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Arellano CJ, Kram R. The energetic cost of maintaining lateral balance during human running. *J Appl Physiol* 112: 427–434, 2012. First published November 3, 2011; doi:10.1152/japplphysiol.00554.2011.—To quantify the energetic cost of maintaining lateral balance during human running, we provided external lateral stabilization (LS) while running with and without arm swing and measured changes in energetic cost and step width variability (indicator of lateral balance). We hypothesized that external LS would reduce energetic cost and step width variability of running (3.0 m/s), both with and without arm swing. We further hypothesized that the reduction in energetic cost and step width variability would be greater when running without arm swing compared with running with arm swing. We controlled for step width by having subjects run along a single line (zero target step width), which eliminated any interaction effects of step width and arm swing. We implemented a repeated-measures ANOVA with two within-subjects fixed factors (external LS and arm swing) to evaluate main and interaction effects. When provided with external LS (main effect), subjects reduced net metabolic power by 2.0% ($P = 0.032$) and step width variability by 12.3% ($P = 0.005$). Eliminating arm swing (main effect) increased net metabolic power by 7.6% ($P < 0.001$) but did not change step width variability ($P = 0.975$). We did not detect a significant interaction effect between external LS and arm swing. Thus, when comparing conditions of running with or without arm swing, external LS resulted in a similar reduction in net metabolic power and step width variability. We infer that the 2% reduction in the net energetic cost of running with external LS reflects the energetic cost of maintaining lateral balance. Furthermore, while eliminating arm swing increased the energetic cost of running overall, arm swing does not appear to assist with lateral balance. Our data suggest that humans use step width adjustments as the primary mechanism to maintain lateral balance during running.

locomotion; stability; economy

MAINTAINING LATERAL BALANCE during human walking requires active control, which appears to be accomplished by two primary mechanisms: *1*) varying step width from step-to-step and *2*) arm swing (3, 12, 14, 21). In walking, external lateral stabilization (LS) reduces the need for the active control of maintaining lateral balance and thus reduces energetic cost (12, 14, 21). Donelan et al. (14) demonstrated that external LS reduced step width (by 46%), step width variability (by 31%), and energetic cost (by 6%). Reductions in step width and step width variability are thought to reflect a reduction in the need for the muscles to actively control lateral balance (3, 14) and can therefore explain the reduction in energetic cost. Due to the design of Donelan et al.'s external LS apparatus (14), subjects were required to walk without arm swing, which may exact a greater energetic cost to maintain lateral balance. However, by eliminating arm swing, the experiment of Donelan et al. provided important insights into how humans use step width as an effective and independent mechanism for maintaining lateral balance during walking.

Eliminating arm swing itself increases the energetic cost of walking by 5–12% (11, 21, 23). This increase in energetic cost may be in part due to an increase in the cost to maintain lateral balance. Ortega et al. (21) found that when walking without arm swing, external LS reduces energetic cost by 6%, similar to the finding of Donelan et al. (14). On the contrary, Ortega et al. demonstrated that when walking with arm swing, external LS reduced energetic cost by only 3%. It appears that the 6% cost of maintaining lateral balance originally found by Donelan et al. was due to eliminating arm swing. It has been noted that the reduction in energetic cost with external LS may also be due to providing a restoring torque about the waist, potentially counteracting the whole body angular momentum about the vertical axis (6, 21). As Ortega et al. mentioned, their device assisted with balance in both the "lateral" and "twisting" directions, but the magnitude of a potential restoring torque applied to the waist was not measured in their study. Thus it remains difficult to determine how much of the reduction in the energetic cost of walking is due to improvements in lateral balance alone. However, since the forces were applied mainly in the lateral direction, it seems reasonable that the major effect of the external LS system was to assist with lateral balance. Although controversial (6), we can presume that arm swing during walking assists with some aspect of lateral balance and based on the study of Ortega et al. (21), one can infer that the active control of lateral balance comprises at most 3% of the net energetic cost of normal walking. Overall, it appears that the energetic cost of maintaining lateral balance during walking depends on both step width adjustments and arm swing.

To date, we do not fully understand whether these balance control mechanisms (i.e., step width adjustments and arm swing) used in human walking are also important for maintaining lateral balance during human running. Our previous study (1) demonstrated a fundamental difference in how humans use step width adjustments to maintain lateral balance in walking vs. running. For example, humans prefer to walk with a moderate step width [8–13% leg length (LL) (13, 21)] but prefer to run with a step width near zero (1). A step width near zero suggests that there may be little need for the active control of maintaining lateral balance in running. We also showed that eliminating arm swing during running increases step width variability (by 9%), which was associated with an increase in energetic cost (by 8%). The increase in step width variability suggests that eliminating arm swing increases the active control required to maintain lateral balance, which may explain the 8% increase in energetic cost. In general, we concluded that arm swing plays an important role in the control of lateral balance during running (1). However, the associated increase in step width variability when arm swing was eliminated did not establish cause and effect between lateral balance and energetic cost. We cannot be certain that the observed decrease

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in lateral balance provides a causal explanation for the increase in energetic cost when running without arm swing. Thus we felt that it was necessary to determine if arm swing contributes to lateral balance in running by carrying out a more direct experimental approach.

