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BIOMECHANICAL EFFECTS OF CARRYING A UNILATERAL LOAD ON HUMAN BIPEDALISM, AS INDICATED BY FOOTPRINT TRAIL PARAMETERS

ABSTRACT: The evolution of bipedalism allowed hominins to carry tools, food and infants more easily. Now, humans carry suitcases, grocery bags and toolboxes that sometimes comprise a large percentage of body weight. Preliminary research suggested that when humans carry a heavy, unilateral load we in-toe on the side opposite the load and narrow step width. This project was undertaken to test the idea that both of these changes in foot placement help compensate for the imbalance produced while carrying a load in one hand. If so, a greater load should produce greater effects.

Thirty subjects of both sexes were asked to walk on a paper runner, while wearing paint-soaked socks and carrying a canvas bag with various loads: empty; 7% of body weight; 14%; and 21%. Foot angle, medio-lateral foot placement, step width and step length were recorded and analyzed. Cadence, speed and duty factor were calculated from videos made during the footprinting procedure.

As load increased, foot angle decreased (intoeing), as did step width. Also, step length decreased and the variability of all footprint trail parameters increased with increasing load. The increased variability in foot placement was taken as an indication of "staggering" and partially explains the shorter steps. Cadence, speed and duty factor did not vary significantly with different loads, suggesting that changes in temporal characteristics of gait did not account for the changes in footprint trail parameters.

Studies of modern footprint trails are the best way to understand ancient footprint trails. The results here suggest that the unusual foot angle and step length of one of the Laetoli footprint trails are best understood as resulting from pathology, rather than carrying or species differences.

KEY WORDS: Foot angle – Step width – Step length – Staggering

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INTRODUCTION

Understanding the mechanics of human bipedalism is fundamental to understanding the origin and evolution of the hominins, since bipedalism is often taken as the defining feature of Hominini. A number of explanations for the origins of bipedalism have been proffered, and several of them involve freeing the upper limbs for carrying such things as infants, food or tools (Day, Wickens 1980, Iwamoto 1985, Amaral 1989, 2008, Hewes 1964, Videan, McGrew 2002, Lovejoy 1981, 1988, Brace 1995, 2004). If freeing the upper limbs for carrying was indeed the primary adaptive value of bipedalism, then even the earliest hominins must have been affected by the mechanical properties and constraints of carrying. Therefore, our best hope of understanding our ancestors' methods of locomotion lies in our understanding of modern human locomotion, especially while carrying a load.

Winter, Ruder & MacKinnon (1990) discussed the complexity of the task of bipedal walking and opined that "the control of balance during that task is especially challenging". They cite three facts about human anatomy and locomotion that contribute to the challenge: (1) because we are bipedal, we spend about 80% of the time with only one supporting foot on the ground; (2) our body mass is distributed like an "inverted pendulum", with about 2/3 of the body mass (that which resides in the head, arms and trunk) balanced about 2/3 of body height above the ground; and (3) the center of gravity of the body does not pass within the foot during normal walking, but instead passes along the medial border of the foot. This last point means that the body's center of gravity, projected as a point on the ground, often does not pass within the footprints left during normal walking. Balance in the medio-lateral direction is therefore dynamic, requiring the center of gravity to be kept oscillating between the centers of pressure of the two feet.

When a heavy load, such as a backpack or heavy shoulder bag, changes the weight distribution of the body, the most common response is to adjust posture by leaning away from the load (Garciaguirre *et al.* 2007). That this is a compensatory mechanism has been established by the repeated observation that heavier loads elicit greater responses (Martin, Nelson 1986, Li *et al.* 2003).

In addition to postural adjustments, footfall modifications may also be made. For example, when carrying an asymmetrical load, subjects typically reduce speed, shorten their strides and increase the percentage of the stride in which both feet are on the ground (Martin, Nelson 1986, Wang *et al.* 2001). Although several studies like these have examined the same three parameters (speed, stride length, support time), little work has been performed on footfall parameters which might be evidenced by footprints, despite the fact that balance in the coronal plane "is achieved by the initial placement of the foot which determines the coarse control" of the upper body (Winter *et al.* 1990). Thus, when changes are made to the system, such as by carrying a load in one hand, changes in the placement of the foot will also be necessary. These changes should be amenable to footprint analysis.

As part of his research into the function of the upper limbs, Webb (1989) performed a brief biomechanical analysis of carrying a heavy load by hand. The subjects were asked to walk under three loading conditions: walking comfortably with no load (unloaded); carrying a 30 lb (13.6 kg) bag of books in one hand (unilateral load); carrying the same bag in front of them, with both hands (frontal load). Several gait parameters were recorded, including stride length, right and left step length, step width and foot angle. The results of that pilot study showed significant changes in some gait parameters when carrying a unilateral load, but none when carrying a frontal load. Specifically, when carrying the load in one hand, step width decreased, as did foot angle on the unloaded side. It was hypothesized that we make these two changes to bring the free-side foot further under the load to help balance it. However, no further test of this idea was performed at the time of the original study. Furthermore, the original study involved only seven subjects who carried the same load, regardless of their own body size, with the result that the load varied between 17% and 28% of the subjects' body weights (Webb 1989). And, for the initial study, footprint measurement techniques had never been tested for consistency or precision.

The current project, with more subjects and better experimental control, was undertaken as an indirect test of the hypothesis that both of these changes in foot placement are part of our compensation for the imbalance that occurs while carrying a load in one hand. If the hypothesis is true, a greater load should produce more marked effects. Based on the original research (Webb 1989), it is expected that foot angle and step width would both decrease with increasing load. Since these methods are hypothesized to be attempts to regain balance when carrying a unilateral load, the amount of change in foot angle and step width might tell us if particular subjects tended to use one method or the other, or use them to different degrees. Also, it might explain the tendency to trip on one's own feet when the load (e.g., suitcase, toolbox) is particularly heavy, because in-toeing and narrowing the step increase the chances of catching the free-side toes on the heel of the load-side foot. Furthermore, a very heavy load might cause the loss of precise control over walking which would also lead to tripping or staggering. Furthermore, since some temporal characteristics of human walking (e.g., cadence, duty factor) have been linked to changes in foot placement, observations and analyses of such characteristics were included to better understand our results.

