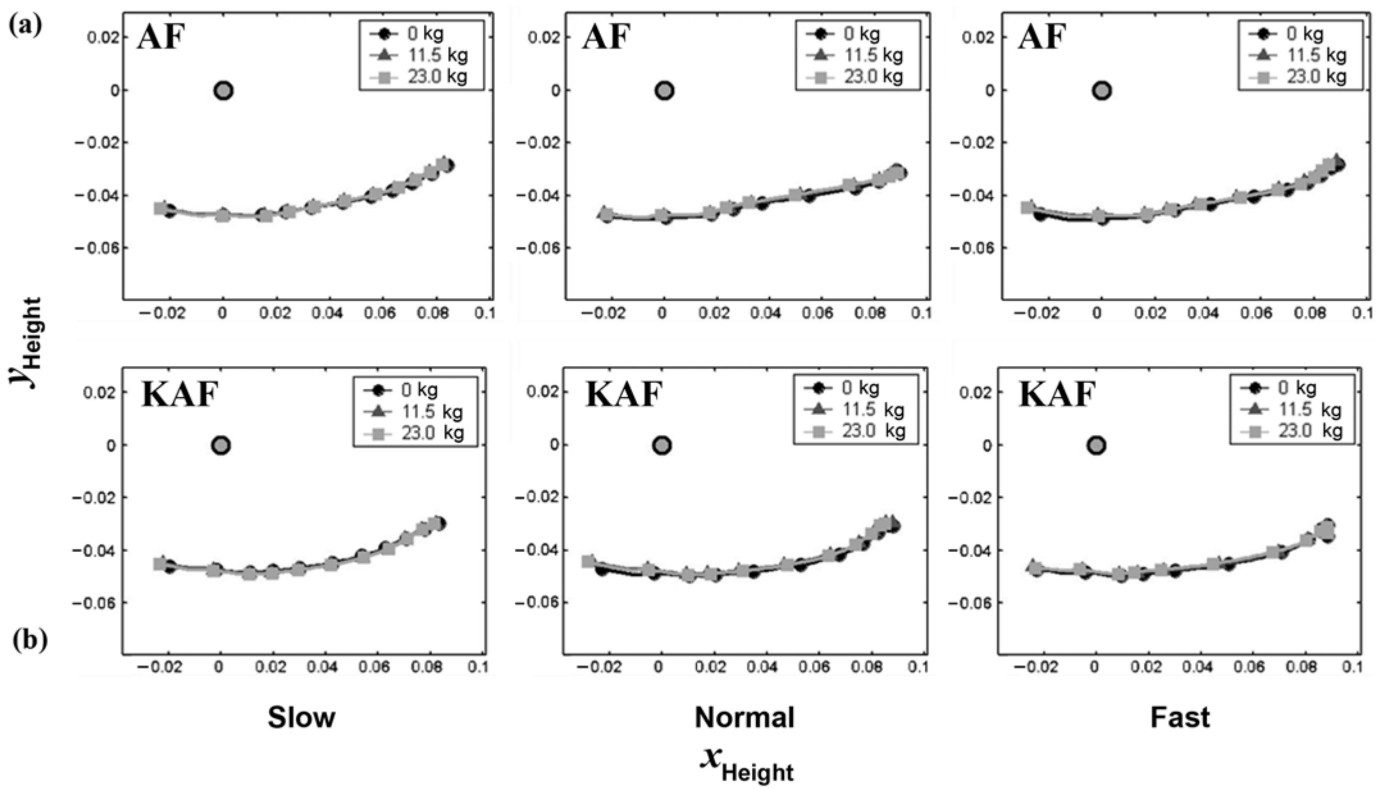

Figure 5.
(a) Average ankle-foot (AF) and **(b)** knee-ankle-foot (KAF) roll-over shapes for weighted-walking study. Plots are normalized by participant height and shown for three walking speeds. Added-weight conditions are indicated with three different symbols and different shades of gray. Roll-over shapes are indicated by black circles when no added weight is carried, by darker gray triangles when 11.5 kg of added weight is carried, and by light gray squares when 23.0 kg of added weight is carried. Weight was always attached to torso as shown in **Figure 2**. Notice that shapes do not change appreciably with added weight. AF roll-over shapes appear to change slightly with increased walking speeds. KAF roll-over shapes do not appear to change appreciably with increased walking speed.