In this study, we investigated if there is a link between energetic cost and lateral balance more directly by applying external LS during running. We addressed several questions. *1*) Is there an energetic cost for maintaining lateral balance during human running with arm swing? *2*) Is there an energetic cost for maintaining lateral balance during human running without arm swing? If so, *3*) does the energetic cost of maintaining lateral balance depend on arm swing? To address these questions, we applied external LS during human running with and without arm swing and measured changes in energetic cost and step width variability. We hypothesized that *1*) external LS would reduce the energetic cost and step width variability of running with arm swing at a zero target step width and that *2*) external LS would reduce the energetic cost and step width variability of running without arm swing at a zero target step width. If arm swing assists with lateral balance, we expect that eliminating arm swing would increase the active control required to maintain lateral balance, thus explaining a portion or all of the increase in energetic cost when running without arm swing. Applying external LS while running without arm swing should reduce the active control that is required to maintain lateral balance, resulting in a greater reduction in energetic cost. As such, we further hypothesized that *3*) the reduction in energetic cost and step width variability would be greater when running without arm swing compared with running with arm swing. If the reduction in energetic cost and step width variability is greater when running without arm swing, then we can conclude that arm swing does assist with some aspect of lateral balance. We developed an experimental design that minimized any interaction effects of step width and arm swing. We controlled for step width by having subjects run at a zero target step width. The reasoning for this was twofold. First, humans prefer to run at a step width that is not significantly different from zero (1). Second, we previously found that when humans run without arm swing, they compensate by increasing their step width (1). Adopting a different step width strategy may itself exact an energetic cost.

In addition to understanding the balance control mechanisms, we want to understand if maintaining lateral balance during running incurs a significant energetic cost. A major theme in our laboratory is to understand the biomechanical basis for the energetic cost of human running. Overall, we have pursued an empirical approach whereby assistive devices are directly attached to the body, with the purpose of facilitating a reduction in the energetic cost demand for generating muscular force and/or performing mechanical work. Our task-by-task approach focuses on the biomechanical task that comprise human running. To date, the energetic cost of running can be partitioned into the biomechanical tasks of *1*) body weight support, *2*) propulsion, and *3*) leg swing (9, 20, 22). Quantifying the energetic cost of maintaining lateral balance will help to complete our overall analysis.

MATERIAL AND METHODS

Twelve subjects participated in this study (9 men and 3 women, age = 25.8 ± 3.9 yr, mass = 67.1 ± 10.0 kg, LL = 95.1 ± 5.2 cm; mean ± SD). Prior to data collection, all subjects provided written informed consent as per the University of Colorado Institutional Review Board. Subjects wore their own shoes, were experienced with treadmill walking and running, and were healthy and injury free. Due to some technical difficulties during a running trial, the data for one subject had to be excluded from the final analysis. Thus the group data for running are for $n = 11$ and the group data for walking are for $n = 12$.

Experimental design. Subjects visited our laboratory on two separate days. The first day served as an acclimation period (~30 min) during which subjects practiced walking (2 trials) and running (2 trials) on a force-measuring treadmill (18) with and without external LS. On the second day, subjects began the session with a standing trial during which they stood quietly for 7 min while we measured their rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide production ($\dot{V}CO_2$) using expired gas analysis (ParvoMedics TrueMax2400, Sandy, UT). As described previously (1), we placed reflective markers on the left and right heel, dorsum of the second toe, and lateral mid-foot of each shoe to provide real-time visual feedback (Motion Analysis, Santa Rosa, CA) of foot placement during each trial (Fig. 1*A*). On the monitor positioned directly in front of each subject, we provided a zero target step width by displaying a single virtual line that was defined by reflective markers positioned in the front and back of the force-treadmill. Following the standing trial, subjects performed two randomized trials of walking *1*) with and *2*) without external LS. Both of the trials consisted of walking (1.25 m/s) without arm swing at a zero target step width. The results of the walking part of this experiment are summarized in APPENDIX A and provide experimental validation for our method of applying external LS in this study.

Following the walking trials, subjects performed four randomized trials of running (3.0 m/s). These trials consisted of running with arm swing at a zero target step width *3*) with and *4*) without external LS, and running without arm swing at a zero target step width *5*) with and *6*) without external LS. To walk and run without arm swing, subjects crossed their arms in front of their chest. We measured $\dot{V}O_2$ and $\dot{V}CO_2$ during each 7-min trial and recorded the three-dimensional motions of the feet (100 Hz) and the vertical ground reaction forces (1,000 Hz) during the last 4 min of each trial.

External lateral stabilization system. Our external LS system is similar to those used in previous human walking experiments (12, 14, 21). We applied lateral forces to the subjects via an adjustable waist belt. Lateral forces were applied using nylon rope and a section of latex rubber tubing that acted as a spring element (McMaster-Carr, model #5234K16, Elmhurst, IL). As shown in Fig. 1*B*, a separate piece of nylon rope ran from the other end of a rope ratchet (Carolina North, Kernersville, NC) toward a pulley that connected to the rubber tubing in series with a force transducer (Omega Engineering, model LLCB-50). We used the rope ratchet to stretch the section of latex rubber tubing and thus adjust the effective stiffness of our external LS system. The lateral distance from the subject's waist to the pulley mounted on the wall measured 6 m in length and helped to minimize any anterior/posterior or vertical forces that may have been inadvertently applied to the subject (14, 21). To accommodate differences in subject height, we adjusted the height of the pulley to ensure that we applied forces horizontally in the lateral direction. On the basis of our pilot study ($n = 5$), we found that applying a lateral force of 90 N on each side of the waist yielded an average effective stiffness of ~2,200 N/m and was most comfortable for walking and running.

