

**ENERGETIC COST AND BALANCE CONTROL MECHANISMS
IN HUMAN LOCOMOTION**

by

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A thesis submitted to the
Faculty of the Graduate School of the
University of Colorado in partial fulfillment
of the requirement for the degree of
Doctor of Philosophy
Department of Integrative Physiology

2012

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Energetic cost and balance control mechanisms in human locomotion
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IRB protocol #
0606.29
11-0112

Arellano, Christopher J. (Ph.D., Integrative Physiology)
Energetic Cost and Balance Control Mechanisms in Human Locomotion
Thesis directed by Associate Professor Rodger Kram

The first three chapters of my dissertation focuses on human running emphasizing step width and arm swing as primary balance control strategies. The fourth chapter tests the hypothesis that arm swing during human walking is primarily passive.

In my first study, I found that when subjects ran (3.0 m/s) at step widths other than their preferred narrow step width or without arm swing, both net metabolic power demand and step width variability (indicator of lateral balance) increased. I interpret greater step width variability as a decrease in lateral balance. My findings suggest that humans prefer to run with a narrow step width and swing their arms so as to minimize energetic cost and improve lateral balance.

In my second study, I found that when running (3.0 m/s) with or without arm swing, external lateral stabilization (LS) results in similar reductions in net metabolic power (~2.0%) and step width variability (~12.0%). I 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 (~8%), arm swing does not appear to assist with lateral balance.

In my third study, I found that compared to non-amputees, sprinters with trans-tibial amputations ran with greater step width and medio-lateral (M-L) foot placement variability, indicating that they have greater challenges with maintaining lateral balance. At faster running speeds up to maximum sprint speed, variability of both step width and M-L foot placement increased in all sprinters, indicating progressive decreases in lateral balance.

In my fourth study, I quantified arm swing amplitudes and shoulder muscle activity while subjects walked with 1) their biological arms and 2) with free-swinging, anthropomorphic,

passive mechanical arms. I found that passive mechanical arm swing resembled the behavior of a horizontally driven pendulum, reaching its largest amplitude as step frequency approached the arm's natural frequency; however, the swinging amplitudes of the passive mechanical arms was much less than the swinging amplitudes of the biological arms. My findings demonstrate that arm swing during human walking is a hybrid system comprising active muscular actuation and passive pendulum dynamics.

DEDICATION

To my wife, Ruth María López, and our soon to be born son.

ACKNOWLEDGMENTS

This dissertation would have not been possible without the funding support from the NASA Harriett G. Jenkins pre-doctoral fellowship program and the Colorado Diversity Initiative. I would also like to thank the following individuals who have supported me through my doctoral program:

Professor Rodger Kram for his continuous support, devotion, and help in shaping my philosophy of science.

Professor Alaa A. Ahmed for her continuous support and allowing me to use her laboratory equipment to collect the data for my dissertation.

Professor Glenda M. Russell and the dissertation support group for their valuable encouragement and support.

Professors Roger M. Enoka, William C. Byrnes, and David M. Klaus for serving on my dissertation committee and for their helpful insights and suggestions.

Members of the Locomotion Laboratory, Neuromechanics Laboratory, Neurophysiology Laboratory, and Applied Exercise Science Laboratory.

Professors Dale Mood and Matthew B. McQueen for helpful comments and assistance with my statistical analyses.

CONTENTS

1. INTRODUCTION	1
The Energetic Cost of Human Walking and Running	1
Walking	4
Mechanics	4
Simple Inverted Pendulum Model	4
Dynamic Walking Model	9
The Energetic Cost of Walking: Task-by-Task	14
Body Weight Support	15
Propulsion	17
Leg Swing	18
Lateral Balance	18
Walking Summary	19
Running	21
Mechanics	21
Performing Mechanical Work	21
Spring Mass Model	24
Generating Muscular Force	27
Energetic Cost of Running: Task-by-Task	28
Body Weight Support	29
Propulsion	30
Leg Swing	30
Lateral Balance	31
Running Summary	32
Overall Conclusion	34
Arrangement of Dissertation	36
2. THE EFFECTS OF STEP WIDTH AND ARM SWING ON ENERGETIC COST AND LATERAL BALANCE DURING HUMAN RUNNING	37
A. Abstract	37
B. Introduction	38
C. Methods	42
D. Results	45
E. Discussion	51
3. THE ENERGETIC COST OF MAINTAINING LATERAL BALANCE DURING HUMAN RUNNING	55
A. Abstract	55
B. Introduction	56
C. Methods	60
D. Results	69
E. Discussion	73
APPENDIX A	77
APPENDIX B	81
4. LATERAL BALANCE AND FOOT PLACEMENT ACROSS RUNNING SPEEDS: A COMPARISON OF SPRINTERS WITH AND WITHOUT TRANS-TIBIAL AMPUTATIONS	

A. Abstract	85
B. Introduction	86
C. Methods	87
D. Results	95
E. Discussion	103
5. ARM SWING DURING HUMAN WALKING: ACTIVE AND PASSIVE CONTRIBUTIONS TO A HYBRID SYSTEM	107
A. Abstract	107
B. Introduction	108
C. Methods	113
D. Results	117
E. Discussion	120
APPENDIX C	127
APPENDIX D	131
6. SUMMARY	135
1. The effects of step width and arm swing on energetic cost and lateral balance during running	135
2. The energetic cost of maintaining lateral balance during human running	135
3. Lateral balance and foot placement across running speeds: A comparison of sprinters with and without trans-tibial amputations	136
4. Arm swing during human walking: Active and passive contributions to a hybrid- system	137
Future Directions	138
7. REFERENCES	140

LIST OF TABLES

Table 2.1 Data for subjects running (3 m/s) at the various conditions of target step widths, Arms, and No Arms (mean $\pm$ SEM).....	47
Table 3.1 The effective stiffness (mean $\pm$ SD) of the external LS system and intraclass R values measured for trial 1 and trial 2.....	66
Table 3.2 Net metabolic power, respiratory exchange ratio, step width, step width variability, and step frequency (mean $\pm$ SEM).	80
Table 4.1 Anthropometric and biomechanical characteristics for each unilateral (UL), bilateral (BL), and non-amputee (NA) sprinter. For sprinters with unilateral and bilateral trans-tibial amputations, total mass includes prosthesis mass.....	89
Table 4.2 Mean, standard deviation, and MANOVA results for comparisons of step width and step width variability (data collapsed across speed).....	97
Table 4.3 Mean, standard deviation, and MANOVA results for comparisons of M-L foot placement and M-L foot placement variability (data collapsed across speed).	101
Table 5.1 Amplitude of biological and passive mechanical arm swing at each step frequency condition ($n = 10$; mean $\pm$ SEM).	119
Table 5.2 Normalized EMG amplitude of the posterior and anterior deltoid while swinging the biological arms at each step frequency condition ($n = 10$; mean $\pm$ SEM).....	119
Table 5.3 Natural resonant frequency of biological and passive mechanical arms (mean $\pm$ SD).	128
Table 5.4 Amplitude of the horizontal motion of the reflective marker placed the shoulder joint and pivot point of the biological and passive mechanical arms during each step frequency condition (mean $\pm$ SEM).....	132
Table 5.5 Amplitude of biological and passive mechanical arm swinging as predicted from a horizontally driven pendulum subjected to variable step frequencies during walking.....	134

LIST OF FIGURES

Figure 1.1 (A) $\dot{V}O_2$ (ml O_2 /kg/min) and (B) Cost of Transport (ml O_2 /kg/meter) as a function of speed (m/s). The curves are developed from the classic data published by Cotes and Meade (1960) and Margaria et al. (1963), respectively..... 3

Figure 1.2 In walking, the kinetic energy (KE) fluctuations are out of phase with the gravitational potential energy (GPE) fluctuations. This out of phase behavior allows for inverted pendulum like exchange of mechanical energy. 6

Figure 1.3 Mechanical power curve demonstrating the rate of positive mechanical work performed by the trailing leg (P_{trail}) and the rate of negative mechanical work performed by the leading leg (P_{lead}) as quantified by the individual limbs method (ILM; Donelan et al. 2002). The inverted pendulum approach (Cavagna et al., 1977), also known as the combined limbs method (CLM), mistakenly cancels the simultaneous rate of positive and negative work performed during double support. The gray shaded areas underneath the solid line represent the CLM and the area underneath the dashed lines represents the ILM. The data are calculated from the ground reaction forces of two separate force platforms and represent a single step while a subject walked over-ground at ~ 1.25 m/s. 8

Figure 1.4 (A) The dynamic walking model predicts an energetic cost for the mechanical work performed during the step-to-step transition. The two leg forces F_1 and F_2 redirect the COM velocity from a downward and forward direction to an upward and forward direction, sweeping an angle 2α . (B) Experimental measurements of humans walking at constant step frequency but increasing step length support the models prediction. (C) Metabolic rate also increases with S^4 and in proportion to COM work rate. Figure and caption adapted from Kuo (2007)..... 11

Figure 1.5 (A) The dynamic walking approach predicts an energetic cost for the forced motion of the leg. (B) The energetic cost may be estimated from an isolated leg swinging experiment, in which the leg is moved about the hip with roughly the same torque and angle amplitudes as walking. (C) Experimental measurements show that the cost of leg swinging increases sharply, and in approximate agreement with the F^4 prediction. Figure and caption adapted from Kuo (2007). 13

Figure 1.6 The energetic cost of walking is partitioned into distinct phases or biomechanical tasks as outlined by the three different approaches, i.e. the dynamic walking approach, the forward dynamic simulation approach, and the task-by-task approach..... 20

Figure 1.7 During running, the kinetic energy (KE) and gravitational potential energy (GPE) fluctuations of the COM are in-phase during the stance phase. Figure adapted from Farley and Ferris (1998). 22

Figure 1.8 (A) Spring-mass model of human running. The model consists of a mass and a single leg spring (which connects the foot and COM). This figure depicts the model at the beginning of the stance phase (left most position), at the middle of the stance phase (leg spring is oriented vertically) and at the end of the stance phase (right most position). The leg spring has an initial

length, L_0 , at the beginning of the stance phase, and ΔL represents its maximal compression. The dashed spring-mass model shows the length of the uncompressed leg spring. Thus, the difference between the length of the dashed leg spring and the maximally compressed leg spring represents the maximum compression of the leg spring, ΔL . The downward vertical displacement of the mass during the stance phase is represented by Δy and is substantially smaller than ΔL . Half of the angle swept by the leg spring during the ground contact time is denoted by θ . Figure and caption adapted from Farley and Gonzalez (1996). (B) This figure illustrates the fluctuations in the kinetic energy (KE), gravitational potential energy (GPE), and elastic energy (EE). During the 1st half of stance, the KE and GPE of the COM compress the linear leg spring and elastic energy (EE) is stored within the spring. During the 2nd half of stance, the linear leg spring recoils and releases the EE, which is converted back into KE and GPE. 25

Figure 1.9 The energetic cost of running is partitioned into the biomechanical tasks as outlined by the cost of generating force approach and the task-by-task approach. 33

Figure 2.1 Rear-view of foot placement relative to the midline of the body during human running. The vertical GRF can reach values of 2-3 times body weight and is the largest component making up the resultant GRF during human running. Placing the foot along the midline of body aligns the vertical ground reaction force close to the whole body center of mass (COM), effectively reducing the moment generated about the COM along the A-P axis. Modeled after McClay and Cavanagh (1994). 40

Figure 2.2 Real-time visual feedback of foot placement (monitor providing a top-down view) during the target step width conditions. We placed reflective markers on the left and right feet. Adjustable markers defined the left and right virtual lines. The distance between the virtual lines was set as a percentage of leg length (%LL). For a target step width of 0 %LL, we projected a single virtual line along the middle of the treadmill. Visual feedback was displayed on a computer monitor (30 x 47 cm) positioned in front of each subject (~ 0.5 m). 43

Figure 2.3 Net metabolic power ($n = 10$; mean $\pm$ SEM) versus step width (% LL). Arms denotes running at the preferred step width with arm swing. No Arms denotes running at the preferred step width without arm swing. Quotation marks indicate that subjects were unable to match the target step widths (“%LL”). The data demonstrate that running with step widths other than the preferred step width increases net metabolic power demand. Similarly, running without arm swing increases net metabolic power demand indicating that arm swing is important for reducing energetic cost. 48

Figure 2.4 Step width (%LL) from step-to-step (#) during the preferred (Arms, open circles) and 25 %LL condition (filled circles) for a single subject. The broken lines (white) represent the average step width and the shaded regions (gray) represent the ($\pm$) standard deviation about the average step width. Note the larger standard deviation in the 25%LL condition as compared to the preferred condition. For clarity, the other target step width conditions are not shown. 49