MATERIALS AND METHODS

For the current project, a procedure similar to that described by Uetake (1992) was employed. Specifically, subjects were asked to walk in paint-soaked socks, on a paper runner 24 inches wide (61 cm) and 7 m long. The first and last meter of each trail were ignored, since subjects accelerate and decelerate when starting and stopping each walking trial. Therefore, only 5 m of each trail were used in the analysis. Because no significant differences between unloaded trials and front-loaded trials were found in the original study (Webb 1989), only unloaded and unilaterally loaded walking trials were tested, here.

Thirty subjects (12 male, 18 female) were asked to walk on a paper runner while wearing thin, tight socks soaked in paint and carrying a canvas bag with various loads. Age, sex, stature and body weight are as described in *Table 1*. From stature and body weight, body mass index (BMI) was calculated for each subject. All subjects gave informed consent, after the testing protocol was explained to them, and the protocol was approved by the Institutional Review Board for Human Subjects of Kutztown University. When asked, no subjects reported any musculo-skeletal or neurological conditions that might hinder them or affect their performance in the experiment. No subject exhibited or reported any sign of slipping during walking trials. Also, during the measurement phase of the experiment, each footprint was examined for signs of slippage, but none was found. The protocol, used in previous studies by us (Webb, Bratsch 2017), was therefore considered safe for subjects and legitimate for measurement of footprint trails.

TABLE 1: Descriptive statistics of human subjects.

	Mean	Std. Dev.	Min	Max
Age (yr)	21.5	5.6	18	45
Female	20.4	4.1	18	36
Male	23.1	7.2	18	45
Stature (m)	1.67	0.10	1.50	1.88
Female	1.61	0.05	1.50	1.70
Male	1.78	0.06	1.67	1.88
Weight (kg)	73.4	18.6	47.9	111.0
Female	63.6	14.9	47.9	107.7
Male	88.1	13.1	61.8	111.0

Subjects made footprint trails under four loading conditions. 'A' trails were made without any weights, merely an empty canvas bag (0.6 kg); for 'B' trails, barbell weights totalling 7% of the subject's body weight were added; for 'C' trails the subjects carried a moderate load of 14% of their body weight; and for 'D' trails they carried the heaviest load, 21% of body weight (*Figure 1*). Short-term practice over a period of only five or ten minutes affects subjects' use of foot placement techniques while carrying a heavy load (Webb, Bratsch 2017). Therefore, all subjects were given about ten minutes of practice carrying a heavy load in the same manner as required by the experiment, prior to engaging in measured trials. Also, to reduce any confounding effects of short-term practice, the order of the walking trials (A–D) was randomized. In this study, the amount of the load was more precisely controlled by using barbell weights and a computer

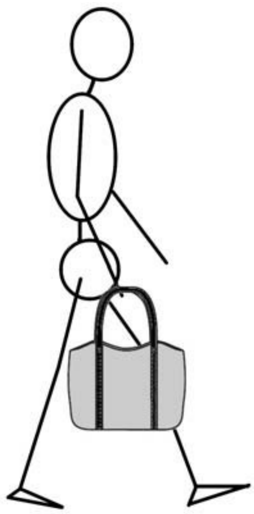

FIGURE 1: Diagram of subject carrying bag during experiment. Drawn from still frame of video.

program which calculates the appropriate combination of weights needed. A test of the computer output and the precision of the estimations was performed, prior to soliciting subjects, and again after all trials' weights were calculated. The amount of deviation from the target weight (7%, 14% or 21% of body weight) was less than 0.5 kg (between -0.365 and 0.427 kg), and 84% of the actual weights used were within 0.25 kg of the target weight. Thus, the degree of the unbalancing load was much more precisely controlled in this case.

Subjects were videotaped while walking using a single consumer-grade (1080p) video camera (Samsung HMX-Q20) placed 12 meters lateral to the walkway. The camera operated at the rate of 30 frames per second. To determine the timing of events (e.g., heel strike), videos were viewed on a large screen monitor (Apple iMac 27") and frames were counted while the video was forwarded, frame-by-frame. From analysis of the videos, temporal characteristics of gait were calculated, specifically cadence (steps/s), speed (m/s) and duty factor (percent of stride each foot is in contact with the ground). To calculate duty factor, two strides near the middle of each walking trial were measured: one stride of the load-side foot and one of the free-side foot.

The footprint trail parameters which were measured were foot angle, step length and distance to the edge of the paper runner. From the edge distance of consecutive footprints, step width was calculated. Also, the footprint marking and measuring techniques of Webb, Bernardo & Hermenegildo (2006), which have been tested for precision and repeatability, were used for the current project. The measurement techniques are depicted in *Figure 2*. Foot angle was measured along the longitudinal axis of the footprint, as defined by the midpoint of the ball and the midpoint of the heel. Measurements were made by naïve observers, in that about 8–12 footprints were measured from each footprint trail, trails were measured out of order (i.e., not in the order in which they were made, and not all trails of the same subject were measured on the same day), and trails were labelled with letters (A, B, C, D) rather than percent body weight. This protocol was used to reduce observer bias by making it nearly impossible for any observer to know what to "expect" among the measurements of any given footprint trail.