between added weight and the walking speed ($p = 0.34$ for radius, $p = 0.23$ for XARC, and $p = 1.00$ for arc length). All three roll-over shape parameters had significant main effects due to walking speed ($p < 0.01$ for radius, $p < 0.01$ for XARC, and $p = 0.01$ for arc length). However, regarding walking speed, post hoc tests indicated that although the radii and forward shifts (XARC) were not significantly different between slow and normal walking speeds ($p = 1.00$ for radius; $p = 0.20$ for XARC), both parameters were significantly different when we compared slow-to-fast ($p = 0.03$ for radius; $p = 0.02$ for XARC) and normal-to-fast speeds ($p = 0.03$ for radius; $p < 0.01$ for XARC). For arc lengths, the post hoc test indicated that although the arc lengths were significantly different when we compared slow-to-normal ($p = 0.03$) and slow-to-fast speeds ($p = 0.049$), arc lengths were not significantly different when we compared normal-to-fast walking speeds ($p = 1.00$). The results of the statistics tests are further illustrated in bar graphs **(Figures 7** and **8**).

DISCUSSION

Simple models of walking, such as the inverted pendulum or even the rocker-based inverted pendulum [14], have dynamics that do not depend on the mass of the body. However, these models assume “stiff” characteristics of the legs; i.e., the legs conform to the same effective geometries throughout the stance phase to uphold the body mass. In reality, this feature must be accommodated through the use of musculoskeletal mechanisms. With increased load to the torso, an increased loading can be expected on the lower-limb systems studied in this paper (i.e., the ankle-foot and knee-ankle-foot musculoskeletal

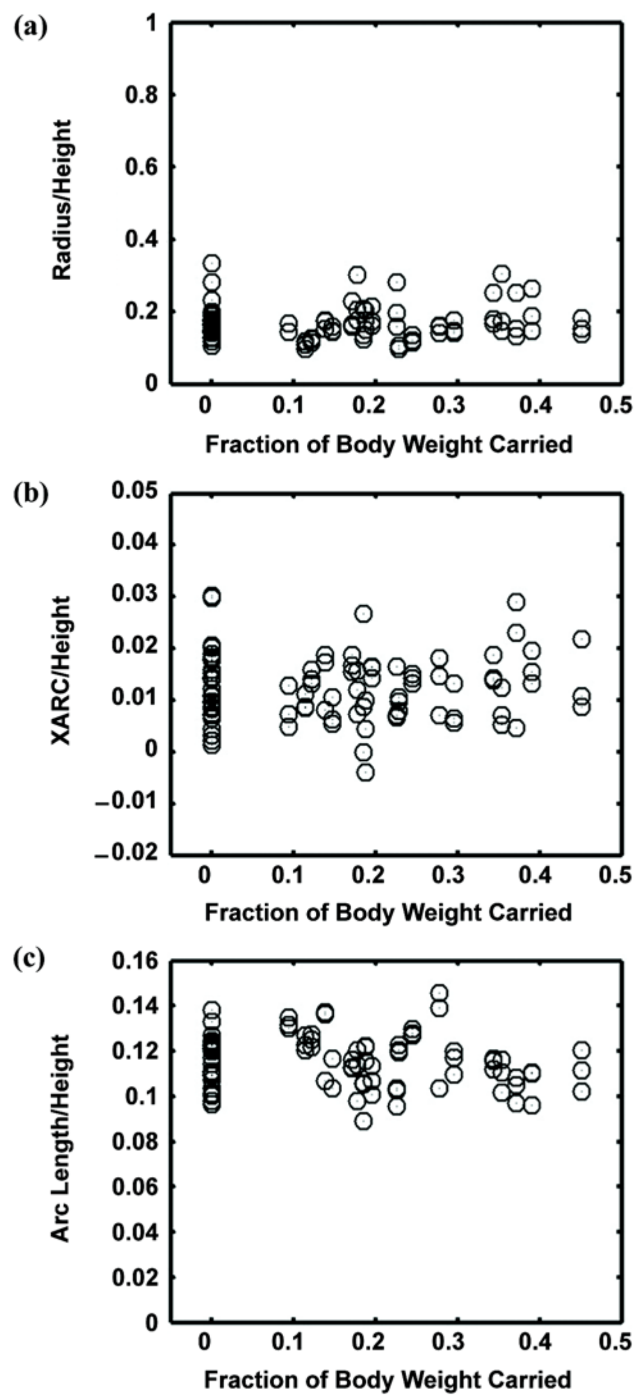

Figure 6.
Best-fit circular arc parameters for knee-ankle-foot (KAF) roll-over shapes vs. fraction of body weight carried (**Figure 3** shows an explanation of parameters): **(a)** radius, **(b)** forward shift of arc, and **(c)** arc length. Characteristics do not change significantly with increased amounts of added weight to torso, as would be expected from roll-over shapes seen in **Figures 4** and **5** and from results of analysis of variance. Median radius/height, XARC/height, and arc length/height values for all KAF roll-over shapes were 0.16, 0.007, and 0.11, respectively. "XARC" refers to forward shift of arc.