Figure 2.5 Step width variability ($n = 10$; mean $\pm$ SEM) versus step width (% LL). The data demonstrate that running with step width other than the preferred step width increases step width variability. Similarly, running without arm swing increases step width variability indicating that arm swing improves lateral balance. 50

Figure 3.1 We provided external 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 (in blue). 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 subject's 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 ¼ inch rope ratchet (Carolina North Mfg, 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 x 47 cm²) 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). 63

Figure 3.2 From the middle of each pole, we connected a separate rope to one end of a ¼ inch 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. 64

Figure 3.3 Net Force (N) versus medio-lateral 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 seconds (open circles 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. 67

Figure 3.4 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; mean±SEM). 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, open circles; Arm Swing, filled circles), 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..... 72

Figure 3.5 Experimental set-up (A) and the medio-lateral 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 text). 82

Figure 4.1 (A) Rear view of a sprinter with a left-side trans-tibial amputation running on a force-measuring treadmill. (B) Side view of the running-specific prosthesis. For measurements of step width and M-L foot placement relative to the midline, we placed reflective markers at the distal end of the running-specific prosthesis(es) and heel of the shoe for the UL. 91

Figure 4.2 Representative foot placement trajectories while running at 3 m/s. For a sprinter without trans-tibial amputations (Non-Amputee), a sprinter with a unilateral trans-tibial amputation (Unilateral), and a sprinter with bilateral trans-tibial amputations (Bilateral), the position of the reflective markers (open circles) indicates the instances of initial contact for each step while running. Note that foot placement of the unilateral (AL) and bilateral (LAL and RAL) sprinter are depicted by the “foot print” based on the distal portion of a running shoe sole bonded to the bottom of the running-specific prosthesis. We depict foot placement of the biological legs (LL, RL, and UL) by the outline of the entire sole of a running shoe (not drawn to scale). 93

Figure 4.3 (Left column) To measure the step width and M-L foot placement relative to the midline of the body, we extracted the position of the reflective markers (open circles) at the instances of initial contact for each step while running. (Middle column) We defined step width as the M-L distance between markers during successive instants of initial contact. In this example, the non-amputee sprinter made initial contact with the left foot first. The position of the left heel marker defines the neutral axis (dashed reference line) for the position of the right heel marker at initial contact. Placing the right foot to the right of the line is considered a positive step width, to the left of the line a negative step width, and placement on the line as zero. For the subsequent step, the sign convention changes. (Right column) We defined M-L foot placement relative to the midline of the body as the distance between the heel marker and the pelvis COM. Following a similar convention as McClay and Cavanagh (1994), the position of the pelvis COM defines the midline of the body (dashed reference line). Placing the left foot to the left of the line is considered..... 94

Figure 4.4 Step width and step width variability across running speeds up to maximum sprint speed. (a) As non-amputee sprinters approached maximum speed, they decreased their step width to near zero (b and c) while sprinters with trans-tibial amputations did not show a strong tendency to decrease their step width as they approached maximum speed. As speed increased up to maximum sprint speed, step width variability increased in all sprinters (d, e, and f), indicating progressive decreases in lateral balance. For each variable, we present the equation and adjusted r^2 values from a least-squares linear regression analysis..... 98

Figure 4.5 M-L foot placement and M-L foot placement variability across running speeds up to maximum sprint speed. (a and d) Across running speeds, non-amputees exhibited symmetrical foot placement patterns between their RL and LL. As non-amputee sprinters approached maximum speed, they placed their foot closer to the midline of the body; however, each leg exhibited greater M-L foot placement variability. (b and e) On the contrary, sprinters with unilateral trans-tibial amputations exhibited asymmetrical foot placement patterns between the AL and UL. They tended to place the AL closer to the midline as compared to the UL, but the

AL exhibited greater M-L foot placement variability. (c and f) The sprinter with bilateral trans-tibial amputation also exhibited asymmetrical foot placement patterns. Although he tended to place the LAL closer to the midline of the body as compared to the RAL, M-L foot placement was similar across speed. On the contrary, both the LAL and RAL exhibited a large increase in M-L foot placement variability at faster running speeds. For each variable, we present the equation and adjusted r^2 values from a least-squares linear regression analysis. 102

Figure 5.1 (A) Reflective marker and EMG sensor placement for measurements of biological arm swing, leg swing, and deltoid muscle activity. From segment lengths and masses, we calculated the position of the biological arm's center of mass relative to the shoulder and computed the angle with respect to the horizontal axis in the sagittal plane. (B) During passive mechanical arm swinging, subjects wore our custom-built device, consisting of a set of football shoulder pads linked to aluminum bars that hung vertically from their pivot point. Reflective markers were placed at the locations of the ball bearing hinge, middle of the bar, and at the end of the lead mass. Damping of the passive mechanical arms was very small (~ 1.05 Ns/m) and equivalent to only 3.0% of the critical damping (~ 35.59 Ns/m) that would cause them not to swing (described in Appendix C). 112

Figure 5.2 (A) Biological arm swing and raw EMG activity for a subject walking at 70%, 100%, and 130% of their preferred step frequency (data for right arm only). Across step frequencies, the average EMG amplitude for the (B) posterior and (C) anterior deltoid resembled V-shaped and L-shaped trends, respectively ($n = 10$). Subjects significantly increased posterior deltoid activity when walking at slower and faster step frequencies (when compared to 100% preferred step frequency, * denotes $p < 0.05$). In contrast, subjects only increased anterior deltoid activity at slower step frequencies (when compared to 100%, * denotes $p < 0.05$). The anterior deltoid remained nearly silent at preferred and faster step frequencies. Values are means $\pm$ SEM. 121

Figure 5.3 (A) Amplitude of biological and passive mechanical arm swing for a representative subject walking at 70%, 100%, and 130% of preferred step frequency. (B) Average swing amplitude of biological (filled) and passive mechanical arms (open) while walking across normalized step frequencies (mean $\pm$ SEM; $n = 10$). At faster step frequencies (110-130%), the amplitude of biological arm swing decreased by 30% while passive mechanical arm swing stayed the same. At slower step frequencies (70%-90%), both biological and passive mechanical arm swinging amplitude increased steadily. The amplitude of passive mechanical arm swinging was clearly much less than that of biological arm swinging. Further, while biological right and left arm swinging was always found to be out-of-phase, passive right and left mechanical arm swinging exhibited a range of in-phase to out-of-phase oscillations. Note, when compared to 100% preferred step frequency, the black * denotes $p < 0.05$ for biological arm swing and the gray * denotes $p < 0.05$ for passive mechanical arm swing. 122

Figure 5.4 (A) Resonance diagram for a horizontally driven pendulum subjected to different frequencies. (B) Resonance diagram comparing the biological arms natural frequency (0.81 Hz) to the preferred step frequency adopted during walking (1.74 Hz). (C) Biological arm swing amplitude measured experimentally (closed circles) and predicted (open circles) for a horizontally driven anthropomorphic pendulum subjected to different step frequencies during walking (described in Appendix D). Across step frequencies, the predicted arm swing amplitude underestimates the swinging amplitudes of the biological arms. (D) Passive mechanical arm

swing amplitude measured experimentally (closed squares) and predicted (open squares) for a horizontally driven anthropomorphic pendulum subjected to different step frequencies during walking. Across step frequencies, the arm swing amplitudes predicted for a horizontally driven pendulum matches well with the swinging amplitudes of the passive mechanical arms. At the slowest step frequency (1.22 Hz = 70%), passive mechanical arm swinging reached the largest amplitude when approaching resonance. Note, the linearized approximation for a horizontally driven pendulum is not valid for angles much greater than 20 degrees, thus we do not include the predicted amplitude for the slowest step frequency..... 124

1. INTRODUCTION

During locomotion, the human body can be treated as a biological machine that utilizes chemical energy to generate muscular force and/or perform mechanical work. As Rodolfo Margaria (Margaria, 1976) stated, the “study of locomotion first requires the determination of the energy cost of this exercise and secondly a detailed analysis of the mechanical work performed”. Thus, in this literature review, I first compare and contrast the energetic costs of walking and running across a range of speeds. Next, I examine the fundamental features of human locomotion by describing the movements and mechanics of the whole body center of mass (COM). While focusing on the whole body COM dynamics, I will also discuss some simple but distinct biomechanical models that have been useful in revealing the principles of human locomotion. Then, I address an historical question that has been prevalent in the literature, i.e. “What is the biomechanical basis for the energetic cost of human locomotion?” I attempt to answer this question by providing a brief, chronological account of the major approaches to explaining the energetic cost of human walking and running.

The Energetic Cost of Human Walking and Running

When humans walk or run, the muscles require metabolic energy to generate force and perform mechanical work to move the body from one point to another. The rate of oxygen consumption ($\dot{V}O_2$) provides a good indication of the rate of metabolic energy consumption (Brooks et al., 2004). The $\dot{V}O_2$ is highly dependent on speed and the fundamental relation between the $\dot{V}O_2$ and the speed of walking and running exhibits two patterns (Figure 1.1A). As demonstrated by Ralston’s (Ralston, 1958) classic study, the $\dot{V}O_2$ (ml O_2 /kg/min) for walking increases curvilinearly as speed increases, which has been replicated by many others (Bobbert, 1960; Cotes and Meade, 1960; Givoni and Goldman, 1971; Martin et al., 1992; Menier and

Pugh, 1968; van der Walt and Wyndham, 1973; Wyndham et al., 1971). In contrast, the $\dot{V}O_2$ for running increases linearly as speed increases (Bøje, 1944; Hogberg, 1952; Margaria et al., 1963; Menier and Pugh, 1968; van der Walt and Wyndham, 1973; Wyndham et al., 1971).

The relation between the $\dot{V}O_2$ for walking and running versus speed demonstrates some important distinctions (Figure 1.1A). First, walking above a speed of 2.0 m/s is energetically expensive, exceeding the $\dot{V}O_2$ required for running at a similar speed. Second, the slope of the walking curve (for speeds > 2.0 m/s) is steeper than the slope of the running line indicating a greater demand for oxidative metabolism. Third, at speeds slower than 2.0 m/s, walking is energetically cheaper than running. As speed increases, humans transition between walking and running at ~ 2.0 m/s and thus minimize metabolic energy demand (Mercier et al., 1994).

One of the most interesting problems highlighted in the early literature was to understand at which speed walking and running is most economical. For example, is there a particular speed of walking or running that minimizes the energy cost of moving a unit mass a unit distance, defined as the cost of transport (expressed in ml O_2 /kg/meter)? Indeed, the relation between the cost of transport and speed depicts a U-shaped curve for walking with a minimum at 1.1 m/s. (Figure 1.1B).