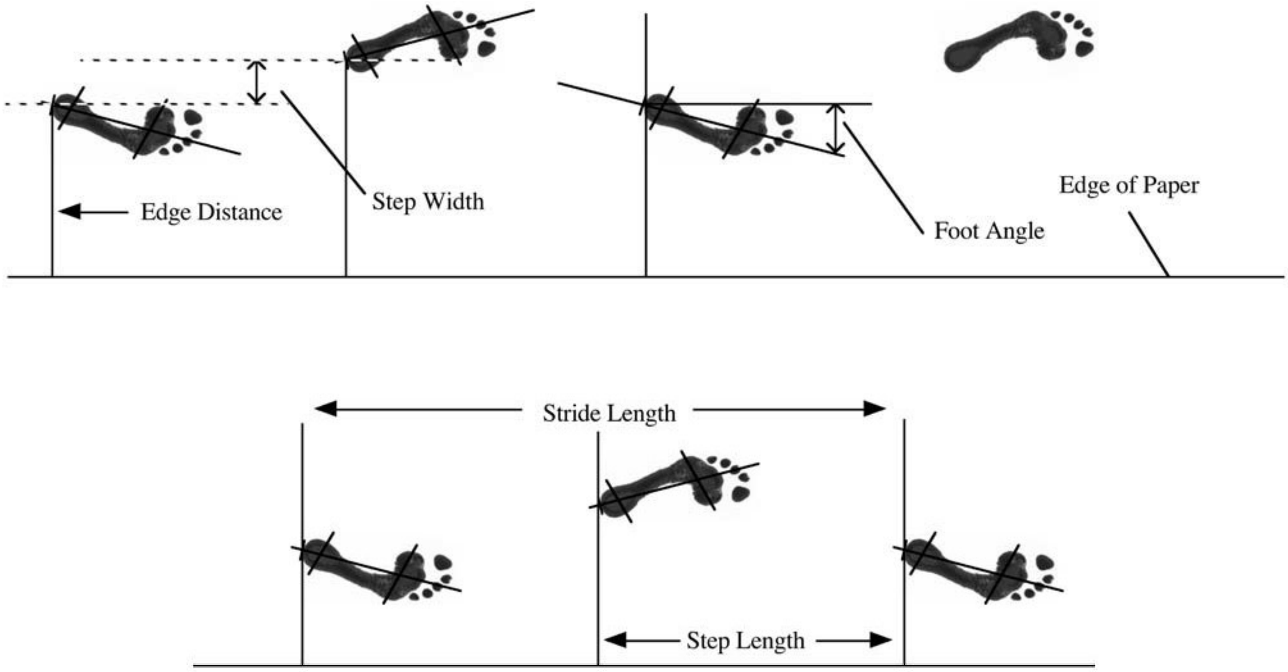

FIGURE 2: Measurement techniques. Individual footprints were marked according to Webb et al. (2006), with the longitudinal axis defined by the midpoints of the ball and heel. Distance to the edge of the paper was made from the posterior heel point, where the longitudinal axis crosses the heel print. Foot angle is measured relative to the line of progression which is taken to be parallel to the edge of the paper. Positive angles indicate out-toeing. Figure is taken from Webb and Bratsch (2017).

Most measured footprint trail parameters were analyzed by means of Linear Mixed Effects (LME) modelling. This technique is an improvement over t-tests and ANOVA, since it allows the variance within each footprint trail to be included in the analysis, whereas t-tests and ANOVA require averaging parameters (e.g., foot angle, step length), to collapse them into a single datum for each footprint trail. Another advantage of LME modelling is that other factors that might covary with the results can be included (Fitzmaurice *et al.* 2004). Therefore, each of the three dependent variables of primary interest in this study (foot angle, step length, step width) was separately analyzed by means of LME modelling. For foot angle and step length, the loaded side and the free (unloaded) side were analyzed separately. In each model, footprint trail was treated as a categorical variable and the overall fixed effect for trail was examined. Thus, there were four groups: 0%, 7%, 14% and 21% of body weight being carried (labelled A, B, C and D trails, respectively). A compound symmetric correlation structure was used to account for within-subject correlation. Post-hoc pair-wise comparison analyses were also conducted, and Bonferonni-adjusted p-values were reported, thereby adjusting for multiple comparisons. In order to explore the effects of sex, walking speed and body mass index (BMI) on each footprint trail parameter, a separate model, including these three covariates, was fitted for each outcome, and the results below indicate the cases in which they showed significance.

Twenty-one percent of one's body weight is a considerable load for many people, especially when carried in one hand. Such a heavy load is difficult to control and sometimes leads to staggering. We may consider "staggering" to be a poorly controlled, erratic form of locomotion. Therefore, a measure of staggering would be the statistical variation among any of the measurable parameters of gait (e.g., step length, instantaneous path of progression, foot angle). This changes "staggering" from a categorical variable in the vernacular sense, to a continuous variable in the scientific sense, and allows us to talk about degrees of staggering and to apply appropriate statistical tests.

Comparisons of statistical variation among the parameters measured here were made with repeated measures ANOVA (RM ANOVA), with the category of loading trials as a between-subjects factor. Again, there were four groups: A, B, C and D trails. In the case of foot angle and step width, standard deviation was calculated for each footprint trail, then that number was used as a single datum for each subject for each walking trial. For step length, coefficient of variation (CoV) was used. CoV was not suitable for foot angle and step width, since these values are often close to zero and sometimes negative.

To test the degree to which subjects used the two techniques of main interest (in-toeing and narrowing of step), correlation analyses were performed. Correlations were sought between foot angle with no load (A trails) and the amount by which foot angle changed with the heaviest load (D trails), using Pearson's correlation coefficient. The same test was performed for step width (i.e., A trails compared to the difference between A and D trails). This provided a test of the amount that individuals used one or both techniques.

Finally, because some temporal characteristics of gait (speed, cadence and duty factor) have been linked to the parameters which were measured in this study, it was necessary to consider their effects. To be sure that temporal factors were not confounding the results described below, RM ANOVA tests were performed, since there was exactly one datum for each walking trial for each subject. Speed, cadence and duty factor were analyzed separately, and in these analyses, differences among the four different loading regimes were sought, with the four loading percentages as the within factor (A trails, B trails, etc.). Relationships among speed, cadence and duty factor themselves were assessed with simple linear regressions.

All statistical tests were two-sided and p-values <0.05 were considered statistically significant. For all LME models, Statistical Analysis Software (SAS, version 9.4, Cary, N.C.) was used. To generate graphs of data, Statview 4.02 (Abacus Concepts) was used. Simple linear regressions were also produced in Statview 4.02.

RESULTS

Foot angle

Foot angle on the free (or unloaded) side showed an absolute decrease with increasing load, when all individuals were grouped (*Figure 3*). Foot angle decreased steadily, from an average of 4.98° in A trails to 2.65° in D trails. A Linear Mixed Effects model showed the trend to be significant ($p < 0.0001$). SAS output automatically displayed a comparison between D trails and each of the others (*Table 2*). From these results, we see that D trails with 21% of body weight showed significantly lower foot angles than all others. However, the extent to which particular subjects reduced foot angle on the free side was not significantly correlated with their initial, "natural" foot angle. Thus, subjects with relatively large foot angles on the free side did not reduce foot angle any more than subjects with small foot angles, even when carrying the heaviest load.

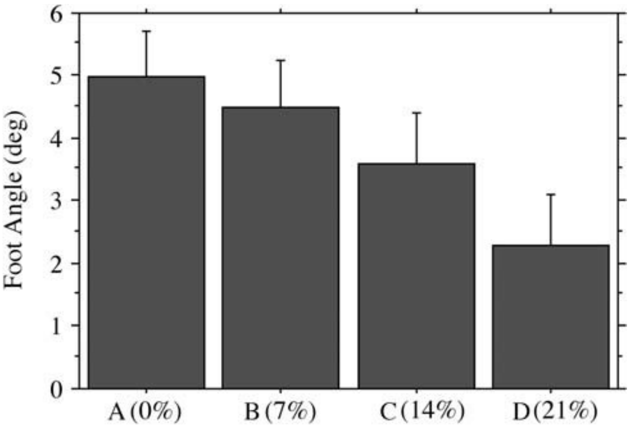

FIGURE 3: Average foot angles of the free-side foot decreased significantly with increasing load. For this and all bar graphs, whiskers indicate one standard error above the mean.