In situ stiffness measurements. We developed a novel in situ calibration method to ensure that our external LS system applied an effective stiffness that was similar in magnitude for each subject and for each walking and running trial. Before each trial with external LS, subjects stood in a tandem stance along the middle of the treadmill while we applied a lateral stabilizing force of 90 N to each side of the

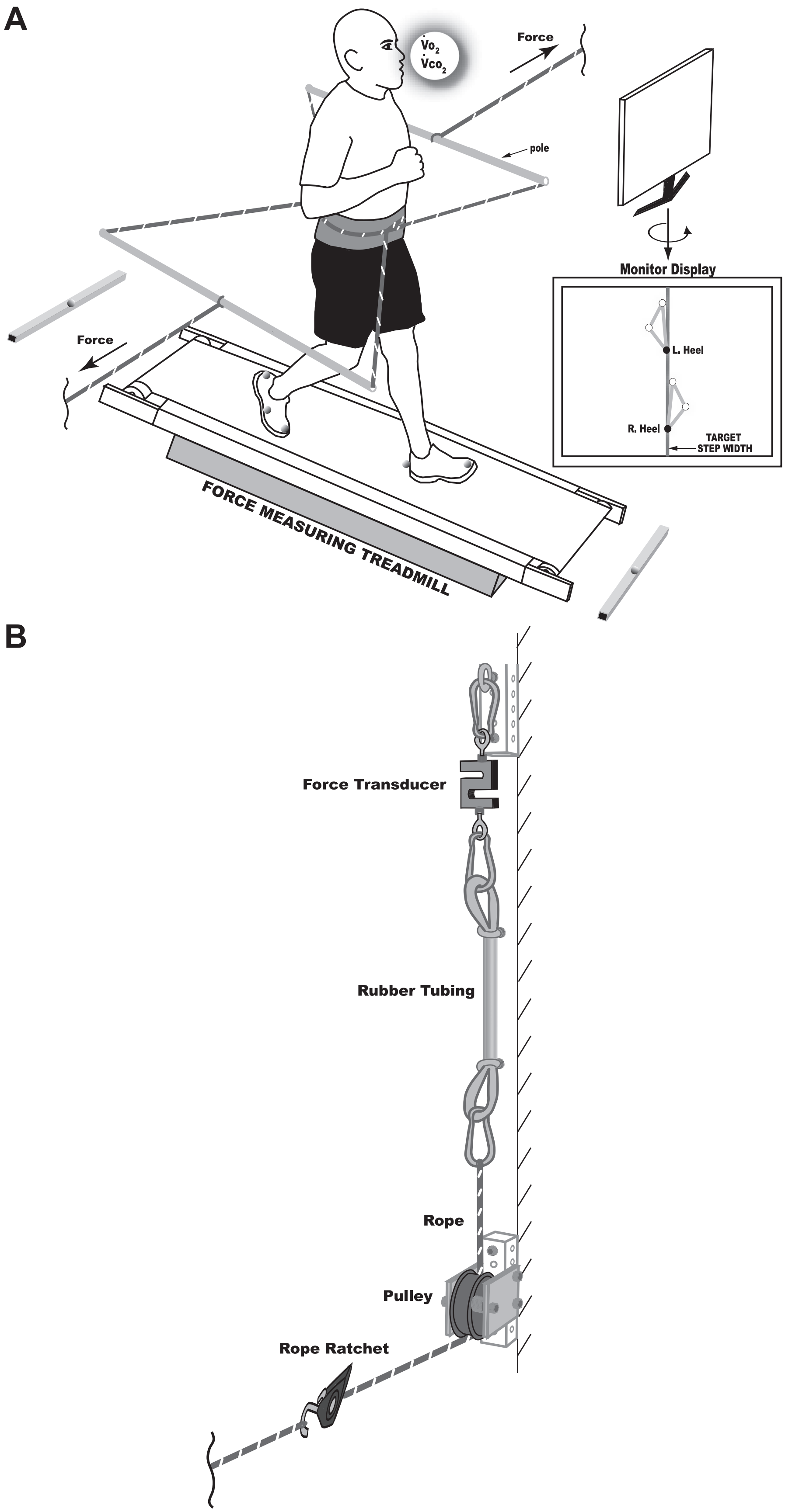

Fig. 1. *A*: we provided external lateral stabilization (LS) and real-time visual feedback of foot placement (monitor providing a top-down view) during all walking and running trials. We applied lateral forces about the waist via lightweight carbon fiber poles. The poles served two purposes. First, we could apply a relatively large elastic force by running a rope (~3.0 m) through the pole (~1 m) and around the waist such that the right pole pulled on the left side of the hip. Similarly, a separate rope (~3.0 m) ran through the left pole (~1 m) and pulled on the right side of the hip. Second, the separation of the poles from the waist allowed for subjects to swing their arms without any restrictions from the stabilizing apparatus. From the middle of each pole, we attached a separate rope that connected to one end of a 1/4-in. rope ratchet (Carolina North, Kernersville, NC), which allowed us to adjust the force applied by the external LS system. To control for step width, we placed reflective markers on the left and right feet and projected a single virtual line along the middle of the treadmill by placing a reflective marker on the front and back of the treadmill. Visual feedback was displayed on a computer monitor (30 × 47 cm^2) positioned in front of each subject (~0.5 m). In addition, we fixed the position of the mouthpiece, which was used to measure rates of metabolic energy, so that subjects had to maintain their position on the force treadmill. This mouthpiece placement also helped minimize any anterior-posterior forces applied by the external LS system (for clarity, mouthpiece configuration not shown). *B*: from the middle of each pole, we connected a separate rope to one end of a 1/4-in. rope ratchet in series with a pulley, a piece of latex rubber tubing (resting length ~0.20 m, outside diameter ~0.02 m, inside diameter ~0.01 m, wall thickness ~0.003 m), and a force transducer mounted to the wall. We hung the force transducer and rubber tubing vertically to reduce any inertial effects of the external LS system.