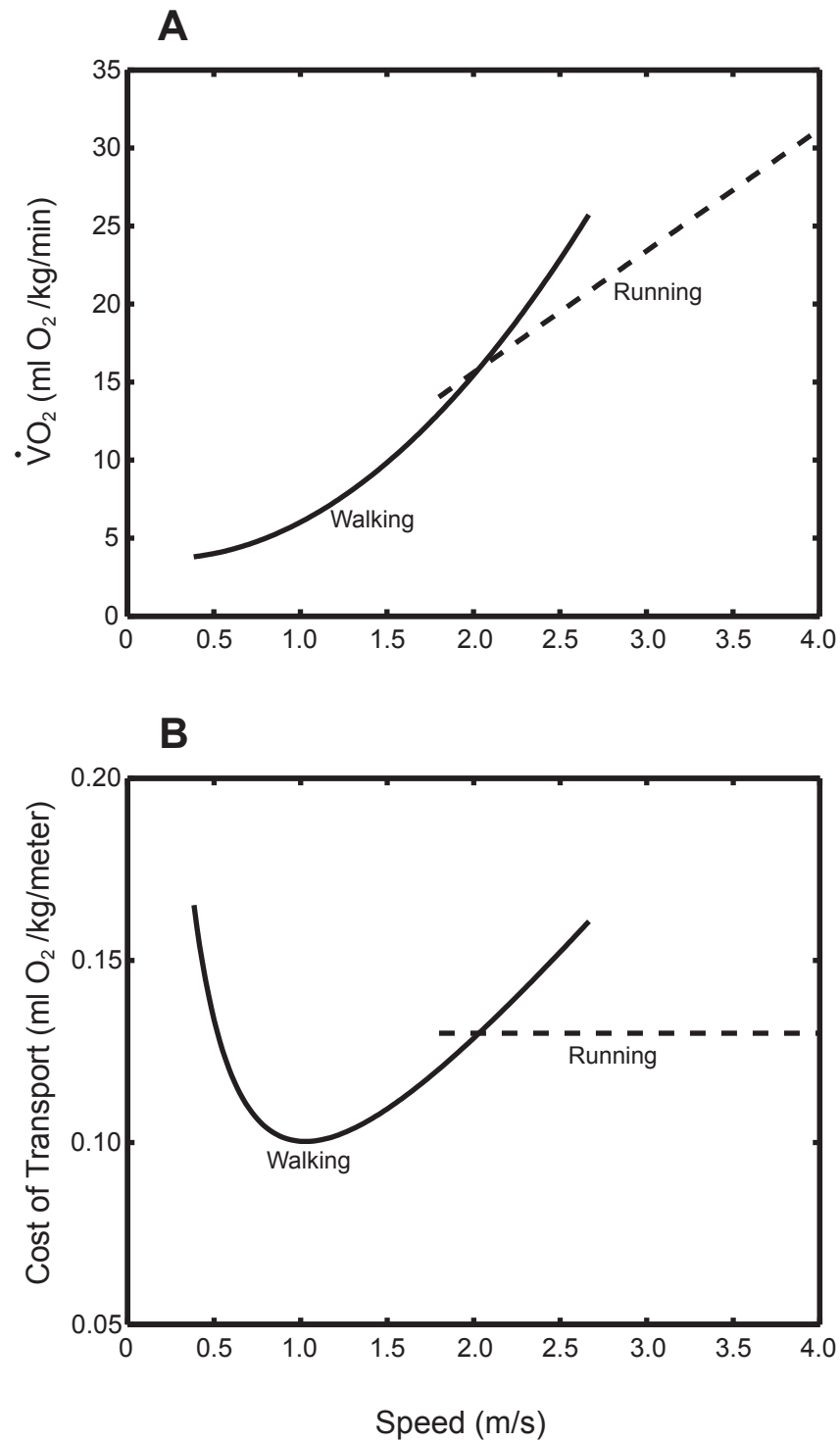


Figure 1.1 (A) $\dot{V}O_2$ (ml O_2 /kg/min) and (B) Cost of Transport (ml O_2 /kg/meter) as a function of speed (m/s). The curves are developed from the classic data published by Cotes and Meade (1960) and Margaria et al. (1963), respectively.

When humans are asked to walk at their freely chosen speed, they adopt a speed ($\sim 1.0\text{--}1.3$ m/s) that nearly minimizes the cost of transport (Ralston, 1958). As humans deviate from their preferred walking speed, the cost of transport increases. Although the optimal walking speed differs for each individual, a fundamental principle is that humans prefer a walking speed that minimizes the cost of transport. In contrast, the cost of transport for running is independent of speed since the $\dot{V}O_2$ increases linearly with speed (Margaria, 1976; Margaria et al., 1963).

Walking

In this section, I focus first on center of mass (COM) movements and mechanical energy fluctuations. Second, I describe the biomechanical basis for the net energetic of walking by summarizing the findings from both modeling and human walking experiments.

Mechanics

Simple Inverted Pendulum Model

Human walking can be simplified to the mechanical energy fluctuations of the whole body COM (Farley and Ferris, 1998). The mechanical energy fluctuations of the COM are important because they directly relate to metabolic energy cost. Human walking can be described as alternating phases of single support and double support. Single support occurs when only one leg is in contact with the ground. During this time, the relatively stiff leg supports the COM as it arcs up and over. Double support occurs briefly when both legs are in contact with the ground. During this time, the support of the COM transitions from the trailing leg to the leading leg, which has been termed the step-to-step transition (Donelan et al., 2001).

Historically, biomechanists have focused on the mechanical work performed on the COM with an emphasis on quantifying the mechanical energy exchange of kinetic (KE) and

gravitational potential energy (GPE; Figure 1.2). At the beginning of single support, the COM starts at its lowest position, reaches a maximum height at the mid-support, and then drops again to its lowest position preparing for the transition to the leading leg. Using a single force platform, Cavagna and Kaneko (Cavagna and Kaneko, 1977) noted that changes in KE and GPE of the COM are out of phase, resembling the mechanical behavior of an inverted pendulum. The exchange of KE into GPE results in small fluctuations in the total mechanical energy ($ME_{\text{total}} = KE + GPE$), which reduces the total external mechanical work performed by the lower leg muscles (Cavagna and Margaria, 1966; Cavagna et al., 1963; Elftman, 1966; Ralston and Lukin, 1969).

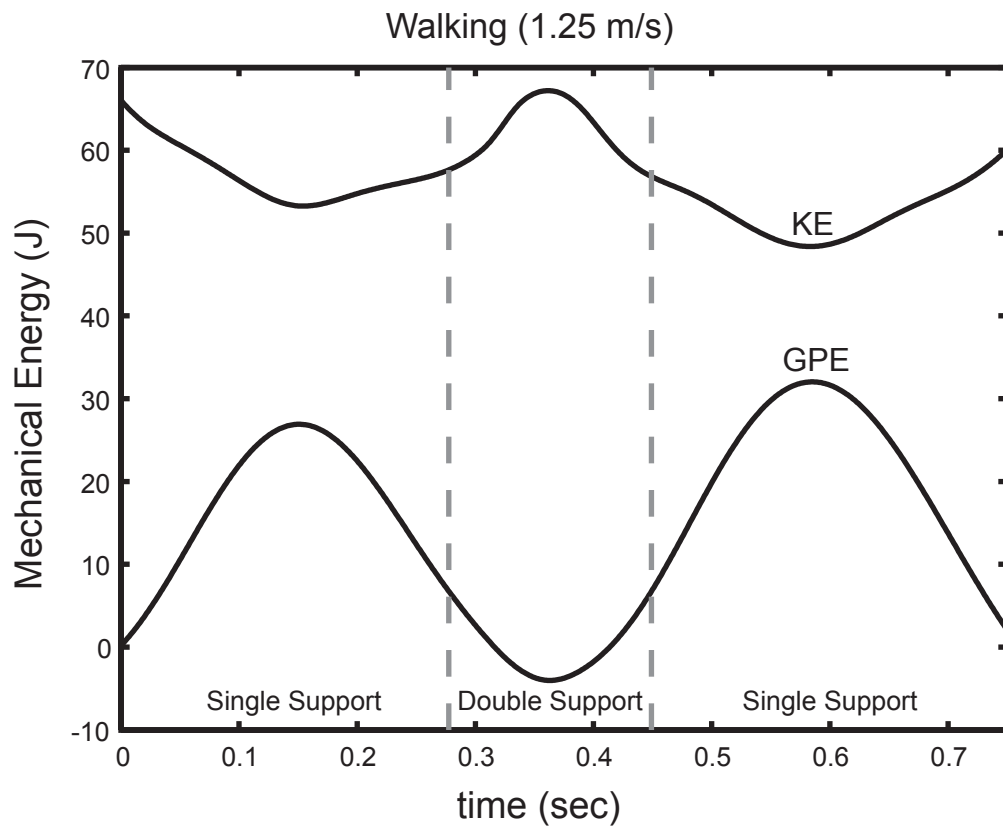


Figure 1.2 In walking, the kinetic energy (KE) fluctuations are out of phase with the gravitational potential energy (GPE) fluctuations. This out of phase behavior allows for inverted pendulum like exchange of mechanical energy.

The GPE of the COM is greatest at the middle of single support since the COM height is at a maximum and the KE is at a minimum (Figure 1.2). Some experimental evidence suggests that the exchange of GPE and KE reduces the energetic cost of walking. At the optimal walking speed, the percent recovery of mechanical energy calculated over an entire step reaches a maximum (~60-70%). At speeds other than optimal, the percent recovery decreases and more mechanical work is required to lift and accelerate the COM (Cavagna et al., 1977; Cavagna et al., 1976). Interestingly, the total external mechanical work per unit distance (cal/kg/km) is at a minimum near the optimal walking speed (Cavagna et al., 1976).

Although the inverted pendulum approach has been useful for understanding the mechanical energy fluctuations of the COM during walking, a major limitation is that this approach underestimates the total external mechanical work. Cavagna's (1975) original measurements of the external mechanical work performed during a step used a single force platform that combined the individual leg ground reaction forces. As illustrated in Figure 1.3, the inverted pendulum approach mistakenly cancels out the simultaneous positive and negative mechanical work performed during the step-to-step transition (Donelan et al., 2002b).

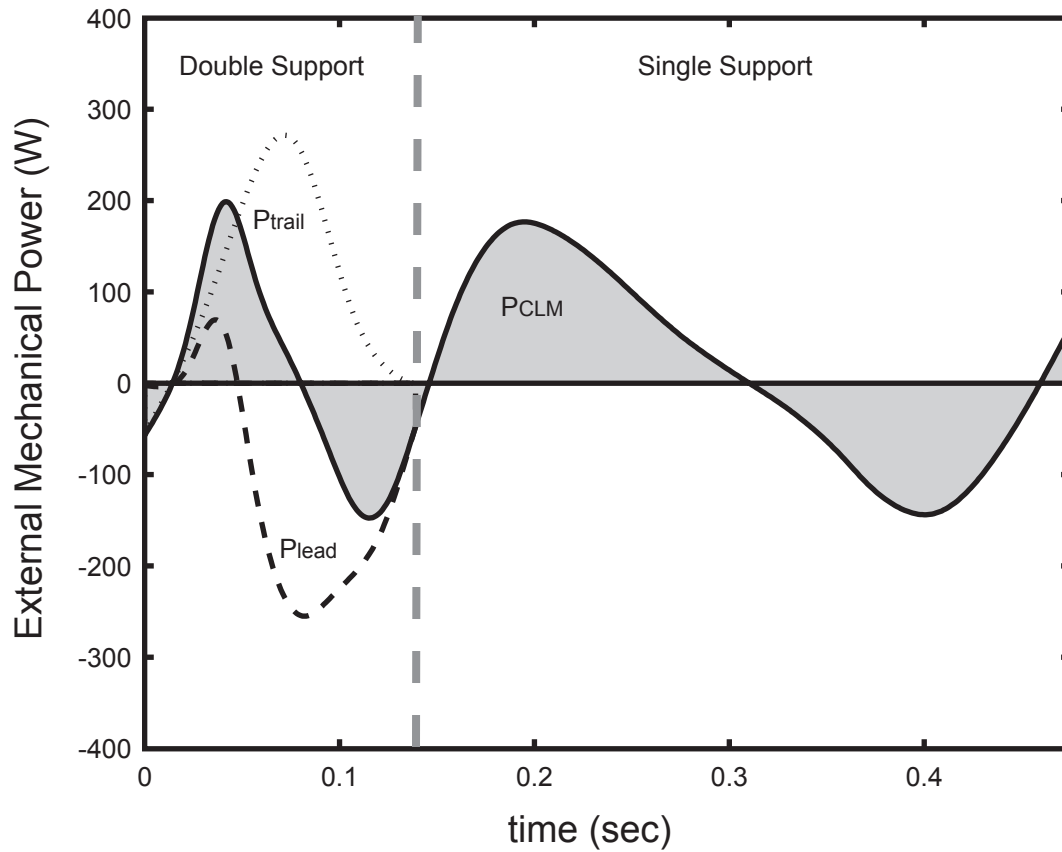


Figure 1.3 Mechanical power curve demonstrating the rate of positive mechanical work performed by the trailing leg (P_{trail}) and the rate of negative mechanical work performed by the leading leg (P_{lead}) as quantified by the individual limbs method (ILM; Donelan et al. 2002). The inverted pendulum approach (Cavagna et al., 1977), also known as the combined limbs method (CLM), mistakenly cancels the simultaneous rate of positive and negative work performed during double support. The gray shaded areas underneath the solid line represent the CLM and the area underneath the dashed lines represents the ILM. The data are calculated from the ground reaction forces of two separate force platforms and represent a single step while a subject walked over-ground at ~ 1.25 m/s.

Dynamic Walking Model

Initially, McGeer (McGeer, 1990b) developed a passive dynamic walking model (DW) for the purpose of building a simple two-dimensional walking machine. Simulations of the DW model produced leg motions comparable to the walking mechanics observed in humans. Kuo (Kuo, 2001) extended the DW model with the intention of explaining the energetic cost of walking. His model predictions along with human experiments indicate that the energetic cost of walking is the sum of 1) performing mechanical work during the step-to-step (S-S) transition and 2) the muscular force required for leg swing (Kuo, 2007).