TABLE 2: Foot angle on the free side for each footprint trail compared to D trails.

	Mean Difference (deg)	p-Value
A–D	2.32	<0.001
B–D	2.30	<0.001
C–D	1.37	0.002

On the other hand, pairwise comparisons between footprint trails were not generally significant for the foot angle on the loaded side, but the overall effect was significant ($p = 0.013$). This seems to be driven mostly by the significant increase in foot angle from unloaded A trails to heavily loaded D trails ($p = 0.025$).

When sex, speed and body mass index (BMI) were included in the model, it was found that sex and speed had no effect on foot angle ($p > 0.41$). However, BMI was positively correlated with foot angle on both sides. Therefore, regardless of the amount of load, or the side on which it was carried, greater BMI was associated with wider foot angles (free-side $p = 0.0006$; loaded-side $p = 0.0065$).

Step width

Step widths conformed to our expectations. In accordance with our hypothesis, there was a significant decrease in step width with the added load (LME model p-value < 0.0001). In pairwise comparisons with D trails, all others were significantly higher ($p \le 0.0123$). Sex, walking speed and BMI were not significant factors in the LME model ($p > 0.24$) for step width. Results are presented graphically in *Figure 4*.

An additional test of the reduction in step width with regard to original (unweighted) step width showed an interesting result. Correlation analysis yielded a correlation coefficient $r = -0.409$ ($p = 0.0266$), indicating that those with the widest normal step widths were the ones who reduced step width the most when carrying a heavy load.

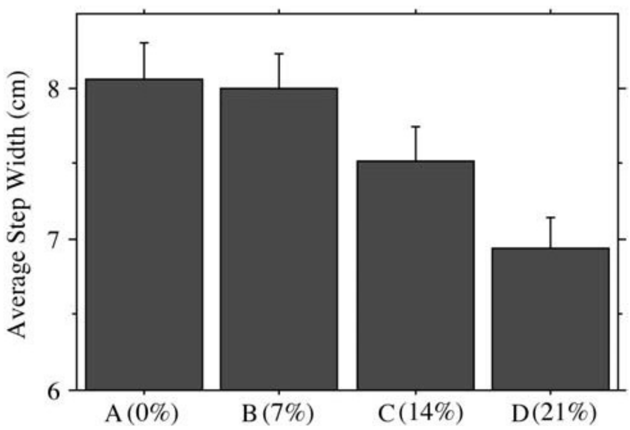

FIGURE 4: Step width in all four walking trials showed a decrease with increasing load.

Step length

Table 3 presents the results of separate linear mixed-effects (LME) models for step length on the loaded side and on the free side. In this case, the parameter "Trail" refers to the LME assessment of the overall significance of the amount of load on step length. All pair-wise comparisons between D trails and other trails showed significant decreases in step length with increased load. When loaded side and free side are combined, the effect of load is also very significant ($p < 0.0001$). *Figure 5* shows average step lengths for each load regime (both sides combined). In all comparisons, the greater loads elicited smaller steps.

In an additional LME model of step length, sex, speed and BMI were included. As found by many previous authors, speed was significantly related to step length ($p < 0.0001$), but sex and BMI were not (p-values ranged from 0.35 to 0.78).

TABLE 3: Linear Mixed Effects comparisons of step length on the loaded and free sides. "Trail" refers to the whole model, and paired comparisons with D trails are so labelled (e.g. "A–D" means A minus D).

	Loaded		Free	
	Mean Difference (cm)	p-value	Mean Difference (cm)	p-value
Trail	–	<0.0001	–	<0.0001
A-D	5.58	<0.0001	5.77	<0.0001
B-D	4.10	<0.0001	4.18	<0.0001
C-D	1.89	0.0002	2.53	<0.0001

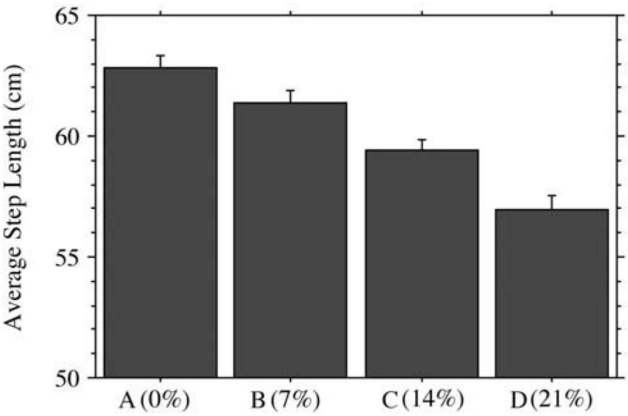

FIGURE 5: Step length decreased with increasing load. See text for tests of significance.

Staggering

Standard deviations of foot angle on the free side increased with increasing load (*Figure 6*). On the free side, the LME model showed a significant effect ($p = 0.003$) overall. The A, B and C trails were nearly identical in standard deviation. The dramatic increase in standard deviation of the D trails apparently drives the significant result for the free side. Specifically, D trails with 21% of body weight evinced significantly greater variability than any of the other trails ($p \leq 0.026$ in pairwise comparisons). On the loaded side, however, there were no significant results ($p = 0.337$), even though there was a monotonic increase in standard deviation as load was increased.

Standard deviation of step width did not show any trend toward increase with increasing load. C Trails, with 14% of body weight were the most variable, while D trails were similar to B trails in having the least variation.

Analysis of step length also provided evidence of staggering when a heavy load was carried. Because step length decreased with increasing load (see above), the same absolute amount of variation might appear more important in the heavily loaded D trails than in the others. Therefore, coefficient of variation (CoV) of step length was analyzed, since it is standardized by the average value of step length in a given series (i.e., A trails). Results for the loaded side indicated no significant differences in pair-wise comparisons. However, the whole LME model produced a significant effect ($p = 0.038$). For step length on the free side, the whole model also indicated a significant increase in variability ($p = 0.0001$). In this case, A, B and C trails were similar, but D trails were significantly more variable than any of the others.