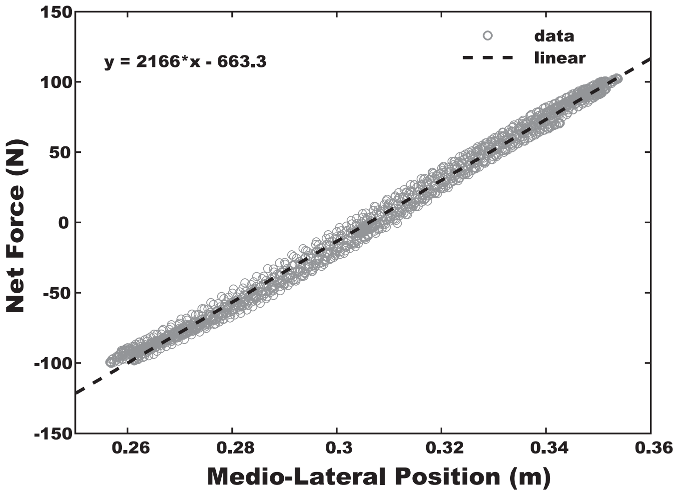

Fig. 2. Net force (N) vs. mediolateral position (m) for a single subject swaying about a virtual line projected on the monitor display. We simultaneously recorded the changes in net force and the position of the sacrum marker for 15 s (○ in gray). To estimate the effective stiffness of the external LS system, we applied a least squares linear regression model to the data (dashed line). The slope (~2,166 N/m) of the linear regression equation represents the effective stiffness of the external LS system.

subject's waist. Using the monitor display (Fig. 1*A*), we projected a single virtual line along the middle of the treadmill while providing subjects visual feedback of a reflective marker placed on the back of the waist harness located at sacrum level. We instructed subjects to sway from side to side about the virtual line for 15 s. During this time, we recorded the mediolateral position of the sacrum marker and changes in the lateral forces detected by the left and right force transducers (LabView, National Instruments, Austin, TX). From the mediolateral position of the sacrum marker and the net force acting on the subject's waist, we calculated the slope of the net force vs. mediolateral position curve to yield the effective stiffness of the external LS system (Fig. 2). Subjects repeated this procedure twice before beginning each walking or running trial to ensure that the appropriate stiffness was applied (Table 1). Our in situ calibration method yielded similar effective stiffness values for all trials. The intraclass R values (all equal to 0.97) indicate that our in situ method for quantifying the effective stiffness of our external LS system provided a highly reliable measure. To complement our in situ method, we replicated the logarithmic decrement method used by previous lateral stabilizing human walking experiments (12, 14, 21). We provide a detailed description of the logarithmic decrement method in APPENDIX B.

Data analysis. For each trial, we calculated the metabolic power from the average $\dot{V}O_2$ and $\dot{V}CO_2$ during the last 3 min (4). We then computed the net metabolic power by subtracting the average metabolic power during standing from the average metabolic power during each running trial. As described previously (1), we calculated the average step width, step width variability, and step frequency during the last 401 consecutive steps that occurred during the last 3 min for each running trial. Step width was defined as the mediolateral (M-L) distance between the right and left heel markers during successive instances of initial contact. We defined step width variability, an indicator of lateral balance, as the standard deviation about the average step width (1, 3, 12, 14, 21). We normalized step width and step width variability by dividing each variable by leg length (trochanter height) and multiplying by 100. Thus step width and step width variability are reported as a percentage of leg length (%LL).

Statistical analysis. We performed repeated-measures ANOVAs with two within-subjects fixed factors (external LS and arm swing). For each dependent variable (net metabolic power, step width, step width variability, and step frequency), this statistical analysis yields *1*) a within-subjects main effect for external LS, *2*) a within-subjects main effect for arm swing, and *3*) an external LS-by-arm swing interaction effect. Following this statistical analysis, we performed planned comparisons between *4*) running with arm swing, with and without external LS and *5*) running without arm swing, with and without external LS using paired *t*-tests. Statistical significance was set at an α level =0.05 (SPSS, Chicago, IL). All values are reported as means ± SE unless noted otherwise.

RESULTS

External LS main effect. When grouping the data across arm swing and no arm swing conditions, external LS during running significantly reduced the demand for net metabolic power by 2.0% (11.56 ± 0.22 W/kg without external LS vs. 11.34 ± 0.27 W/kg with external LS, $F(1,10) = 6.168$; $P = 0.032$; Fig. 3*A*) and also significantly reduced step width variability by 12.3% (1.92 ± 0.13 %LL without external LS vs. 1.69 ± 0.12 %LL with external LS, $F(1,10) = 12.578$; $P = 0.005$; Fig. 3*C*). However, external LS did not affect step width (1.91 ± 0.32 %LL without external LS vs. 1.82 ± 0.26 %LL with external LS, $F(1,10) = 0.295$; $P = 0.599$; Fig. 3*B*) or step frequency (2.93 ± 0.07 Hz without external LS vs. 2.92 ± 0.06 Hz with external LS, $F(1,10) = 0.041$; $P = 0.843$; Fig. 3*D*).

Arm swing main effect. When grouping the data across conditions of without and with external LS, eliminating arm swing during running significantly increased the demand for net metabolic power by 7.6% (11.03 ± 0.27 W/kg with arm swing vs. 11.87 ± 0.23 W/kg without arm swing, $F(1,10) = 34.186$; $P < 0.001$; Fig. 3*A*) but did not significantly affect step width (1.90 ± 0.30 %LL without external LS vs. 1.83 ± 0.28 %LL with external LS, $F(1,10) = 0.323$; $P = 0.582$; Fig. 3*B*) or step width variability (1.80 ± 0.15 %LL without external LS vs. 1.80 ± 0.13 %LL with external LS, $F(1,10) = 0.001$; $P = 0.975$; Fig. 3*C*). Eliminating arm swing during running significantly increased step frequency by 1.7% (2.90 ± 0.06 Hz with arm swing vs. 2.95 ± 0.07 Hz without arm swing, $F(1,10) = 6.419$; $P = 0.030$; Fig. 3*D*).