In the DW model, the COM motion follows an arc with a velocity directed forward and upward at the beginning of single support and then the velocity is directed forward and downward at the end of single support (Figure 1.4A). During the S-S transition, the COM velocity must be redirected from a downward to an upward direction, which requires positive and negative mechanical work by the trailing and leading legs, respectively (Donelan et al., 2002a; Kuo, 2007). The individual limbs method accounts for the opposing mechanical limb work and provides a more complete view of how the total external mechanical work relates to the energetic cost of walking (Donelan et al., 2002a; Donelan et al., 2002b).

As a consequence of the S-S transition, mechanical energy is lost due to the initial collision of the leading leg with the ground (Figure 1.4A). The trailing leg must replace this lost energy by performing positive mechanical work to restore and redirect the COM velocity (Alexander, 1991; Donelan et al., 2002a). With the premise that metabolic rate is proportional to the mechanical work rate, the DW approach proposes that the mechanical work performed during the S-S transition explains a major portion of the energetic cost of walking (Donelan et al., 2002a). The model mathematically predicts that the mechanical work rate increases in

proportion to the fourth power of step length (S^4) with step frequency held constant. If mechanical work rate is proportional to metabolic rate, the model also predicts that metabolic rate will also increase with S^4 , with step frequency held constant. Thus, when varying step length alone, both mechanical work rate and metabolic rate will increase with S^4 . Human walking experiments support these predictions (Donelan et al., 2002a; Kuo, 2007; Kuo et al., 2005) demonstrating that both mechanical work rate and metabolic rate increases with S^4 (Figure 1.4B and C). The similar trends suggest that S-S transition work exacts a proportional energetic cost.

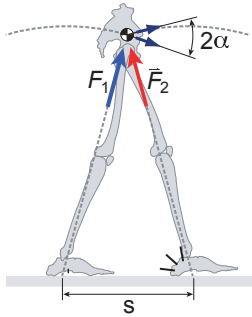
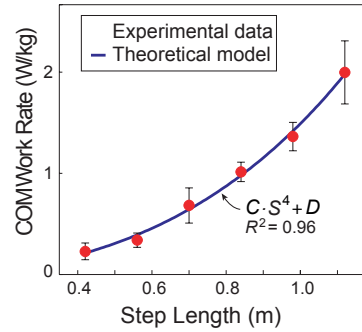
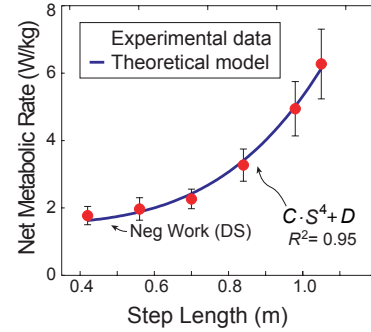
A. Step-to-Step Transitions**B. COM Work Rate vs. Step Length****C. Metabolic Rate vs. Length**

Figure 1.4 (A) The dynamic walking model predicts an energetic cost for the mechanical work performed during the step-to-step transition. The two leg forces F_1 and F_2 redirect the COM velocity from a downward and forward direction to an upward and forward direction, sweeping an angle 2α . (B) Experimental measurements of humans walking at constant step frequency but increasing step length support the models prediction. (C) Metabolic rate also increases with S^4 and in proportion to COM work rate. Figure and caption adapted from Kuo (2007).

The DW model also predicts that leg swing exacts an energetic cost (Figure 1.5). Humans increase walking speed by choosing an optimal combination of increasing step length and step frequency. Faster leg swing motion results in a proportional increase in the energy cost of walking. The model mathematically predicts that metabolic rate increases in proportion to the fourth power of step frequency (F^4) with leg swing amplitude held constant (Doke and Kuo, 2007). As seen in Figure 1.5C, isolated human leg swinging experiments support the model predictions (Doke and Kuo, 2007; Kuo, 2007). Kuo and colleagues have proposed that the leg swing is dominated by the cost of producing cyclic muscle force rather than performing mechanical work (Doke and Kuo, 2007; Kuo, 2007). Although the cost of cyclic muscle force hypothesis is difficult to test directly, models of human walking indicate that leg swing resembles the behavior of a swinging pendulum (Mochon and McMahon, 1980). Ideally, a short burst of muscle activity initiates leg swing and the energy supplied by gravity facilitates the forward motion of the leg. Then, a short burst of hip flexor muscle activity arrests leg swing preparing the leg for ground contact (McMahon, 1984). Further evidence from Doke and Kuo (2007) shows that the relation between metabolic rate and step frequency still held when energy cost of isolated leg swing was measured at a constant rate of mechanical work. Thus, muscle force, not mechanical work rate, provides a simple explanation for the energetic cost of leg swing.

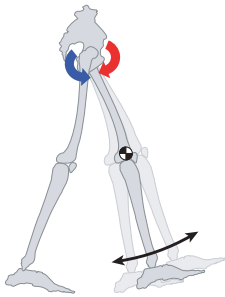
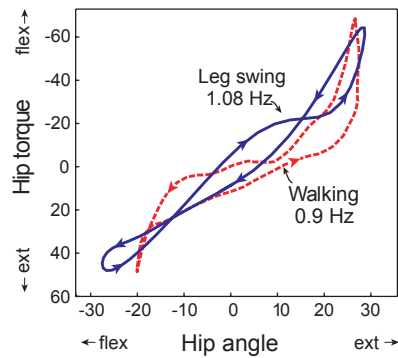
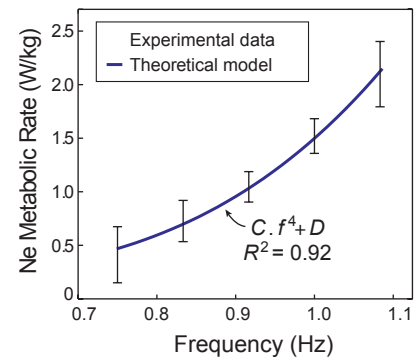
A. Forced Leg Motion**B. Hip Torque vs. Hip Angle****C. Metabolic Rate vs. Frequency**

Figure 1.5 (A) The dynamic walking approach predicts an energetic cost for the forced motion of the leg. (B) The energetic cost may be estimated from an isolated leg swinging experiment, in which the leg is moved about the hip with roughly the same torque and angle amplitudes as walking. (C) Experimental measurements show that the cost of leg swinging increases sharply, and in approximate agreement with the F^4 prediction. Figure and caption adapted from Kuo (2007).

The DW approach suggests that the S-S transition and swinging the legs are the major determinants of the energetic cost of walking. The predictions from the DW model are consistent with the results of human experiments. Insights from other human experiments, however, reveal that other biomechanical tasks exact a significant portion of the energetic cost of walking.

The Energetic Cost of Walking: Task-by-Task

An historical approach to explaining the energetic cost of walking was to relate the rate of mechanical work performed by the muscles to the rate of metabolic energy consumption. It appears that this approach stemmed from isolated muscle experiments (Fenn, 1924; Hill, 1938), which provided evidence that muscles must utilize metabolic energy in order to generate muscular force and perform mechanical work. Following this line of reasoning, several individuals attempted to measure the mechanical work performed during locomotion (Elftman, 1966; Fenn, 1930a, b), but Cavagna and colleagues must be credited with providing the most comprehensive measurements at the time.

From these measurements, Cavagna characterized the mechanics of walking as that of an inverted pendulum (IP). As previously mentioned, the DW approach extends the IP model and partitions the energetic cost of walking primarily into two tasks: the S-S transition and leg swing. A recent and alternative approach to explaining the energetic cost of walking is offered by computerized forward dynamic simulations (FDS) of human walking. In this approach, musculo-tendon actuators representing the major muscles of the lower leg drive the walking simulation. A few experiments to date have utilized this approach (Neptune et al., 2004; Umberger, 2010) and some of their findings are in disagreement with the predictions of the DW model (discussed in the following sections).

Kram and colleagues 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. Instead of separating the energetic cost of walking into distinct phases of single and double support as adopted by the DW and FDS approach, this task-by-task approach focuses on the biomechanical tasks that comprise human walking. These biomechanical tasks are defined as 1) body weight support, 2) propulsion, 3) leg swing, and 4) lateral balance. By externally reducing the energetic demand for a biomechanical task during walking, it can be inferred that the particular task accounts for a certain fraction of the energetic cost of walking. These human walking experiments, to a certain degree, have supported the predictions of the DW model, but they also reveal other important determinants for the energetic cost of walking. Thus, the goal of this section is to compare and contrast the tasks identified from these major approaches as well as quantify the energetic cost for each biomechanical task.

Body Weight Support

Muscular force is required to support the weight of the body during single and double support. Theoretically, the IP and DW model can only relate the COM mechanical work rate to the metabolic rate of walking. Thus, a major limitation of both models is that predictions cannot be made about the cost of generating muscular force to support body weight. In addition to body weight support, muscles perform external mechanical work to lift the COM during single support (Cavagna et al., 1976). However, the IP and DW models both predict that no mechanical work is required during single support because of the underlying assumption that there is a perfect exchange of KE and GPE. Thus, these models have assumed that the energetic cost of lifting the COM during single support is negligible. In contrast, the FDS approach suggests that muscles

perform the greatest amount of mechanical work in order to lift the COM during single support, indicating a major energy cost (Neptune et al., 2004). Although the predictions of this model have been controversial (Kuo and Donelan, 2009; Neptune et al., 2009), a recent FDS model estimating the rates of metabolic energy utilization in individual muscles during the full gait cycle supports this idea (Umberger, 2010). In this model, single support accounted for 34% of the energy cost of walking. However, caution should be taken when interpreting this value as it represents both the energy cost of generating muscle force to support body weight and lifting the COM during single support.

Whether the mechanical work required to lift the COM is energetically significant needs further evaluation. Nonetheless, there is no doubt that muscular force (with or without performing mechanical work) is required to maintain a stiff limb in order to support body weight. Several human experiments provide evidence that metabolic energy is required for body weight support. Farley and McMahon (1992) estimated the energetic cost of supporting body weight by utilizing a suspension system whereby a constant, upward force was applied to the whole body COM. By systematically reducing body weight from 1G to .25G (G is gravity), it was estimated that the energetic cost of body weight support was ~44% (Farley and McMahon, 1992). A later experiment using an improved suspension system (Grabowski et al., 2005) estimated that body weight support accounted for ~28%. Given that Grabowski et al. (2005) utilized a low-friction rolling trolley and included a sample of 10 subjects (vs. fixed pulley design and $n = 4$ in Farley and McMahon, 1992), it seems reasonable to infer that the energetic cost of supporting body weight accounts for ~28% of the net energetic cost of human walking.

Propulsion

In walking, metabolic energy is also used to propel the whole body COM in the forward direction. Propulsion, as a biomechanical task, is defined as the muscular effort required to accelerate the COM during single support as well as redirect the COM velocity during double support (Gottschall and Kram, 2003; Grabowski et al., 2005). The biomechanical task of propulsion is equated to the S-S transition where the redirection of the COM velocity occurs. According to the DW approach, S-S transitions account for ~60-70% of the net energetic cost of walking (Kuo et al., 2005). But at a similar speed, the FDS approach estimates that S-S transitions represent only 37% of the net energetic of walking (Umberger, 2010). Other experiments have partitioned out the energetic cost of propulsion by applying an optimal horizontal aiding force (%10 BW) about the waist. The aiding force reduces the need for the muscles to provide forward propulsion. This method estimates that propulsion accounts for ~47% of the net energetic cost of walking (Gottschall and Kram, 2003). Another means of quantifying the energetic cost of propulsion is by measuring the difference in energy cost while walking with the independent effects of added weight and added mass. When walking with added loads carried about the waist, the energetic cost increases due to the effects of 1) a greater muscular force generation to support body weight due to gravity and 2) a greater muscular force generation dedicated toward propulsion in order to accelerate and redirect a greater amount of mass. By comparing the differences in the energetic demand while walking with added body weight and added body mass, Grabowski et al. (2005) estimated that propulsion accounted for ~45% of the net energetic cost of walking which is in close agreement with the findings of Gottschall and Kram (2003). Thus, we can infer that propulsion accounts for ~45-47% of the net energetic cost of walking.