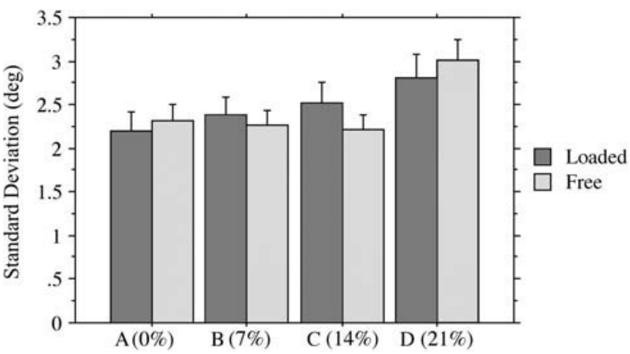

FIGURE 6: Standard deviations of foot angle. As a measure of "staggering", standard deviation shows a monotonic increase with increasing load among the loaded-side footprints and a dramatic increase with the 21% load on the free side.

Temporal characteristics

Average speed for all subjects for each of the four walking trials was approximately 1.1 m/s. However, RM ANOVA, with sex included, showed two main differences. First, male subjects walked faster under all conditions ($p = 0.021$), with an average difference of 0.135 m/s. Second, there was a significant slowing with increased load ($p = 0.005$; *Figure 7*). Pairwise comparisons showed that the slowest walking trials with the heaviest load (D trials, at 21% body weight) are what caused the significant result, since D trials were significantly slower than A and B trials, but no other pairwise comparisons were significant.

Males evinced a slightly lower cadence than females, but it was insignificant (RM ANOVA $p = 0.42$), as was the test for cadence under the various loading conditions ($p = 0.175$). There was also no clear pattern of change in cadence, with C trials (14% body weight) evincing the highest cadences.

Despite these results for speed and cadence, a close relationship between them was found (*Figure 8*), as has been shown by many other authors. In this case, the regression of speed on cadence produced a highly significant positive slope ($p < 0.0001$) and a fairly tight fit of the data to the line (R-squared = 0.706). The same type of relationship was found when each loading regime was analyzed separately. For all four types of load (0% through 21%), there was a significant positive slope, ranging from 0.731 to 1.22, and a good fit to the regression line ($0.57 \leq R^2 \leq .86$).

Duty factor, or support time expressed as percent of stride time, also showed no significant results in an RM ANOVA test. For example, there was no significant difference in duty factor between the loaded side and the free side ($p = 0.74$). The duty factor, in all cases, was approximately 66%, a value very similar to those found by others for unencumbered walking at similar speeds (e.g., Donelan, Kram 1997, Alexander 2004). The p-value for duty factor evaluated by amount of load (or trail type) was 0.26, when load-side and free-side were averaged; when separated, the effect of the load was also insignificant (load-side p-value = 0.19; free-side = 0.46).

Because the load-side duty factor and the free-side duty factor were very similar, they were averaged to perform an additional test: the averaged duty factor was regressed on speed (*Figure 9*). The correlation analysis shows a rather poor correlation between duty factor and speed ($R^2 = 0.11$), but the relationship is significant ($p = 0.003$ for slope of the regression line). The 95% confidence interval for slope is from -0.072 to -0.016. There is, therefore, a slight negative correlation between duty factor and speed, as found by other authors (e.g., Nardello *et al.* 2011).

DISCUSSION

The results of this project agree with those from the original study (Webb 1989). In addition, some new results were obtained which further elucidate the biomechanical effects of carrying on modern human bipedalism.

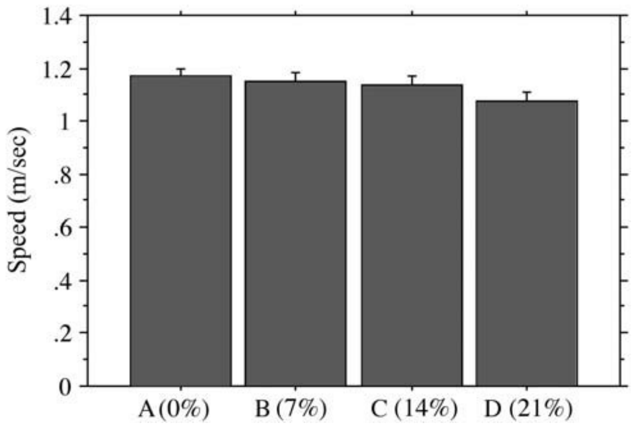

FIGURE 7: Average speed under four loading conditions. The noticeable decrease in speed with increasing load was significant in group form (RM ANOVA p-value = 0.005), and D (21% body weight) walking trials were significantly slower than A and B trials ($p < 0.034$).

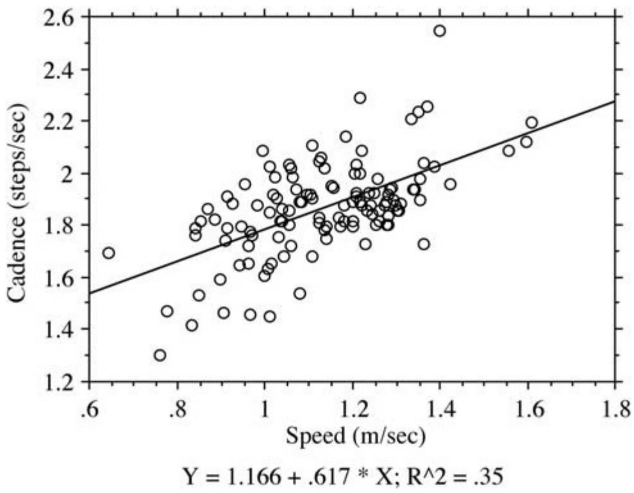

FIGURE 8: Cadence vs. walking speed for all walking trials. The slope of the regression line is significantly different from zero at the $p < 0.0001$ level.

Foot angle

A small amount of out-toeing is normal in modern human bipedalism (Selby-Silverstein *et al.* 2001, Toda *et al.* 2003), and our results for the unweighted A trails (averaging 6.8 deg of out-toeing) are consistent with the findings of others who studied unencumbered

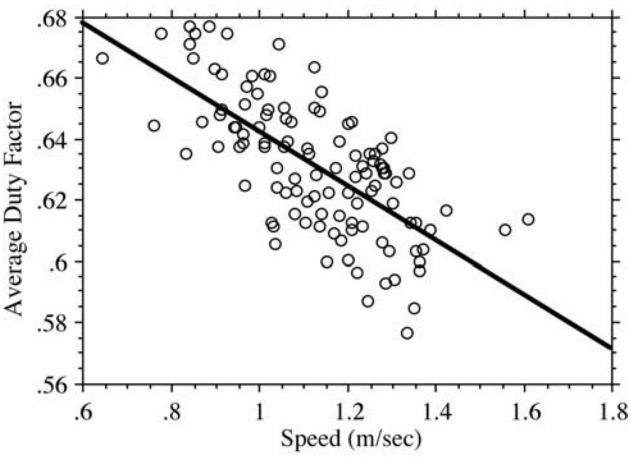

FIGURE 9: Duty factor vs. walking speed for all walking trials. Each point on the graph represents the average of loaded-side and free-side duty factors plotted against the speed of that trial. Here, p = 0.003 for the slope.