Table 1. *Effective stiffness of the external LS system and intraclass R values measured for trial 1 and trial 2*

	Prior to Walking		Prior to Running		Prior to Running	
	No arm swing		Arm swing		No arm swing	
Trial	*1*	*2*	*1*	*2*	*1*	*2*
Stiffness, N/m	2,164 (261)	2,194 (291)	2,250 (270)	2,301 (301)	2,243 (232)	2,281 (223)
*i*ntraclass R	0.97*		0.97†		0.97‡	

Values are means ± SD. LS, lateral stabilization. *F test not significant ($P = 0.207$) between the stiffness measured at *trials 1* and *2*. †F test not significant ($P = 0.061$) between the stiffness measured at *trials 1* and *2*. ‡F test not significant ($P = 0.073$) between the stiffness measured at *trials 1* and *2*. When combining all trials, the F test was not significant ($P = 0.70$) and the intraclass R = 0.95.

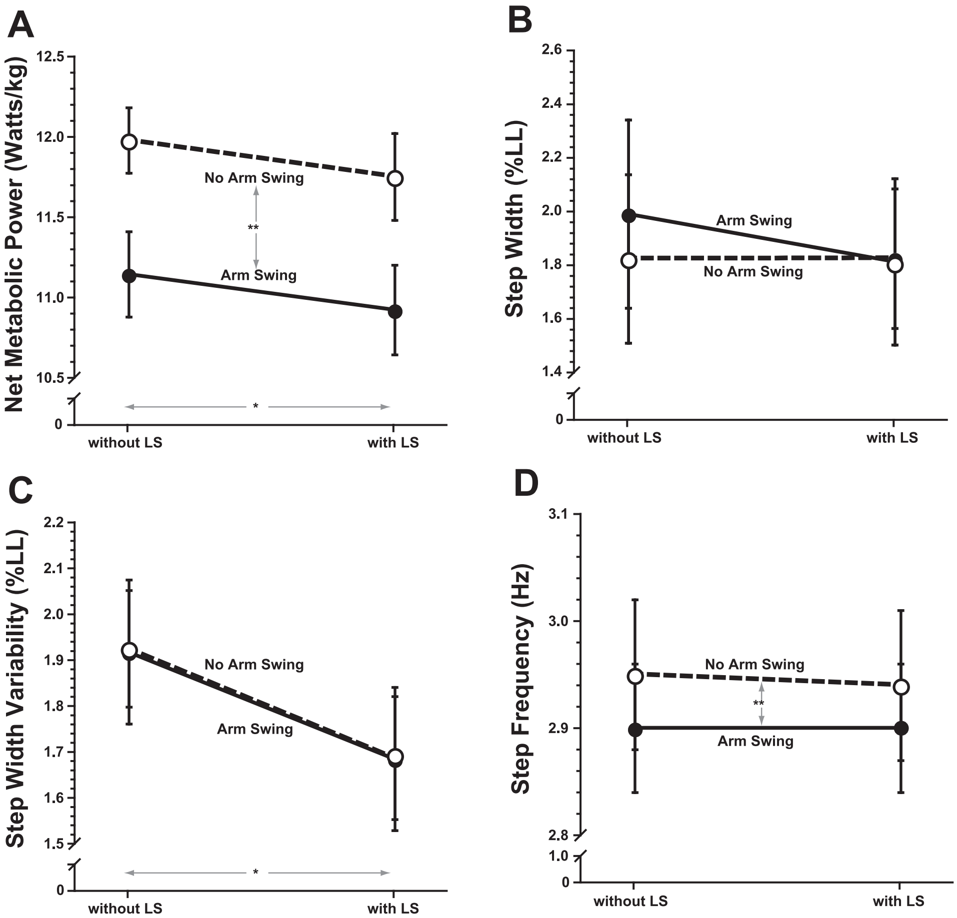

Fig. 3. Net metabolic power (*A*), step width (*B*), step width variability (*C*), and step frequency (*D*) while running without and with external LS, both with and without arm swing (n = 11; means ± SE). When provided with external LS (solid and dashed lines), subjects significantly reduced net metabolic power demand (*significant external LS effect, P = 0.032) and reduced step width variability (*significant external LS effect, P = 0.005), but did not change step width or step frequency. When eliminating arm swing (No Arm Swing, ○; Arm Swing, ●), subjects significantly increased net metabolic power demand (**significant arm swing effect, P < 0.001) and step frequency (**significant arm swing effect, P = 0.030), but did not change step width or step width variability. The absence of a significant interaction effect for net metabolic power, step width, step width variability, and step frequency indicates that external LS resulted in a similar reduction in net metabolic power and step width variability when running with or without arm swing. Note that by defining external LS and arm swing as two within-subjects fixed factors in a repeated-measures ANOVA, the graphical layout of the data allows interpretation of any main and/or interaction effects on net metabolic power, step width, step width variability, or step frequency.