Leg Swing

During single support, the contralateral leg is in a state of flexion during early swing and in extension during late swing. Knee flexion during leg swing is important so that the foot clears the ground. As mentioned previously, the DW approach proposes a cost for leg swing that is due to generating muscular force. Estimates from isolated human leg swinging experiments indicates that leg swing explains ~33% of the net energetic cost of walking (Doke et al., 2005). Similar findings are observed from the FDS approach where leg swing accounts for ~29% of the net energetic cost of walking (Umberger, 2010). To empirically quantify the energetic cost of leg swing in human walking, Gottschall and Kram (Gottschall and Kram, 2005) reduced the energetic demand for leg swing initiation and propagation by independently applying a forward pulling force at the waist and a forward pulling force at each foot. Applying an optimal pulling force at the waist (10% BW) reduced the energetic cost of walking by ~42%. Applying an optimal combination of a pulling force at the waist (10% BW) and a pulling force at each foot (3% BW) further reduced the net energetic cost of walking by ~10%. From these data, the authors deduced that leg swing accounts for ~10% of the net energetic cost of human walking.

Lateral Balance

The DW model reveals that the sagittal plane dynamics are passively stable¹ while the frontal plane dynamics are inherently unstable (Kuo, 1999; McGeer, 1990b). However, frontal plane motion can be stabilized through feedback control via adjustments of media-lateral (M-L) foot placement from step-to-step. Insights from the DW model led to the hypothesis that humans require greater active control in the frontal plane compared to the sagittal plane due to the

¹ The term stable refers to the ability of DW model to resist small perturbations about a fixed point. If the periodic motion is stable, small perturbations about the fixed point will shrink and the model will yield a walking gait pattern. Otherwise, if the periodic motion is unstable, small perturbations about the fixed point will grow and the model will not yield a walking gait pattern (McGeer, 1990).

inherent instability of frontal plane motion (Bauby and Kuo, 2000). As such, the active control of lateral balance during walking is expected to exact an energetic cost. Donelan et al. (2004) tested this hypothesis with an external lateral stabilization device that applied external lateral forces about the waist while walking. These external lateral forces reduced the need for the active control of lateral balance and results in a decrease in the energetic cost of walking. Donelan et al.'s experiment and similar follow up experiments indicate that the active control of lateral balance exacts an energetic cost of ~3-9% (Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008).

Walking Summary

The different approaches described here partition the net energetic cost of human walking into distinct phases or different biomechanical tasks (Figure 1.6). The DW and FDS approach partition the net energetic cost of walking into distinct phases. The DW approach focuses on two distinct phases with a relative cost required for S-S transitions and leg swing. In contrast, the FDS approach predicts that the S-S transition, single support, and leg swing exact an energetic cost. The DW approach ignores any cost of single limb support. The task-by-task approach indicates other costs such as maintaining lateral balance and body weight support, which are not captured by the DW or FDS approach. However, all three approaches share a common finding in that the S-S transition comprises the largest component of the net energetic cost of human walking.

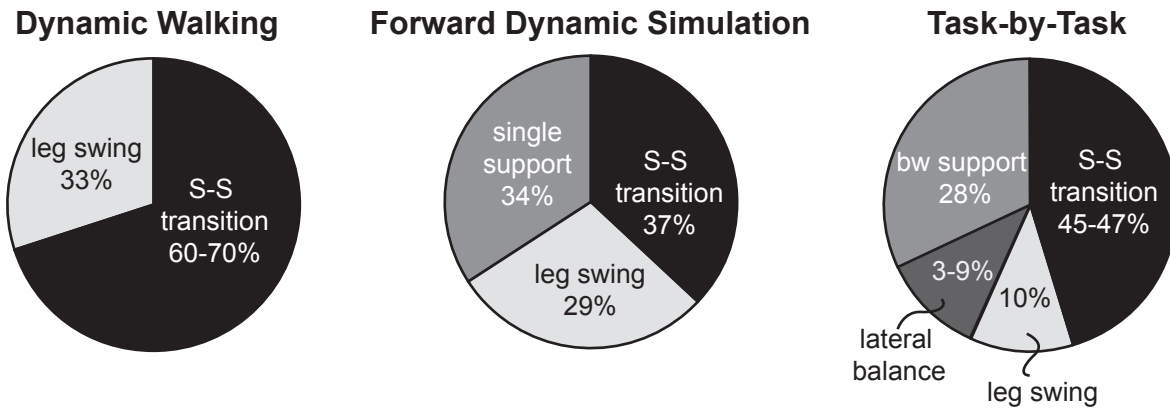


Figure 1.6 The energetic cost of walking is partitioned into distinct phases or biomechanical tasks as outlined by the three different approaches, i.e. the dynamic walking approach, the forward dynamic simulation approach, and the task-by-task approach.

Running

In this section, I first review the mechanics of human running by describing the COM movements and mechanical energy fluctuations. Second, I describe the biomechanical basis for the net energetic cost of human running by summarizing the findings from several key experiments.

Mechanics

Performing Mechanical Work

Initially, researchers attempted to apply the same mechanical work approach as for walking, i.e. exclusively considering the mechanical energy fluctuations of the COM (Cavagna et al., 1964; Fenn, 1930b). The mechanical energy fluctuations in running are fundamentally different from those observed in walking. Running is essentially a bouncing movement. At the beginning and end of the stance phase, the KE and GPE of the COM are at a maximum when the COM height is greatest (Figure 1.7). The KE and GPE reach their minimum values during mid-stance when the COM height is at its lowest position. In running, the inverted-pendulum like exchange (calculated as percent recovery) is negligible (0-4%) because the KE and GPE fluctuations of the COM are nearly in-phase (Cavagna et al., 1964; Cavagna et al., 1976; Farley and Ferris, 1998).

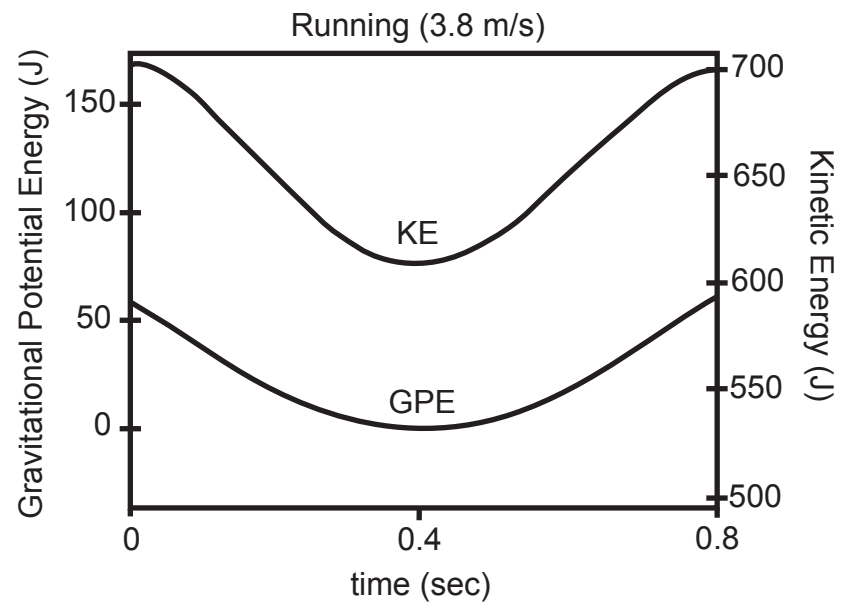


Figure 1.7 During running, the kinetic energy (KE) and gravitational potential energy (GPE) fluctuations of the COM are in-phase during the stance phase. Figure adapted from Farley and Ferris (1998).

Since there is negligible inverted-pendulum like exchange of mechanical energy during running, some authors initially proposed the idea that the muscles perform negative and positive mechanical work on the COM during the stance phase (Cavagna et al., 1976). During the 1st half of stance, negative work is required to decelerate (decreasing KE) and lower the COM (decreasing GPE) and during the 2nd half of stance, positive work is required to accelerate (increasing KE) and lift the COM (increasing GPE). During the negative work phase, active muscles were assumed to undergo lengthening and during the positive work phase, active muscles were assumed to undergo shortening to restore the total mechanical energy of the COM. Thus, the main explanation for the energetic cost of running was based on the presumption that the muscles themselves are performing external mechanical work on the COM.

Fenn (1930a, b), using kinematic data of the moving limbs and force plate measurements, was the first to relate the rate of total mechanical work (both internal and external²) and rate of metabolic energy during human running. However, these experiments were performed at maximum running speeds (~7.5 m/s) and are out of the scope of this review. Using similar methods, Cavagna and colleagues extended Fenn's findings by quantifying the changes in the total mechanical work across a range of running speeds (~3-6 m/s). It was found that the rate of total mechanical work increased linearly with speed (Cavagna et al., 1964) which resembled that of the linear relation observed between the $\dot{V}O_2$ and speed (Figure 1.1A).

Although the changes in the rate of external mechanical work vs. speed paralleled the changes in the $\dot{V}O_2$ vs. speed in running, Cavagna et al. (1964) realized that this approach overestimated the mechanical efficiency of skeletal muscle. Mechanical efficiency is defined as the ratio of the rate of mechanical work to the rate of metabolic energy consumption and

² Internal mechanical work refers to the movement of the limbs relative to the whole body center of mass. External mechanical work refers to the movement of the whole body center of mass relative to the ground.

describes the muscles ability to convert metabolic energy into useful mechanical work. Mechanical efficiency of skeletal muscle reaches a maximum value of 25% (Hill, 1922; Margaria, 1976). However, the mechanical efficiency measured in human running ranged from 40-50% (Cavagna et al., 1977; Cavagna and Kaneko, 1977; Cavagna et al., 1964), obviously much higher than the 25% efficiency limit. Thus, a large portion of the mechanical work performed on the COM could not be explained by the mechanical work performed by the muscles.

Spring Mass Model

The observation that the mechanical efficiency of running was much higher than 25% led Cavagna and colleagues to propose the idea that ‘elastic elements’ within the musculoskeletal system store and return mechanical energy. Therefore, Cavagna characterized human running as a bouncing ball (Cavagna et al., 1964) and this analogy was later extended to a spring-mass model whereby the leg acts as a linear spring (Blickhan, 1989; McGeer, 1990a; McMahon and Cheng, 1990; McMahon et al., 1987). In this model (Figure 1.8), the spring behavior of the leg is due to the elastic stretch and recoil of the muscles, tendons, and ligaments (Alexander, 1984; Ker et al., 1987).

The linear leg spring in the model compresses and stores mechanical energy during the 1st half of the stance phase and then recoils and releases the mechanical energy during the 2nd half of the stance phase (Figure 1.8). The energy released by the linear leg spring provides the mechanical energy for the acceleration of the body in the upward and forward direction. The elastic property of the linear leg spring is characterized by the leg stiffness (k_{leg}), which is defined as the ratio of the peak force of the spring to the peak displacement of the spring (Farley et al., 1993).

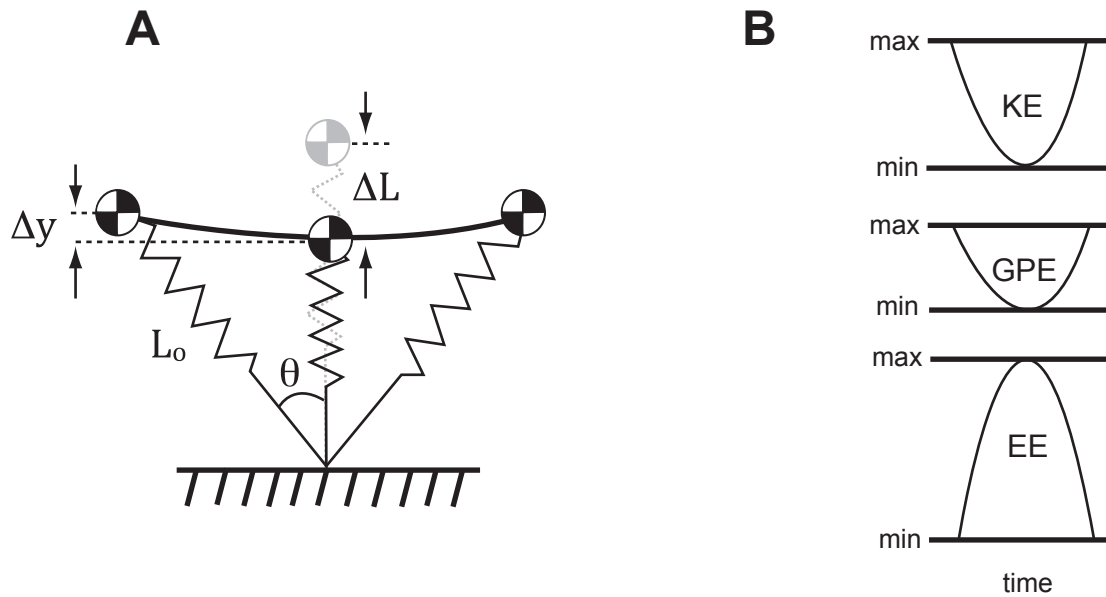


Figure 1.8 (A) Spring-mass model of human running. The model consists of a mass and a single leg spring (which connects the foot and COM). This figure depicts the model at the beginning of the stance phase (left most position), at the middle of the stance phase (leg spring is oriented vertically) and at the end of the stance phase (right most position). The leg spring has an initial length, L_0 , at the beginning of the stance phase, and ΔL represents its maximal compression. The dashed spring-mass model shows the length of the uncompressed leg spring. Thus, the difference between the length of the dashed leg spring and the maximally compressed leg spring represents the maximum compression of the leg spring, ΔL . The downward vertical displacement of the mass during the stance phase is represented by Δy and is substantially smaller than ΔL . Half of the angle swept by the leg spring during the ground contact time is denoted by θ . Figure and caption adapted from Farley and Gonzalez (1996). (B) This figure illustrates the fluctuations in the kinetic energy (KE), gravitational potential energy (GPE), and elastic energy (EE). During the 1st half of stance, the KE and GPE of the COM compress the linear leg spring and elastic energy (EE) is stored within the spring. During the 2nd half of stance, the linear leg spring recoils and releases the EE, which is converted back into KE and GPE.