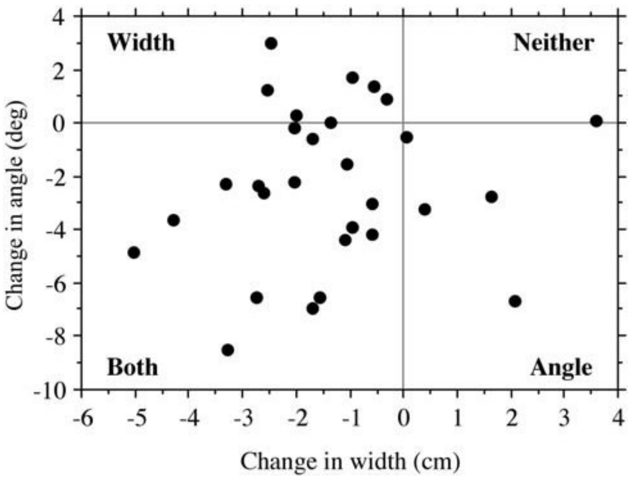

FIGURE 10: Change in foot angle vs. change in step width, from A (0%) to D (21%) loads. The x- and y-axes are drawn to emphasize the four quadrants of the Cartesian plane. Note that only one individual did not change foot angle or step width to accommodate the heavy load, thus using neither of the expected techniques.

subjects. However, as expected, foot angle on the free side showed a significant decrease with increasing load. This is a clear indication that one way in which humans compensate for a load carried in one hand is to rotate the contralateral foot medially, towards the load. This places the anterior foot further under the load during the single-support phase, when the lower limb on one side is required to support both the body and the unbalancing load on the opposite side. The average change in foot angle is small, being a little over 3 deg from unweighted to heavily weighted walking trials, although this represents a decrease of almost half. Also, using simple trigonometry, we can calculate that for a foot length of 25 cm, this is equivalent to moving the toes about 1 cm closer to the loaded side. This, in combination with other gait changes, seemed to be sufficient to support a unilateral load of 21% of the subjects' body weight.

Step width

As expected, the additional compensatory technique of decreasing step width was also used. Decreasing step width would also bring the supporting foot further under the load, in this case by about 1.4 cm. This agrees well with the original study by Webb (1989) which showed a decrease in step width of ≈1 cm. An additional test was performed, to determine the importance of narrowing the step among various subjects. To do so, initial (unencumbered) step width was compared to the amount of decrease in the most heavily weighted trials for each subject. This showed a significant trend for those with a wide step to narrow it more than those who naturally have a narrow step. The correlation between initial step width and the difference between D and A trails was -0.409 ($p = 0.027$, $R^2 = 0.168$). Thus, those with a narrow step to begin with did not narrow it very much, whereas those with wide steps narrowed them the most, when carrying the unilateral load.

Since subjects used both methods of handling the unbalancing load, in-toeing and narrowing step width, we chose to consider the degree to which individuals used both techniques. Using the difference between the most heavily weighted walking trials and the unweighted trials (D–A) as an indicator of the amount that each technique was employed, a regression of the change in foot angle against the change in step width was produced (*Figure 10*). Since reduction in either parameter results in a negative value, those who used both techniques to accommodate the heavy load were plotted in quadrant III of the Cartesian plane, those who used neither were in quadrant I, and those who reduced only foot angle or only step width were plotted in the fourth and second quadrants, respectively. Of 29 subjects who provided useful data, only one (3%) used neither method of accommodation; 14% used only in-toeing, 24% used only step width reduction, and 59% used both methods. Thus, these two gait adjustments, singly or in combination were used extensively (97%) by the subjects to support the unilateral load. However, it should be noted that no significant pattern of combining the two methods was found (p-value of slope = 0.46; $r^2 = 0.021$). This suggests that, though the majority of subjects used both techniques, there was no real correlation between the two in terms of the extent to which they were used.

A confounding factor in the analysis of step width is the fact that it is inversely related to speed. Orendurff et al. (2004) found that, as subjects increased speed they decreased step width. Their subjects walked at speeds of 0.7, 1.0, 1.2, and 1.6 m/s as directed by the investigators, and step width decreased by a total of 5.0 cm over that range of speeds. Orendurff et al. (2004) interpreted their data as indicating a reduced need for balance in the coronal plane when walking at higher speeds, since the center of mass of the body oscillates less (medio-laterally) at higher speeds. However, the effect that they describe cannot explain the results obtained here. Since speed decreased significantly with the heaviest load, we would expect step width to increase rather than decrease, as observed. We can, therefore, be fairly certain that the decrease in step width was a compensation for the unilateral load being carried, rather than a result of changing speed.

Step length

The initial study, involving only 7 subjects, showed a slight and insignificant decrease in step length while carrying the load unilaterally and frontally. In the current project, with many more subjects, significant differences were found.

In our LME model comparing individual performance among the four loading regimes, step length decreased with the increasing load, especially with a heavy load. This may be due to several factors relating to effort, discomfort and balance. When we walk, our center of mass oscillates about 3 cm in the sagittal plane as we alternately "vault" over one lower limb, then the other (Inman *et al.* 1981). The length of each step (or more accurately, the lengths and angles of the limbs at the moment of heel strike) determines the amount of vertical displacement (Helene 1984) with shorter steps yielding smaller vertical movements (Webb 1996). Since some muscular action is required to lift the body over the supporting limb and to stabilize the body medio-laterally (Carlsöö 1972, Winter *et al.* 1990, Griffin *et al.* 2003), increasing body weight by 21% increases the effort necessary to take each step. This can be accomodated in part by taking smaller steps, reducing the distance that the center of mass must be raised. Furthermore, because ground reaction forces, especially in the vertical direction, are related to body mass, adding weight to the body can increase these forces and thereby increase discomfort and risk of injury (Kinoshita 1985). This effect may be heightened when, as in the present study, subjects walk on a hard floor without the benefit of cushioned shoes. While wearing cushioned shoes, changes in step length and cadence due to increased load are often insignificant (Hong, Cheung 2003), but not always (Pascoe *et al.* 1997) and other compensatory mechanisms may also be used to lessen vertical ground reaction forces (Singh, Koh 2009).