External LS-by-arm swing interaction effect. There were no significant interaction effects for net metabolic power (F(1,10) = 0.002; P = 0.964; Fig. 3*A*), step width (F(1,10) = 0.276; P = 0.611; Fig. 3*B*), or step width variability (F(1,10) = 0.005; P = 0.943; Fig. 3*C*), indicating that regardless of running with or without arm swing, external LS resulted in a similar reduction in net metabolic power and step width variability. For example, when running with arm swing, external LS significantly reduced the demand for net metabolic power by 2.0% (11.15 ± 0.27 W/kg without external LS vs. 10.92 ± 0.28 W/kg with external LS, P = 0.006) and reduced step width variability by 12.2% (1.92 ± 0.16 %LL without external LS vs. 1.68 ± 0.16 %LL with external LS, P = 0.006), but did not change step width (1.99 ± 0.35 %LL without external LS vs. 1.81 ± 0.31 %LL with external LS, P = 0.566). When running without arm swing, external LS significantly reduced the demand for net metabolic power by 1.9% (11.98 ± 0.20 W/kg without external LS vs. 11.75 ± 0.25 W/kg with external LS, P = 0.027) and reduced step width variability by 12.4% (1.93 ± 0.13 %LL without external LS vs. 1.69 ± 0.13 %LL with external LS, P = 0.004), but did not change step width (1.82 ± 0.31 %LL without external LS vs. 1.82 ± 0.26 %LL with external LS, P = 0.994). There was no significant interaction effect for step frequency (Fig. 3*D*), indicating that external LS did not change step frequency when running with (2.90 ± 0.06 Hz without external LS vs. 2.90 ± 0.06 Hz with external LS, P = 0.735) or without (2.95 ± 0.07 Hz without external LS vs. 2.94 ± 0.07 Hz with external LS, P = 0.709) arm swing. Overall, our data reveal that the reduction in net metabolic power and step width variability with external LS did not depend on arm swing. In addition, when provided with external LS, the reduction in net metabolic power and step width variability during running were not affected by changes in step width or step frequency.

Standing metabolic power and respiratory exchange ratio values. The average metabolic power and respiratory exchange ratios (RER) during quiet standing were 1.54 ± 0.11 W/kg and 0.83 ± 0.04, respectively. The average RER value across all walking and running trials was 0.87 ± 0.01 (values expressed as mean ± SD). RER values <1.0 indicate that metabolic energy was provided primarily by aerobic metabolism (5).

DISCUSSION

Overall, our experimental findings support our first and second hypothesis, suggesting that in running, external LS improves lateral balance and reduces energetic cost. We found that external LS reduced energetic cost (by 2%) and step width variability (by 12%) to a similar extent while running with or without arm swing. As expected, eliminating arm swing increased the energetic cost of running. However, applying external LS when running without arm swing did not reduce energetic cost or step width variability to a greater extent, thus we reject our third hypothesis. We conclude that eliminating arm swing does not increase the active control required to maintain lateral balance and the increase in energetic cost must be due to some other aspects of balance control. Our study demonstrates that *1*) maintaining lateral balance comprises ~2% of the net energetic cost of human running and that *2*) arm swing is not an important mechanism for maintaining lateral balance during running.

When running at a zero step width with and without arm swing, we found that external LS reduced net metabolic power by 0.23 and 0.23 W/kg, respectively. We also found that external LS reduced step width variability by 0.24 and 0.24 %LL, respectively. Step width was similar when running with and without external LS, which is evidence that our method for controlling step width was effective. Our results demonstrate that the reduction in the magnitude of step width variability coincided with a reduction in net metabolic power with external LS, which was independent of changes in average step width. A reduction in step width variability with external LS indicates a reduction in the active control needed to maintain lateral balance (12, 14). Thus the reduction in step width variability provides the best explanation for the reduction in net metabolic power while running with external LS.

Surprisingly, we found that arm swing had no effect on the cost of maintaining lateral balance during human running. The energetic cost of maintaining lateral balance was nearly the same (2.0% with arm swing vs. 1.9% without arm swing) when arm swing was eliminated during human running. Why did external LS not counteract some portion of the 7.5% increase (11.15 W/kg with arm swing vs. 11.98 W/kg without arm swing) in energetic cost when running without arm swing? It is well known that arm swing in human walking and running plays the major role in counteracting the angular momentum generated by the lower body about the vertical axis, resulting in total whole body angular momentum about the vertical axis that is relatively small and fluctuates about zero (7, 16, 17). When considering the angular momentum about the vertical axis, Hinrichs (17) showed that at a slightly faster running speed of 3.8 m/s, the motion of arm swing constitutes ~80% of the total angular momentum generated by the upper body. Since running without arm swing prevents the arms from generating angular momentum about the vertical axis, it seems reasonable that subjects would compensate to ensure that the upper body angular momentum counteracts the lower body angular momentum about the vertical axis.

If the cost of maintaining lateral balance does not increase when eliminating arm swing, then what is the biomechanical explanation for the increased cost of running without arm swing? One possibility is that eliminating arm swing during running increases the free moment about the vertical axis, requiring greater leg muscular activation and a more costly strategy for maintaining balance. However, Miller et al. (19) found that eliminating arm swing during running does not increase the free vertical moment. An alternative explanation may involve compensatory strategies in torso rotation when arm swing is eliminated. When arm swing was eliminated during the running trials, we observed a tendency for subjects to increase and/or modify torso rotation, which may explain why external LS did not counteract some of the increase in energetic cost. Increasing and/or modifying torso rotation may help to counteract the vertical angular momentum generated by the swinging legs, an idea proposed by Miller and colleagues (19). This compensatory strategy may involve greater activation of the trunk muscles and thus incur an energetic cost. At this time, these explanations are speculative but our future efforts will aim to identify the underlying mechanism(s) that increase the energetic cost of running without arm swing.

In our previous study (1), we perturbed lateral balance by having subjects run with step widths other than preferred. Those data suggest that running at step widths other than preferred increased the need for the active control of lateral balance, thus incurring a greater energetic cost. In this study, we reduced the need for the active control of lateral balance by providing external LS during running and demonstrated reductions in both step width variability and energetic cost. Taken together, our findings lend further support to our idea that there is a link between energetic cost and lateral balance in human running (1).