Leg stiffness is an important determinant of the overall mechanics of human running. Based on the assumptions of the spring-mass model, several experiments reveal that leg stiffness is relatively constant over a range (3-6 m/s) of forward running speeds (Farley et al., 1993; He et al., 1991). Humans run faster by increasing the angle swept by the leg spring (θ) and reducing the vertical excursion of the COM (Δy) during the stance phase. These adjustments allow humans to run faster by bouncing off the ground more quickly. Although leg stiffness remains constant across running speeds, humans can drastically alter their leg stiffness to achieve different stride frequencies at a single running speed (Farley and Gonzalez, 1996) or to accommodate surfaces of different stiffness values (Ferris et al., 1999; Ferris et al., 1998; Kerdok et al., 2002). Overall, humans adjust leg stiffness to maintain similar COM motion and mechanical energy fluctuations (Farley and Ferris, 1998).

Despite the simplicity of the spring-mass model of human running, this analogy presents a paradox for explaining the energetic cost of running. If human running can be modeled as a simple spring-mass system, why does running incur any metabolic energy at all? Theoretically, the external mechanical work on the COM can be performed by the spring itself. Furthermore, this model assumes that the linear spring is perfectly elastic meaning that the same amount of mechanical energy is stored and returned during the stance phase of running. Therefore, the simple spring-mass model cannot explain the energy cost of running but one must realize that the model's main intention was to describe the mechanics of human running and has done so extremely well.

In summary, Cavagna's et al. (1964) mechanical work approach did not provide a satisfactory explanation for the energetic cost of running. However, his approach did point to the

elastic mechanisms involved in human running. These elastic mechanisms greatly reduce the energetic cost of running in humans and other animals (Alexander, 1984; Cavagna et al., 1977; Cavagna et al., 1964). Despite these initial efforts, it was not until 25 years later that a better explanation for the energetic cost of running emerged.

Generating Muscular Force

Since the transformation of metabolic energy into mechanical work did not explain the energetic cost of running (Heglund et al., 1982a; Heglund et al., 1982b; Taylor et al., 1982), C.R. Taylor and colleagues (Taylor et al., 1980) proposed the ‘cost of generating force’ approach. This cost of generating force approach was based on the idea that muscles transform metabolic energy into force and not necessarily mechanical work. By measuring rates of oxygen consumption in a variety of small and large animals (.30kg-120kg) and humans (90kg) while running with various loads (7-27% of body mass), Taylor et al. (1980) observed that rate of oxygen consumption increased in direct proportion to the added mass. Thus, they proposed that the energetic cost of running can be explained by the cost of generating force over time ($\int F dt$) which increases with speed and decreases with body mass. Taylor (Taylor, 1985) further explored this hypothesis by providing evidence that the rate of metabolic energy consumption at physiologically equivalent speeds (trot-gallop transition) did not change in parallel with the mechanical work performed; however, the rate of metabolic energy consumption at physiologically equivalent speeds did change in parallel with the time course of generating force per stride.

Kram and Taylor (Kram and Taylor, 1990) then provided a more refined and comprehensive explanation for the energetic cost of running by demonstrating the rate of metabolic energy was proportional to body weight and inversely proportional to the time of

contact when the foot applies force to the ground (Eq. 1) for a variety of animals ranging in size (30g-140kg) and speed.

$$\dot{E}_{metabolic} = c \cdot \frac{W_b}{t_c} \quad (1)$$

Here, c is defined as the cost coefficient and was found to be nearly constant across animal size and running speed which demonstrates that the energetic cost of running can be explained by the cost of supporting body weight and the time course of generating force during ground contact. Experiments on running humans further supported Taylor's original hypothesis whereby changes in metabolic rate versus speed paralleled the changes in the rate of force generation ($1/t_c$) versus speed (Roberts et al., 1998). Furthermore, *in-vivo* measurements from the lateral gastrocnemius of running turkeys reveal that the major role of the active muscle fibers is to generate force and support body weight and that the majority of the mechanical work is performed by the elastic stretch and elastic recoil of the tendon (Roberts et al., 1997). The elastic stretch and elastic recoil of the Achilles tendon plays a similar role in human running (Alexander, 1984; Alexander and Bennetclark, 1977; Ker et al., 1987) and the ability of the Achilles tendon to store and release elastic energy relies upon the active muscle fibers of the soleus and gastrocnemius. Other anatomical structures (e.g. arch of the foot) also store and return elastic energy (Ker et al., 1987).

Energetic Cost of Running: Task-by-Task

To provide a more detailed explanation for the energetic cost of running, Kram and colleagues have adopted a task-by-task approach (as described in the walking section). The energetic cost of running can be partitioned into the biomechanical tasks of: 1) body weight support, 2) propulsion, 3) leg swing, and 4) lateral balance. Here, I summarize the energetic cost

for each biomechanical task as estimated from the cost of generating force approach and the task-by-task approach.

Body Weight Support

During the stance phase of running, muscles are recruited to generate force, which demands metabolic energy. Although the storage and return of mechanical energy is primarily performed by the elastic elements of the muscle-tendon unit of the lower limb, the muscle fibers still consume metabolic energy to support body weight support and operate the spring-like behavior of the leg (Taylor et al., 1980). The cost of body weight support can be estimated from Eq. 1. Roberts et al. (1998) estimated that 70% of the increase in metabolic rate between running speeds of 2-4 m/s could be explained by the changes in the rate of force generation ($1/t_c$). Other human running experiments that directly manipulate body weight support are consistent with these findings.

For example, Farley and McMahon (1992) reduced body weight from 1G to .25G (as described in the walking section) in running humans and found that the rate of metabolic energy consumption decreased in direct proportion to the amount of body weight support. Those data suggest that body weight support can explain ~100% of the net energetic cost of human running. However, in later experiments, Teunissen et al. (Teunissen et al., 2007) and Wardripp and Kram (Warddrip and Kram, 2007) found body weight support to comprise ~74% and ~65% of the net energetic cost of running, respectively. Both follow up studies utilized a low-friction rolling pulley as opposed to the fixed pulley design used by Farley and McMahon (1992). It appears that the fixed pulley design of Farley and McMahon (1992) not only reduced body weight support but also assisted with forward propulsion resulting in an overestimation for the energetic cost of body weight support.

Propulsion

Although the horizontal ground reaction forces are much smaller in magnitude compared to the vertical ground reaction forces, indirect evidence suggest that generating horizontal forces is an important determinant of the energetic cost of running. Performing external work against an impeding load (applied about the waist) during running exacts an energetic cost that increases linearly with the magnitude of the load (Lloyd and Zacks, 1972; Zacks, 1973). Chang and Kram (Chang and Kram, 1999) later extended those findings by estimating the energetic cost of generating horizontal ground reaction forces while humans ran (3.3 m/s) with a nearly constant horizontal impeding or aiding force applied at the waist. This study demonstrated that generating horizontal propulsive forces is more energetically costly than generating horizontal braking forces. Furthermore, an optimal horizontal aiding force (15% BW) reduced the net metabolic rate by 37% (assuming a standing metabolic rate of 5.5 ml O₂/kg/min). Using essentially the same device, Moed and Kram (Moed and Kram, 2004) confirmed that the net energetic cost forward propulsion is ~40% during running.

Leg Swing

Although leg swing in running serves no direct role in the storage and return of mechanical energy, muscles must be recruited to initially flex the knee and hip, and then extend the knee and hip during the end of swing. Obviously, such muscular actions would exact some energetic cost. To determine the energy cost of leg swing during running, Modica and Kram (Modica and Kram, 2005) provided an aiding force to initiate and propagate the leg forward during swing. These assistive forces reduce the need for the muscles to directly swing the leg. Modica and Kram (2005) inferred that leg swing comprised at most 20% of the net energetic cost of running (3 m/s). Interestingly, Marsh et al. (Marsh et al., 2006) had previously estimated the

cost of leg swing in running birds (guinea fowl) to demand 26% of the net energetic cost. However, Moed and Kram (2004) applied a combination of aiding horizontal forces at the waist and assistive forces at the feet and inferred that leg swing comprised only 10% of the net energetic cost of running. This 10% value was found to be constant across a range of running speeds (2-4 m/s). Wardriipp and Kram (2007) combined body weight support, aiding horizontal forces at the waist, and assistive leg swing forces. They deduced that leg swing comprises only 7% of the net energetic cost of running. Taken together, it appears that the study of Modica and Kram (2005) overestimated the cost of leg swing. It seems reasonable that their assistive leg swing forces also decreased the energetic demand for forward propulsion.

Lateral Balance

Until recently, it was unknown whether lateral balance in running incurred a significant energetic cost. While running, humans prefer a narrow step width so as to minimize energetic cost and optimize for lateral balance (Arellano and Kram, 2011). I suggested that lateral balance in human running would not exact a significant energetic cost since humans prefer a step width near zero that is energetically optimal. This idea was supported by my recent experiment, which provided external lateral stabilization (LS) in human running (Arellano and Kram, 2012). Similar to the walking experiments described previously (Donelan et al., 2004), external LS reduced the need for active control of lateral balance. We inferred that the reduction in energy cost with external LS reflects the energetic cost of maintaining lateral balance during normal running. Our findings indicate that lateral balance demands a net energetic cost of only 2% during running.

Running Summary

The cost of generating force and the task-by-task approach both illustrate that body weight support is the primary determinant of the energetic cost of running (Figure 1.9). In the task-by-task approach, propulsion represents the second largest determinant for the energetic cost of running while leg swing and lateral balance exact relatively small energetic costs. If we sum all the biomechanical tasks, the lower and upper bounds for the energetic cost of running are paradoxically 111% and 126%, respectively. These results are likely due to the overestimation of body weight support and propulsion because of their interactive nature. For example, the plantar flexor muscles have multi-functional roles in contributing to body weight support, forward propulsion, and initiating leg swing; thus, these interaction effects make it difficult to empirically separate these relative costs.

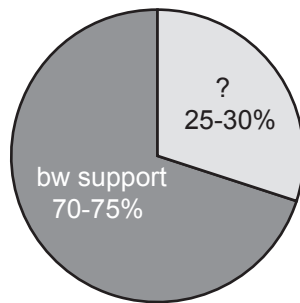
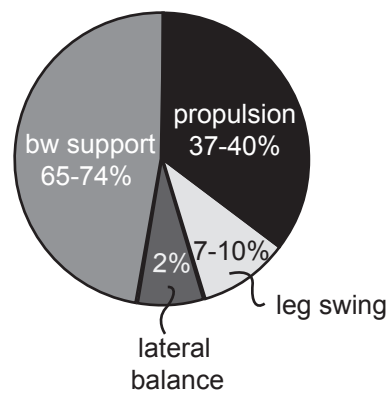
Cost of Generating Force**Task-by-Task**

Figure 1.9 The energetic cost of running is partitioned into the biomechanical tasks as outlined by the cost of generating force approach and the task-by-task approach.