Also, increased mass increases momentum in any direction, making the body+bag system harder to balance and control when moderate and heavy loads are carried. Shorter steps can ameliorate deleterious effects on control by lowering velocity and hence momentum and ground reaction force (Beck *et al.* 1981) or by increasing the frequency of double-contact periods, if speed is maintained (Wang *et al.* 2001). However, when speed is determined *ad libitum* there is not always a significant change in step length or support time (Nottrodt, Manley 1989). Thus, while step length was somewhat reduced in this study, support time as indicated by duty factor did not change significantly.

Winter *et al.* (1990) suggested that shortening swing time (and, therefore, step length) places the foot on the ground before the lateral momentum towards that side becomes too high. With more variable medio-lateral foot placement while carrying a load as shown hereabove, subjects may have benefited from placing their feet more quickly to maintain balance and reduce staggering. However, shortening swing time should have led to increased duty factor, and it did not (see Results). And, in our tests comparing individuals' performance, step length decreased, duty factor and cadence did not. Thus, although cadence and duty factor did not change with increasing load, step length and therefore speed decreased, especially while carrying 21% of body weight.

Both step length and speed showed a monotonic decrease with increasing load. For step length, the change from A trails to D trails was about 6 cm, but it was not the same for males and females. Female subjects' average change was 6.96 cm, but for males it was only 2.31 cm. A similar pattern was observed for speed: females changed speed by 0.12 m/s, whereas males reduced speed by only half that much (0.06 m/s). Furthermore, while a significant sex difference was found in speed, no significant sex difference occurred in step length. Also, as noted above, no change in cadence or duty factor was significant for either sex. It should be noted, however, that females and males were given the same percentages of body weight to carry. Thus, males and females did not react the same way to the same test conditions.

It may be that the range of loads was insufficient to demonstrate the relationships among load, step length and temporal factors, in part because the change in load was relatively small. Some other experimenters used larger percentages of body mass to encumber their subjects, probably increasing the effect on gait. Martin and Nelson (1986), studying college-age men and women in the Reserve Officer Training Corps program, used approximately 1%, 14%, 27%, 45% and 56% of the body masses of their subjects. However, they reported that "*men displayed little change in stride length with increasing load*", with very few significant t-test results. Women, on the other hand, "*displayed a greater sensitivity to the load increases than the men,*" but even then did not show significant changes in all t-tests. For both men and women, they found significant effects of load and sex in two-way ANOVA tests. These results, and those presented here, form a useful contrast with those found by Bampouras & Dewhurst (2016) who studied "older" and "young" women carrying shopping bags. They found that "*carrying shopping bags did not negatively affect postural stability or gait variables*," including stride length and its coefficient of variation and double-support time. However, Bampouras & Dewhurst (2016) used a maximum weight of only 3 kg, which comprised no more than 4.5% of the average body weight of their subjects. Thus, the decrease in step length due to increasing load is often small (as in Martin, Nelson 1986) and will produce significant results in some experiments and statistical analyses, but not in others (Bampouras, Dewhurst 2016).

Staggering under the load

Despite any compensatory techniques used by the subjects, statistical analysis of variance measures indicate that increased loads led to increased staggering. Of the parameters used to determine staggering (foot angle, step width and step length) foot angle and step length showed significant results, adding experimental evidence for the informal observation that body movements become jerky and erratic when we carry a heavy load. This was especially the case while carrying 21% of body weight.

Adolph and Avolio (2000) performed a similar experiment with 14 children aged 14 months. The children were observed under two conditions: wearing a fabric vest of insignificant weight; wearing the same vest loaded with lead shot totalling 25% of body weight. While nearly all of their results were statistically insignificant, they were suggestive of some of the same accommodations made by adults in the present experiment. Adolph and Avolio (2000) also analyzed footprints, made with specially designed shoes, and recorded step length, step width and 'dynamic base'. (Dynamic base was defined as the angle between three consecutive footprints, measured at the posterior-most point of the heel. It is therefore similar to our edge distance and step width, since it is a measure of foot placement relative to the line of progression.) As in the present study, they found that step length decreased with the added load, as did dynamic base. However, step width increased when wearing the weighted vest. They also noted an increase in the amounts of variance in all their measurements, their only significant result being an increase in the coefficient of variance of the dynamic base, a clear indication of staggering like that observed in our subjects.

Temporal characteristics

Orendurff *et al.* (2004) showed that step width decreases with increasing speed (and cadence). Therefore, we might have expected the decrease in step width that was found here to have been prevented by a decrease in speed, as subjects' loads were increased. However, step width decreased with decreasing speed (and increasing load). This suggests that the load was in fact the more important factor determining step width and that, therefore, reduced step width is truly a method of accommodating a unilateral load.

Normally, duty factor is inversely proportional to speed, "*decreasing linearly with increasing velocity*" (Nardello *et al.* 2011), when subjects are unencumbered, walking on a level substrate. That is the case here, as well (*Figure 9*), though the data do not fit neatly on a line ($R^2 = 0.11$).

Others have reported a positive correlation between load and duty factor. In an experiment with guinea fowl, Marsh, Ellerby, Henry & Rubensen (2006) added weights equalling 23% of body mass to the birds and observed them on a treadmill at various speeds. They found that "*loading had no significant effect on swing duration, but caused a significant 4% increase in stance duration*" (Marsh *et al.* 2006: 2054). This is equivalent to an increase in duty factor of something less than 4%. Griffin *et al.* (2003) studied four men and four women traversing a walkway at various speeds while wearing a padded weight belt with up to 30% of body mass added. They noted that: "*At any given speed, stride frequency did not change as subjects carried loads, and duty factor increased only slightly, i.e., 4%, between the unloaded and 30% load conditions*" (Griffin *et al.* 2003: 177). For normal and fast walking speeds, Griffin *et al.* (2003) obtained p-values of less than 0.01 for this relationship between load and duty factor. However, the loading conditions in our study ranged only from 0% to 21% which may not have been enough to elicit a significant change in duty factor. On the other hand, a larger percentage of body weight would have made carrying the load in one hand impossible for some subjects, so our experiment was limited to a smaller range of loads. Finally, in a reversal of this type of experiment, Donelan and Kram (1997) simulated reduced gravity (and therefore reduced body weight) using a body harness and pulley system. The range of simulated gravitational forces was from 25% to 100% (normal). They found that, at a variety of speeds "*duty factors were significantly smaller at lower levels of gravity*" (Donela, Kram 1997: 3198). Once again, changing effective weight by a larger factor (4× in their experiment) may have been required in order to show a significant relationship with duty factor.