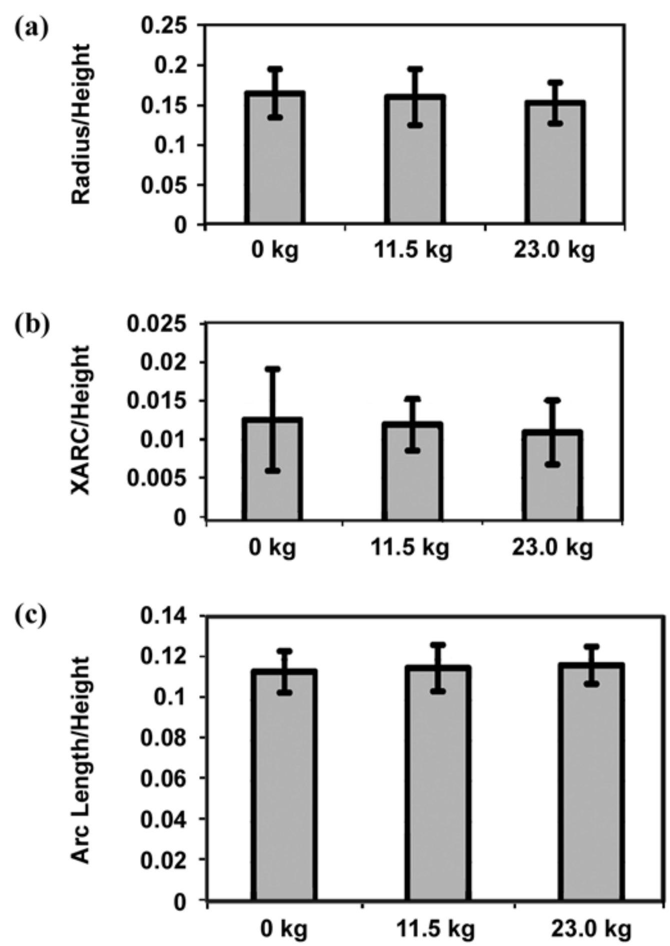

Figure 7.
Circular-arc parameters, normalized by body height, for knee-ankle-foot roll-over shapes, (**a)** radius, **(b)** forward shift, **(c)** arc length vs. added weight to torso; 3 × 3 two-way repeated-measures analysis of variance indicated that added weight did not significantly affect these parameters. Similarly, interaction between added weight and walking speed did not significantly affect these parameters. "XARC" refers to forward shift of arc.

systems). The increased loads to the physiologic systems would cause larger angular displacements in the joints of the foot, ankle, and knee during walking if "quasi-stiffness" in these joints were maintained at a constant level. (The term "quasi-stiffness" refers to changes in joint torques divided by the subsequent displacement changes in the joints [26].) These changes in angular displacement would cause altered roll-over characteristics, i.e., roll-over shapes. In particular, added weight could result in roll-over shapes with smaller radii if quasi-stiffness values are unchanged. Since roll-over shapes were maintained, the physiologic lower-limb systems