Overall Conclusion

In this review, I have outlined the net energetic cost of human walking and running into the biomechanical tasks of body weight support, propulsion, leg swing, and lateral balance. While learning about the energetics and mechanics of human locomotion, I realized that the literature is dominated by the idea that humans prefer to walk or run in a particular manner with the sole purpose of minimizing energetic cost. Thus, the idea that humans walk or run in a manner that minimizes energetic cost is generally accepted. My previous research experience in studying balance in human locomotion (Arellano et al., 2009); however, lead me to believe that optimizing for balance must also play an important role in how humans prefer to walk and run. Since there have been an extensive number of studies investigating balance during human walking, I decided to focus my research on human *running*. Therefore, a main goal of my dissertation was to test the general hypothesis that humans simultaneously minimize energetic cost and optimize for lateral balance while running.

The first goal of my dissertation was to determine how humans maintain lateral balance during running. I chose to study step width and arm swing because several walking studies suggested that humans demonstrate active control of lateral balance by varying step width from step-to-step and by swinging their arms. In addition, previous studies on running reported conflicting results as to whether arm swing helps to minimize energetic cost and improve lateral balance. Because several studies produced conflicting results and did not quantify balance rigorously, my first study investigated the independent effects of step width and arm swing on the energetic cost and lateral balance during running. I found that eliminating arm swing increases net metabolic power demand and step width variability, indicating a decrease in lateral

balance. However, the associated increase in step width variability when arm swing was eliminated did not establish cause and effect between lateral balance and arm swing.

The second goal of my dissertation was to establish cause and effect relations between step width adjustments, arm swing, and lateral balance during human running with a more direct approach. In addition, my second study quantified the energetic cost of maintaining lateral balance when running with and without arm swing by applying external lateral stabilization. I concluded that humans use step width adjustments, not arm swing, as the primary mechanism for maintaining lateral balance during running.

The third goal of my dissertation was to extend my insights on lateral balance to a population of sprinters with and without trans-tibial amputations as well as to understand how running speed up to maximum sprint speed affects lateral balance. Overall, I found that 1) sprinters with trans-tibial amputations have greater challenges with maintaining lateral balance and 2) maintaining lateral balance is more challenging at faster running speeds.

Finally, the fourth study that comprises my dissertation focused on the underlying mechanism(s) that drive arm swing during human walking. This has been a controversial topic since the 1800's and the controversy re-surfaced in articles published in the *Journal of Experimental Biology* and *Proceedings of the Royal Society B: Biological Sciences* in 2009. The underlying claim of these papers is that arm swing is primarily passive. Motivated by this long standing controversy, I addressed the following question: Is arm swing during human walking actively driven by shoulder muscle forces, a passive pendulum-like response to body accelerations, or some combination? The main conclusion from my study is that arm swing during human walking comprises a hybrid system of active muscular actuation and passive pendulum dynamics.

Overall, my dissertation comprises four studies that focus on the balance control mechanisms in human locomotion (see each title below). Chapter 1 was published in the Journal of Biomechanics in 2011. Chapter 2 was published in the Journal of Applied Physiology in 2012. Chapters 3 and 4 will be submitted for publication pending final comments from my co-authors.

Arrangement of Dissertation

1. The effects of step width and arm swing on energetic cost and lateral balance during running
2. The energetic cost of maintaining lateral balance during human running
3. Lateral balance and foot placement across running speeds: A comparison of sprinters with and without trans-tibial amputations
4. Arm swing during human walking: Active and passive contributions to a hybrid-system

2. THE EFFECTS OF STEP WIDTH AND ARM SWING ON ENERGETIC COST AND LATERAL BALANCE DURING HUMAN RUNNING

A. Abstract

In walking, humans prefer a moderate step width that minimizes energetic cost and vary step width from step-to-step to maintain lateral balance. Arm swing also reduces energetic cost and improves lateral balance. In running, humans prefer a narrow step width that may present a challenge for maintaining lateral balance. However, arm swing in running may improve lateral balance and help reduce energetic cost. To understand the roles of step width and arm swing, we hypothesized that net metabolic power would be greater at step widths greater or less than preferred and when running without arm swing. We further hypothesized that step width variability (indicator of lateral balance) would be greater at step widths greater or less than preferred and when running without arm swing. Ten subjects ran (3 m/s) at four target step widths (0, 15, 20, and 25% leg length (LL)) with arm swing, at their preferred step width with arm swing, and at their preferred step width without arm swing. We measured metabolic power, step width, and step width variability. When subjects ran at target step widths less (0%LL) or greater (15, 20, 25%LL) than preferred, both net metabolic power demand (3, 9, 12, 15%) and step width variability (7, 33, 46, and 69%) increased. When running without arm swing, both net metabolic power demand (8%) and step width variability (9%) increased compared to running with arm swing. It appears that humans prefer to run with a narrow step width and swing their arms so as to minimize energetic cost and improve lateral balance.

B. Introduction

Minimizing energetic cost and maintaining lateral balance are important goals in human locomotion. Humans prefer to walk with a moderate step width (~ 12 cm) that minimizes energetic cost (Donelan et al., 2001). In contrast, humans run with a step width near zero (Cavanagh, 1987), which would seem to challenge lateral balance and incur a greater energetic cost. Humans also prefer to swing their arms while walking and walking without arm swing increases energetic cost (Collins et al., 2009; Ortega et al., 2008). In running, arm swing may assist with lateral balance and possibly reduce energetic cost. In this study, we investigated if humans not only minimize energetic cost but also optimize for lateral balance while running.

There are many examples of energetic optimization in human locomotion. The energetic cost of walking per unit distance (cost of transport) plotted as a function of speed exhibits a U-shaped curve with a minimum close to the preferred walking speed (Martin et al., 1992; Ralston, 1958; van der Walt and Wyndham, 1973; Workman and Armstrong, 1963; Zarrugh et al., 1974). Similarly, if walking or running speed is fixed and stride frequency is varied, energetic cost also exhibits a U-shaped relationship with a minimum near the preferred stride frequency (Cavanagh and Williams, 1982; Hogberg, 1952; Holt et al., 1991; Umberger and Martin, 2007). The walk-run transition occurs near the speed where running becomes more economical than walking (Mercier et al., 1994). Thus, the idea that humans prefer to walk or run in a manner that minimizes energetic cost is generally accepted.

Minimizing energetic cost, however, is not the only goal during human locomotion. Maintaining lateral balance is a critical prerequisite that requires active control via sensory feedback (Bauby and Kuo, 2000; Donelan et al., 2004). One way humans demonstrate active control of lateral balance in walking is by varying step width from step-to-step (Bauby and Kuo,

2000) but humans prefer an average step width that minimizes energetic cost. Walking with step widths narrower or wider than preferred is energetically more expensive (Donelan et al., 2001). In contrast, humans run with much narrower step widths (Cavanagh, 1987). Placing the foot along the midline of the body aligns the vertical ground reaction force close to the whole body center of mass (Figure 2.1; (Cavanagh, 1987; McClay and Cavanagh, 1994). Thus, the majority of the center of mass motion is directed in the forward and vertical directions and the side-to-side motion is relatively small (Cavanagh, 1987). Reducing the side-to-side motion of the center of mass may be an effective mechanism for balance control during running. As in walking, the preferred step width in running may minimize energetic cost and facilitate balance control.

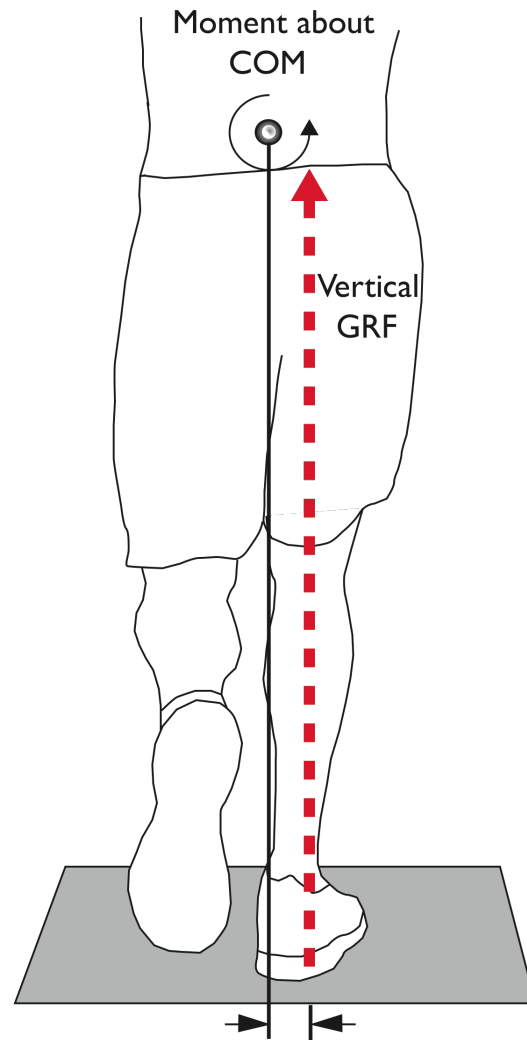


Figure 2.1 Rear-view of foot placement relative to the midline of the body during human running. The vertical GRF can reach values of 2-3 times body weight and is the largest component making up the resultant GRF during human running. Placing the foot along the midline of body aligns the vertical ground reaction force close to the whole body center of mass (COM), effectively reducing the moment generated about the COM along the A-P axis. Modeled after McClay and Cavanagh (1994).

Mann and colleagues inferred that the primary purpose of arm swing in running is to improve balance with no apparent role in reducing energetic cost (Mann and Herman, 1985; Mann, 1981). Experimental studies of running mechanics reveal that arm swing improves balance by 1) counteracting the angular momentum produced by the swinging legs about the vertical axis and 2) reducing the side-to-side motion of the center of mass (Hinrichs, 1987; Hinrichs et al., 1987). Hinrichs et al. (1987) suggested that these mechanical effects might reduce energetic cost and was later supported by Egbunu et al. (Egbunu et al., 1990) who reported a 4% increase in energetic cost of running without arm swing. However, a recent report by Pontzer et al. (Pontzer et al., 2009) concluded that running without arm swing did not affect energetic cost or lateral balance. Because these studies produced conflicting results and did not all quantify balance rigorously, we investigated the independent effects of step width and arm swing on the energetic cost and lateral balance of running. Thus, we designed part of our study to test if there is a link between arm swing and lateral balance. Following the lead of previous studies (Bauby and Kuo, 2000; Donelan et al., 2001; Ortega et al., 2008), we measured step width and its variability as indicators of lateral balance.

We addressed two questions: 1) Why do humans run with a step width near zero? and 2) Why do humans prefer to swing their arms during running? We reasoned that a narrow step width and arm swing reflect important control strategies that help minimize energetic cost and improve lateral balance. We hypothesized that the energetic cost of running would be greater 1) at step widths greater or less than preferred and 2) without arm swing. We further hypothesized that step width variability would be greater 3) at step widths greater or less than preferred and 4) without arm swing.

C. Methods

Ten subjects volunteered for this study as per the University of Colorado IRB (5 males and 5 females, age = 24.4 ± 4.2 years, mass = 65.4 ± 11.7 kg, leg length (LL) = 93.1 ± 5.6 cm; mean $\pm$ s.d.). Subjects wore their own shoes, were experienced with treadmill running, and were healthy and injury-free.

Subjects initially stood on a force measuring treadmill (Kram et al., 1998) while we measured rates of O₂ consumption ($\dot{V}O_2$) and CO₂ production ($\dot{V}CO_2$) for seven minutes using expired gas analysis (ParvoMedics TrueMax2400, Salt Lake City, Utah). We placed reflective markers on the left and right heel, dorsum of the 2nd toe, and lateral mid-foot of each shoe. We provided real-time visual feedback (Motion Analysis Corporation, Santa Rosa, CA) of foot placement during running (Figure 2.2). We created two virtual lines by positioning reflective markers in the front and back of the treadmill.