Although our external LS device and experimental design allowed us to address our hypotheses, there are some limitations of this study. Our main assumption with our method of applying external LS is that our device exclusively assists with lateral balance but our device may also resist twisting motions about the waist. Thus it is possible that any reduction in energetic cost with external LS may be due to stabilizing pelvic/trunk motion. It is also possible that external LS may have reduced the need to swing the arms while running. Measurements of torso and arm swing motion could provide insights into whether external LS assisted with upper body control. Due to our real-time visual feedback method of foot placement, it was not feasible to measure torso and/or arm swing motion because placing reflective markers on the upper body would interfere with the subject's ability to focus on running along the single line that was provided by the monitor display (Fig. 1*A*).

Another potential limitation is that we did not control for step frequency across the running trials. However, step frequency was similar when running with or without external LS. Eliminating arm swing during running increased step frequency by 1.7%, and previous evidence (8) indicates that a 1.7% increase in step frequency would increase the rate of oxygen consumption by <0.5%. Although our experimental design eliminated any interaction effects between step width and arm swing during running, we acknowledge that controlling for step width could be considered a limitation. In a future experiment, it may be worthwhile to measure changes in step width variability and net metabolic power while running with arm swing at the preferred step width. Since humans prefer to run at a step width not significantly different from zero (1), we predict that applying external LS while running with arm swing at the preferred step width would yield a similar reduction in energetic cost and step width variability as observed in this study.

In summary, external LS reduced energetic cost (by 2%) and step width variability (by 12.3%) when running with or without arm swing. We infer that the percent reduction in energetic cost and step width variability while running with external LS reflects the energetic cost of maintaining lateral balance. Thus maintaining lateral balance comprises ~2% of the net energetic cost of human running. Furthermore, eliminating arm swing during running had no effect on the energetic cost of maintaining lateral balance. In conclusion, our data suggest that humans use step width adjustments, not arm swing, as the primary mechanism for maintaining lateral balance during running.

APPENDIX A: WALKING EXPERIMENTS TO VALIDATE OUR METHOD OF APPLYING EXTERNAL LATERAL STABILIZATION

We compared the energetic cost and lateral balance of walking with and without external LS. During these trials, subjects walked at a zero target step width with and without arm swing so that we could compare our findings to previous walking experiments (12, 14). Similar to Donelan et al. (14), we hypothesized that external LS would reduce the energetic cost and step width variability of walking without arm swing at a zero target step width.

For the walking trials, we estimated instances of initial contact by determining the time when the heel marker reached the maximum position in the forward direction (2). We defined this instant in time as the beginning of each step and extracted the mediolateral position of the heel marker. Using these data, we calculated average step width, step width variability, and step frequency during the last 401 consecutive steps that occurred during the last 3 min of each walking trial. For each dependent variable, we compared conditions of walking without and with external LS using one-sided paired *t*-tests (Table 2).

When walking without arm swing and at a zero target step width, external LS significantly reduced net metabolic power by 5.5% (3.13 vs. 2.96 W/kg; $P = 0.018$) and significantly reduced step width variability by 13.8% (1.16 vs. 1.00 %LL; $P = 0.007$). The average step width did not significantly change with external LS (3.83 vs. 3.70 %LL; $P = 0.568$). Lastly, external LS increased step frequency by 1.6% (1.78 vs. 1.81 Hz; $P = 0.017$).

Similar to previous walking experiments (12, 14), we found that external LS reduced energetic cost and step width variability while walking without arm swing at a zero target step width. When walking at a zero step width and without arm swing, we found that external LS reduced net metabolic power by an average of 0.17 W/kg. This absolute reduction in net metabolic power with external LS was less than that observed by Donelan et al. (14) and Dean et al. (12), who reported reductions in net metabolic power of 0.36 and 0.40 W/kg when providing external LS under comparable walking conditions. We also found that external LS reduced step width variability by 0.16% LL while Dean et al. (12), who used an equivalent measure of step width variability, reported a 0.50 %LL reduction in step width variability. However, simply reporting the mean difference in the effect of external LS across studies can lead to an incorrect interpretation, as they do not take into account the amount of variability observed in each study. To compare the effect of external LS across walking studies, we computed the effect size, also known as Cohen's *d*, as suggested by Dunlap et al. (15) and interpret these values based on the classification scheme (small: $d = 0.20$; medium: $d = 0.50$; large: $d = 0.80$) presented by Cohen (10).

When computing the effect of external LS on reducing the demand for net metabolic power, we find that our effect size of 0.66 falls within the effect size range of 0.58 and 0.75 [values derived from the data of Donelan et al. (14) and Dean et al. (12), respectively]. In terms of step width variability, we find that our effect size of 0.80 is slightly less than the effect size of 0.88 as derived from Dean et al. (12). Thus we conclude that our external LS system represents a medium-to-large effect size, comparable to those reported by Donelan et al. (14) and Dean et al. (12).

Finally, we found a small, but significant increase in step frequency (1.7%) while walking with external LS compared with walking without external LS. On the basis of the findings from Umberger and Martin (24), it appears that stride frequency changes less than 5% do not significantly increase net metabolic power. Thus it is unlikely that the 1.7% increase in step frequency observed in this study had a significant effect on net metabolic power. Overall, our walking results help validate our effects of applying external LS during human running.