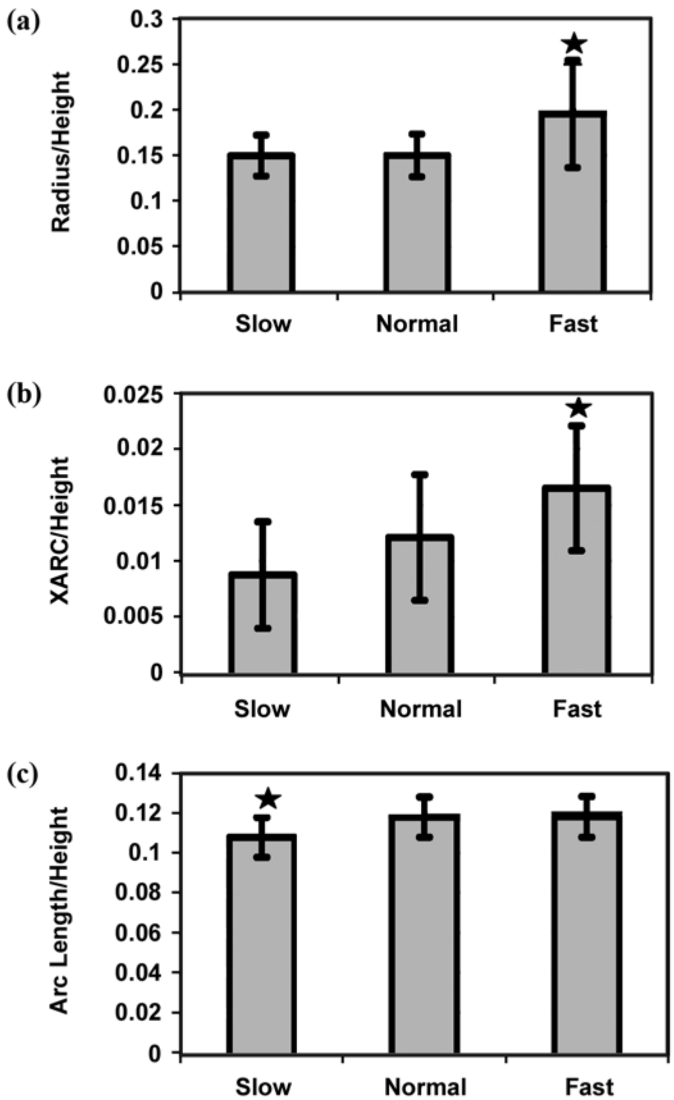

Figure 8.
Circular-arc parameters, normalized by body height, for knee-ankle-foot roll-over shapes, **(a)** radius, **(b)** forward shift, **(c)** arc length vs. walking speed. Post hoc tests indicated that radii and forward shifts for fast walking were significantly higher than those for slow and normal walking. Arc lengths for slow walking were significantly different from those for normal and fast speeds. Stars in plot indicate statistical significance. "XARC" refers to forward shift of arc.

appear to have either adjusted their quasi-stiffness values when increased loads were carried on the torso or the inherent properties of the limb systems made them robust to the changes that were imposed in this study.

The changes in arc lengths of the roll-over shapes seen at slow walking speeds may be related to the step length. At low walking speeds, a person takes smaller steps and uses less of his or her "effective rocker" to "roll-over" to the next step. As the speed is increased, more and more of the rocker is used as the steps are increased. However, the increase in arc length reaches a saturation level when the length of the rocker can no longer be increased because of the physical constraint of the foot's length. This reasoning explains the finding that arc length increases up to normal speeds and then remains constant above these speeds. Further increases in step length beyond the point where the physical constraint of foot length is reached can still be achieved by a person's either rocking about the ends of the feet (actually under the metatarsal heads) or increasing the amount of pelvic rotation.

Changes in forward shift and radius (**Figure 8**) suggest that the knee-ankle-foot system becomes "stiffer" at higher walking speeds. The radius seems to increase, suggesting smaller overall displacements at the heel and at the toe, even though the ground reaction forces have increased amplitudes for higher walking speeds. The shift seems to be somewhat forward as the speed is increased. However, we found no significant differences between radii and forward shifts when subjects walked at slow or normal walking speeds.

Limitations of this study include the fact that trials were not randomized. This factor may have led to fatigue in subjects near the end of the study when they were carrying heavy weights, although if subjects had carried weights by hand they likely would have been more fatigued [2]. Loads were set quantities and were not specific percentages of each person's overall weight. Using set quantities of weight was convenient and allowed us to examine a variety of carried weight percentages because of the variation in the weight of the subjects in the study. However, because of this factor, the relative loading was different for each individual. To reduce the possible effects of fatigue and to speed up the trials, we allowed subjects only a short period of time to adjust to the added weight. We believe that the subjects adjusted to the added weight quite quickly, although further study would be needed to verify this assumption.

For the design of rehabilitation devices such as prostheses or orthoses, a general invariant model for roll-over shapes seems appropriate. Using an invariant roll-over shape as a design goal implies that these devices should be constructed so that they deform under walking loads, thus creating an appropriate roll-over shape that does not change appreciably when persons walk at different speeds or when they carry objects having different weights. The use of materials with nonlinear stiffness properties (i.e., properties of a "hard spring") may be the simplest way to mimic this behavior with the use of passive devices. If the materials are chosen so that the operating weight (e.g., body weight) resides in the steeply sloped portion of the nonlinear stress-strain curve, fluctuations about the operating weight will result in very small displacements.

CONCLUSIONS

Adding weight symmetrically about the torso does not significantly affect the roll-over shapes of the ankle-foot and knee-ankle-foot systems. These musculoskeletal systems appear to adapt to changes in loading to maintain similar roll-over characteristics. Although roll-over shapes of the knee-ankle-foot system are not significantly changed between slow and normal walking speeds, they are altered somewhat at fast speeds. Use of an invariant roll-over shape model seems appropriate in the design and development of prostheses and orthoses because of its simplicity and because these devices are frequently used within the slow-to-normal walking-speed ranges of able-bodied persons.

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