Application to fossil footprints

Testing conditions for this study were different from those which apply to ancient hominins walking in naturally occurring substrates. Specifically, our subjects walked in paint-soaked socks on paper which was laid on a hard, tiled floor, whereas the footprint trails at Laetoli, Tanzania, were made in wet volcanic ash (Masao *et al.* 2016). Wall-Scheffler, Wagnild and Wagler (2015) also had subjects walk on paper, but applied paint directly to their feet. As in this study, they reported no slippage, and they found no significant change in fooprint length or width from changes in paint amount or consistency. Also, Wall-Scheffler *et al.* (2015) performed a reliability assessment of fooprints made on paper with those made on a compressible latex pad. They found no significant differences between the prints made on these hard and soft substrates.

Although we are interested in the origins of hominin bipedalism, in most cases this work cannot be applied to fossil footprints, since there are no controls (i.e., we do not know what ancient hominins were doing – carrying or not carrying – while making the footprints that we have found). Nevertheless, Schmid (2009), in his discussion of the Laetoli hominin footprints in Tanzania, suggested three explanations of the footprint patterns: *"(1) an ape- like morphology of the trunk, (2) individuals carrying bigger objects, or (3) pathological conditions*" (Schmid 2009: 61). He lamented: "*We prefer the first hypothesis, although we cannot falsify the remaining possibilities*" (Schmid 2009: 61). However, in the case of the G-1 footprint trail at Laetoli we can say that the dramatic out-toeing (from 13° to 33°; Tuttle 1987), is not likely to be from carrying, since that is not one of the effects of carrying a unilateral load. The right foot, in particular, evinced remarkable out-toeing with an average of 29° (range 22–33°). It is therefore most likely a minor pathology, as suggested by Tuttle (1987): "*The highly out-toed placement of the right foot by G-1 suggests that its lower limb may have been pathological*" (Tuttle 1987: 513). Thus, the hypothesis that the out-toeing was caused by carrying is not supported, and the pathology hypothesis still stands.

Another hypothesis is that the Laetoli print-makers were walking exactly as their species normally does, and that the differences from modern human footprints are because they walked differently. Recently, this hypothesis was advanced by Hatala, Demes & Richmond (2016): "*The Laetoli footprints provide a clear snapshot of an early hominin bipedal gait that probably involved a limb posture that was slightly but significantly different from our own …*" (Hatala *et al. 2016*: 1). However, since the G-1 trail is very different, not only from modern human trails, but also from the G-2/3 trail in several parameters (Tuttle 1987), it is quite possible that the G-1 hominin was not walking in the usual fashion for its species. Therefore, its prints should probably not be used in analyses of this type.

With the recent discovery of hominin footprints at Laetoli Site S (Masao *et al.* 2016), we can be a little more certain that the G-1 footprints are not usual for this species (putatively *Australopithecus afarensis*). Laetoli Site S contains the footprints of two individuals: S2's print is similar in size to G2's, whereas S1's are the largest at Laetoli (Matao *et al.* 2016). Foot angles of S2 are not available since there is only one (damaged) print. However, for S1 there are 11 measurable prints, though only 10 allow foot angle to be measured. Foot angle on the right side ranges from 3 to 8 degrees, and from 2 to 11 degrees on the left

(Masao *et al.* 2016) (average ±s.d. = 5.3° ±2.3, right; 5.7° ±3.8, left). These values are very similar to the average noted above for our unencumbered subjects (6.8 degrees, range -11 to 19). The G-3 trackway allowed four left prints to be measured and five right prints, producing average foot angles of -3.8° ±1.5 and -1.25° ±1.5 for left and right sides, respectively (excluding G-3/29 R which evinced a foot angle [-19°] very different from all others of G-3) (based on Tuttle 1987: 507). Thus, G-3 and S-1 foot angles were approximately bilaterally symmetrical and similar to modern humans', whereas G-1's were asymmetrical and very different from both its conspecifics' and modern humans'. Therefore, pathology is still the best explanation of G-1's unusual footprint trail.

SUMMARY AND CONCLUSIONS

Carrying may have been an important factor in the evolution of human bipedalism. Therefore, since the present is the key to the past, the best way to understand the mechanics of early hominin bipedalism is to compare evidence of early walking to comparable evidence of modern human and/or ape walking. The most direct evidence we have for earlier hominins' gaits is their footprints, even though ancient footprint trails are quite rare. Therefore, while a good general understanding of locomotion can be obtained from skeletal analysis, gait characteristics that can be reconstructed by studying footprints are probably the most useful for understanding the details of the locomotor behavior of our ancestors, including those characteristics that are affected by load carrying.

Carrying a unilateral load, even a moderate one, changes some of the parameters of gait that, in turn, affect some parameters of footprint trails. We have found that foot angle on the unloaded side decreases incrementally with increasing load, as does step width. These two changes are most likely accomodations to the imbalance caused by the load carried in one hand, but they are not used to the same extent by all people. Indeed, both aspects of footprint trails must be considered, since about 41% of subjects here used only one of these means of accommodation. Furthermore, the extent to which we narrow step width depends on how wide our steps normally are, as evidenced by the fact that those with a naturally wider step tend to narrow their gaits more than those with narrow steps to begin with. This may be an indication that an effective limit to step width is reached when the risk of tripping becomes noticeable to the subject. This is compounded by the fact that increased variability (staggering) in several gait parameters also results from increased loads. The combination of staggering, narrowed step width and in-toeing are likely to add significant risk to carrying a load in one hand, especially when the load is greater than about 21% of body mass.

Although ancient footprint trails are very rare, and they are not always in good enough condition to be analyzed as we have done here, there are nevertheless some hypotheses about our ancestors' locomotion that can be tested. Thus, the unusual gait of the Laetoli hominins, particularly the G-1 trail-maker, cannot reasonably be explained by carrying a heavy load. Therefore, the best approach to understanding the disparity between G-1's gait, on one hand, and G-3's and S-1's gait, on the other hand, still seems to be an investigation of footprint trail parameters under various pathological conditions.

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