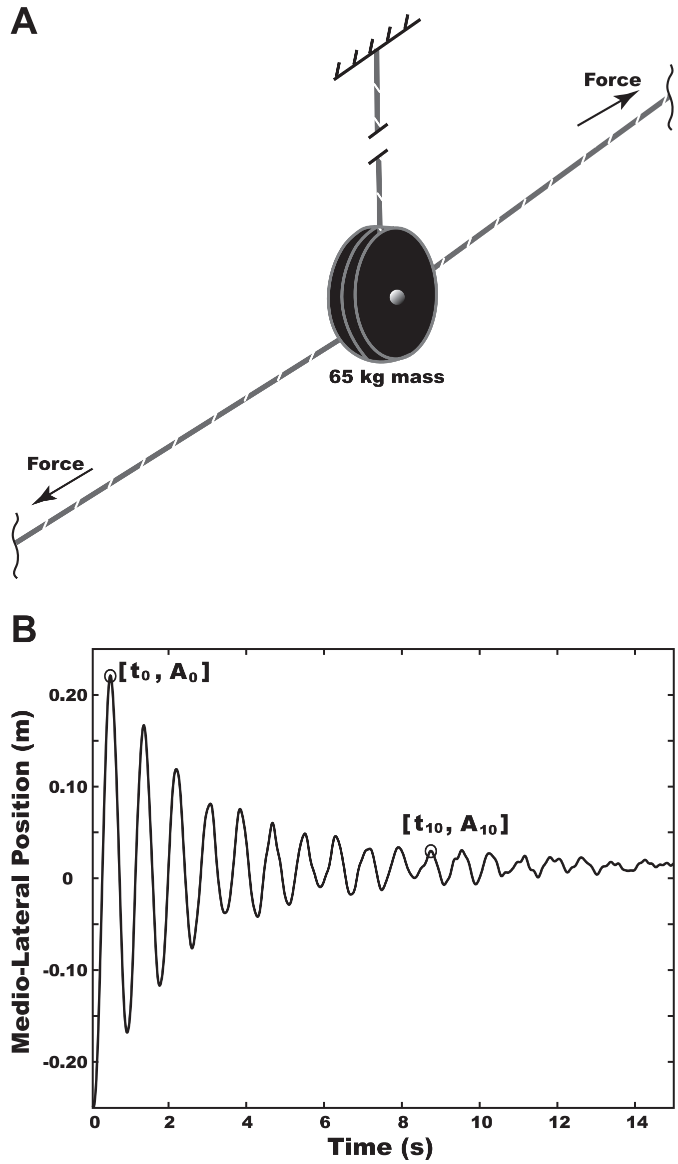

Fig. 4. Experimental set-up (*A*) and the mediolateral decaying oscillation of a known mass attached to the external LS system (*B*). Assuming a one-degree of freedom oscillation, we can compute the effective stiffness, damping ratio, and damping of the system (see APPENDIX B).

Table 2. *Net metabolic power, respiratory exchange ratio, step width, step width variability, and step frequency*

	Walking (No Arm Swing)		
	Without LS		With LS
Net metabolic power, W/kg	3.13 ± 0.07	**P = 0.018**	2.96 ± 0.08
Step width, % LL	3.83 ± 0.33	P = 0.568	3.70 ± 0.445
Step width variability, % LL	1.16 ± 0.05	**P = 0.007**	1.00 ± 0.06
Step frequency, Hz	1.78 ± 0.04	**P = 0.017**	1.81 ± 0.04

Values are means ± SE. LL, leg length. For each comparison that yielded statistical significance, *P* values <0.05 are denoted in bold.

APPENDIX B: LOGARITHMIC DECREMENT METHOD FOR MEASURING THE EFFECTIVE STIFFNESS OF OUR EXTERNAL LATERAL STABILIZATION SYSTEM

Computing Effective Stiffness (k: N/m), Non-Dimensional Damping Ratio (ζ), and Damping (c: N·s/m)

Similar to previous experiments (12, 14, 21), we modeled our external LS system as a second order damped oscillator model. Supported vertically by a long rope, we attached a known mass (65 kg) to the external LS system, displaced the mass from its equilibrium position (~0.05 m) and measured the oscillation of a reflective marker placed on the mass using our motion capture system (Fig. 4, *A* and *B*). The equation for the second order damped oscillator model is defined as follows:

$$\ddot{x} + 2\xi\omega_n\dot{x} + \omega_n^2 x = 0$$

where

$$\omega_n = \sqrt{\frac{\mathrm{k}}{\mathrm{m}}} = \text{natural frequency},\ \xi = \frac{\mathrm{c}}{2\mathrm{m}\omega_n}$$

where c is damping, m is mass, and k is stiffness.

From the logarithmic decrement method, we defined δ as

$$\delta = \frac{1}{10}\ln\left(\frac{\mathrm{A}_0}{\mathrm{A}_{10}}\right)$$

and the natural period ($\mathrm{T_d}$) between 10 cycles as

$$\mathrm{T_d} = \frac{1}{10}\left(t_{10} - t_0\right)$$

From our data (Fig. 4*B*), we computed the changes in amplitude (A_0 and A_{10}) and the natural period ($\mathrm{T_d}$) as follows: $t_0 = 0.5$ and $\mathrm{A}_0 = 0.0221$; $t_{10} = 8.75$ and $\mathrm{A}_{10} = 0.0030$.

Our custom Matlab program yielded the following values: k = 3818.72 (N/m); c = 31.84 (N·s/m); damping ratio = 0.032.

As demonstrated, the logarithmic decrement method yielded an effective stiffness value of ~ 3,800 N/m and a damping value of ~32 N·s/m. Note that the logarithmic decrement method is useful for estimating the effective stiffness of the external LS system as it responds freely (i.e., without external forcing). It seems reasonable that our in situ calibration method would yield a different stiffness value because the external LS system responds differently when a subject applies a force that drives the external LS system away from its equilibrium position. In general, we prefer our in situ calibration method as we could immediately estimate the effective stiffness applied to the subject from trial to trial and ensure the rubber tubing was not damaged.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the authors.

AUTHOR CONTRIBUTIONS

Author contributions: C.J.A. and R.K. conception and design of research; C.J.A. performed experiments; C.J.A. analyzed data; C.J.A. and R.K. interpreted results of experiments; C.J.A. prepared figures; C.J.A. drafted manuscript; C.J.A. and R.K. edited and revised manuscript; C.J.A. and R.K. approved final version of manuscript.

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