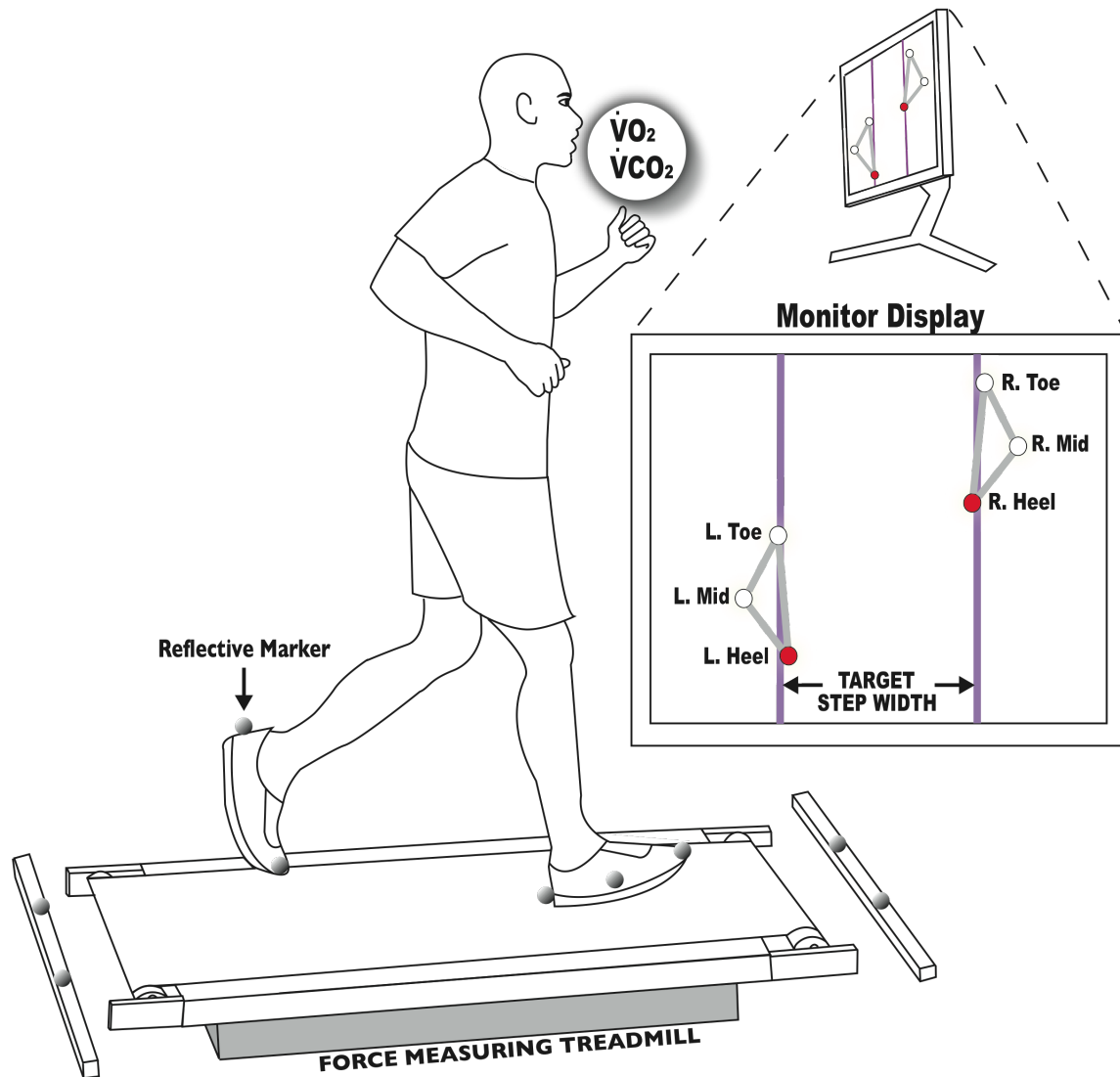


Figure 2.2 Real-time visual feedback of foot placement (monitor providing a top-down view) during the target step width conditions. We placed reflective markers on the left and right feet. Adjustable markers defined the left and right virtual lines. The distance between the virtual lines was set as a percentage of leg length (%LL). For a target step width of 0 %LL, we projected a single virtual line along the middle of the treadmill. Visual feedback was displayed on a computer monitor (30 x 47 cm) positioned in front of each subject (~ 0.5 m).

Subjects ran at 3 m/s on the force-treadmill for randomized conditions of target step widths (0%, 15%, 20%, and 25% LL) with arm swing, at preferred step width with arm swing (Arms), and at preferred step width without arm swing (No Arms). For the target step width conditions, we instructed subjects to position their heel markers on the respective virtual lines at initial contact. The 0% LL condition was accomplished by projecting a single virtual line corresponding to the middle of the treadmill belt. During the No Arms condition, subjects crossed their arms in front of their chest. Subjects ran for seven minutes while we measured $\dot{V}O_2$ and $\dot{V}CO_2$. During the last four minutes, we recorded the three-dimensional motions of the feet (100 Hz) and the ground reaction forces (1000 Hz).

Data Analysis

For each condition, we calculated the average $\dot{V}O_2$ and $\dot{V}CO_2$ during the last three minutes and computed the net metabolic power (Brockway, 1987) by subtracting the average value during standing from the average value during running. We filtered the position data of the left and right heel markers using a 9th order, zero-lag low-pass Butterworth filter with a cutoff frequency of 6 Hz. To determine the instant of initial-contact for each step, we utilized the vertical ground reaction force (GRF) data, which were filtered using a 4th order, zero-lag low-pass Butterworth filter with a cutoff frequency of 50 Hz. Initial contact was identified with a vertical GRF threshold of 10% body weight. To synchronize the kinematic and kinetic data, we down-sampled the filtered vertical GRF data to 100 Hz. As recommended (Owings and Grabiner, 2003), we calculated the average step width and step frequency during the last 401 consecutive steps for each trial. Step width was defined as the medio-lateral (M-L) distance between the right and left heel markers during successive instants of initial-contact. We defined step width variability, an indicator of lateral balance, as the standard deviation about the average

step width (Bauby and Kuo, 2000; Donelan et al., 2001; Ortega et al., 2008). We normalized step width and step width variability by dividing each variable by leg length and multiplying by 100.

Statistical Analysis

We used a repeated measures ANOVA with *a priori* comparisons between the control (Arms) and target step width conditions using Dunnett's multiple comparison method and published data table for a one-sided comparison against a control (Dunnett, 1955, 1964). If Mauchly's test of sphericity was violated at the 0.05 level, we adjusted the degrees of freedom (e.g. Huynh-Feldt) to calculate the critical *t*-statistic. To compare between Arms and No Arms, we used paired *t*-tests with $\alpha = 0.05$ (SPSS Inc., Chicago, IL).

D. Results

Step Width

Subjects consumed energy at a faster rate when running at step widths greater than the preferred step width (Arms) condition. On average, subjects ran with a preferred step width of 3.95% LL (~ 3.6 cm). Compared to the Arms condition, the net metabolic power demand was 9, 12, and 15% greater at target step widths of 15, 20, and 25% LL, respectively (all *p*'s < 0.01, Figure 2.3). At 0% LL, the net metabolic power demand was 3.3% greater ($0.05 > p > 0.01$) when compared to the preferred step width (Arms) condition. Subjects were unable to match the target step widths provided by the real-time visual feedback system (Table 2.1). The average step width at the 15, 20, and 25% LL conditions were consistently narrower than the target step width.

Step width variability was greater at target step widths greater than preferred (Figure 2.4). Compared to the control condition (Arms), step width variability was greater by 33, 46, and 69% at target step widths of 15, 20, and 25% LL, respectively (all *p*'s < 0.01, Figure 2.5). Step width

variability at the 0% LL condition was 7% greater than the Arms condition but the difference was not statistically significant. Subjects used similar step frequencies across the target step width conditions (Table 2.1).

Arm Swing

When running without arm swing, the net metabolic power demand increased by 8% compared to running with arm swing ($p < .0001$, Figure 2.3). Running without arm swing did not change the average step width ($p = .084$) but increased step width variability by 9% ($p = .023$, Figure 2.5). Lastly, step frequency was 2.5% greater when running without arm swing ($p = 0.01$).

Table 2.1 Data for subjects running (3 m/s) at the various conditions of target step widths, Arms, and No Arms (mean $\pm$ SEM).

	Target Step Width (%LL)				Arm Swing	
	0	15	20	25	Arms	No Arms
Net Metabolic Power (Watts/kg)	11.09 $\pm$ 0.32*	11.66 $\pm$ 0.32**	11.94 $\pm$ 0.33**	12.40 $\pm$ 0.36**	10.74 $\pm$ 0.25	11.57 $\pm$ 0.23 [†]
step width (%LL)	2.97 $\pm$ 0.58	11.78 $\pm$ 0.91 [•]	14.31 $\pm$ 1.08 [•]	16.74 $\pm$ 1.05 [•]	3.95 $\pm$ 0.90	4.56 $\pm$ 1.13
step width variability (%LL)	2.32 $\pm$ 0.16	2.90 $\pm$ 0.25 ^{••}	3.17 $\pm$ 0.23 ^{••}	3.66 $\pm$ 0.32 ^{••}	2.17 $\pm$ 0.16	2.37 $\pm$ 0.17 [†]
step frequency (Hz)	2.80 $\pm$ 0.05	2.85 $\pm$ 0.07	2.86 $\pm$ 0.08	2.83 $\pm$ 0.09	2.85 $\pm$ 0.6	2.92 $\pm$ 0.06 [†]

Target Step Width Effect

Net Metabolic Power: * significant difference ($p < 0.05$) between 0 %LL and Arms.

** Significant difference ($p < 0.01$) between 15, 20, and 25 %LL and Arms.

Step Width: [•] significant difference (with conservative Huynh-Feldt adjustment, $p < 0.01$) between target step width and Arms conditions.

Step Width Variability: ^{••} significant difference (with conservative Huynh-Feldt adjustment, $p < 0.01$) between target step width and Arms conditions.

Step Frequency: No main effect detected (with conservative Huynh-Feldt adjustment, $p = 0.55$) across the various target step width conditions.

Arm Swing Effect

Arms vs. No Arms: [†] significant difference ($p < 0.05$) between the Arms and No Arms condition.

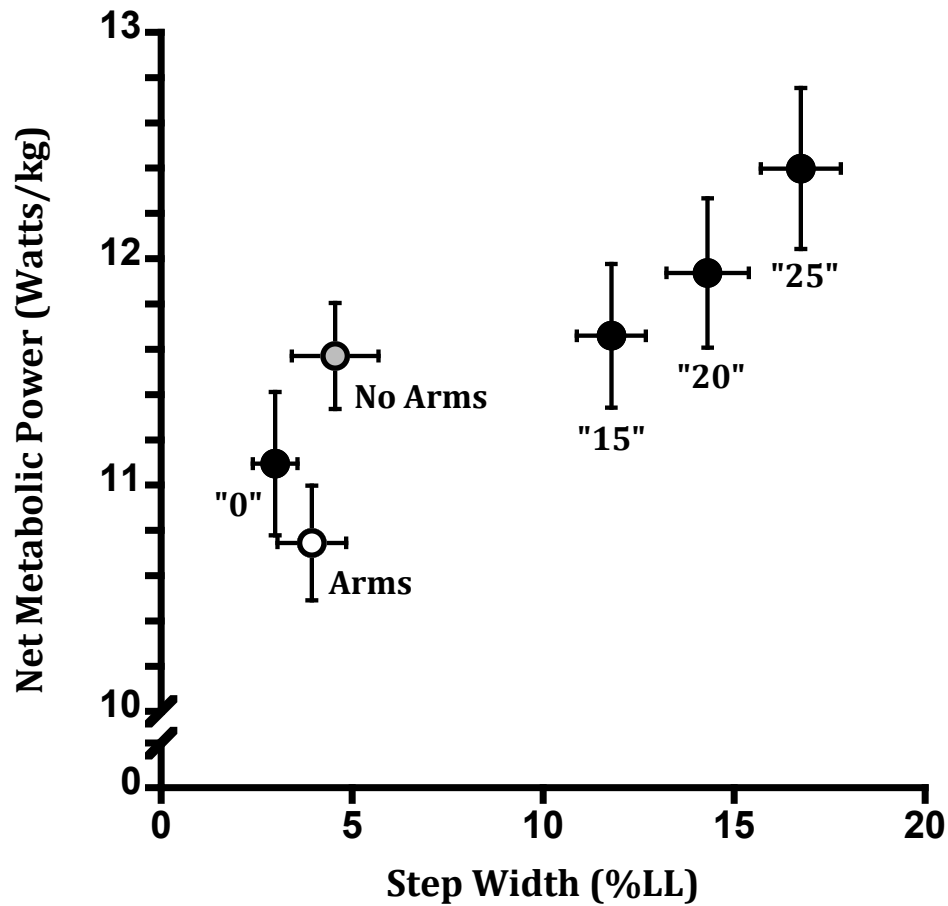


Figure 2.3 Net metabolic power ($n = 10$; mean $\pm$ SEM) versus step width (% LL). Arms denotes running at the preferred step width with arm swing. No Arms denotes running at the preferred step width without arm swing. Quotation marks indicate that subjects were unable to match the target step widths (“%LL”). The data demonstrate that running with step widths other than the preferred step width increases net metabolic power demand. Similarly, running without arm swing increases net metabolic power demand indicating that arm swing is important for reducing energetic cost.

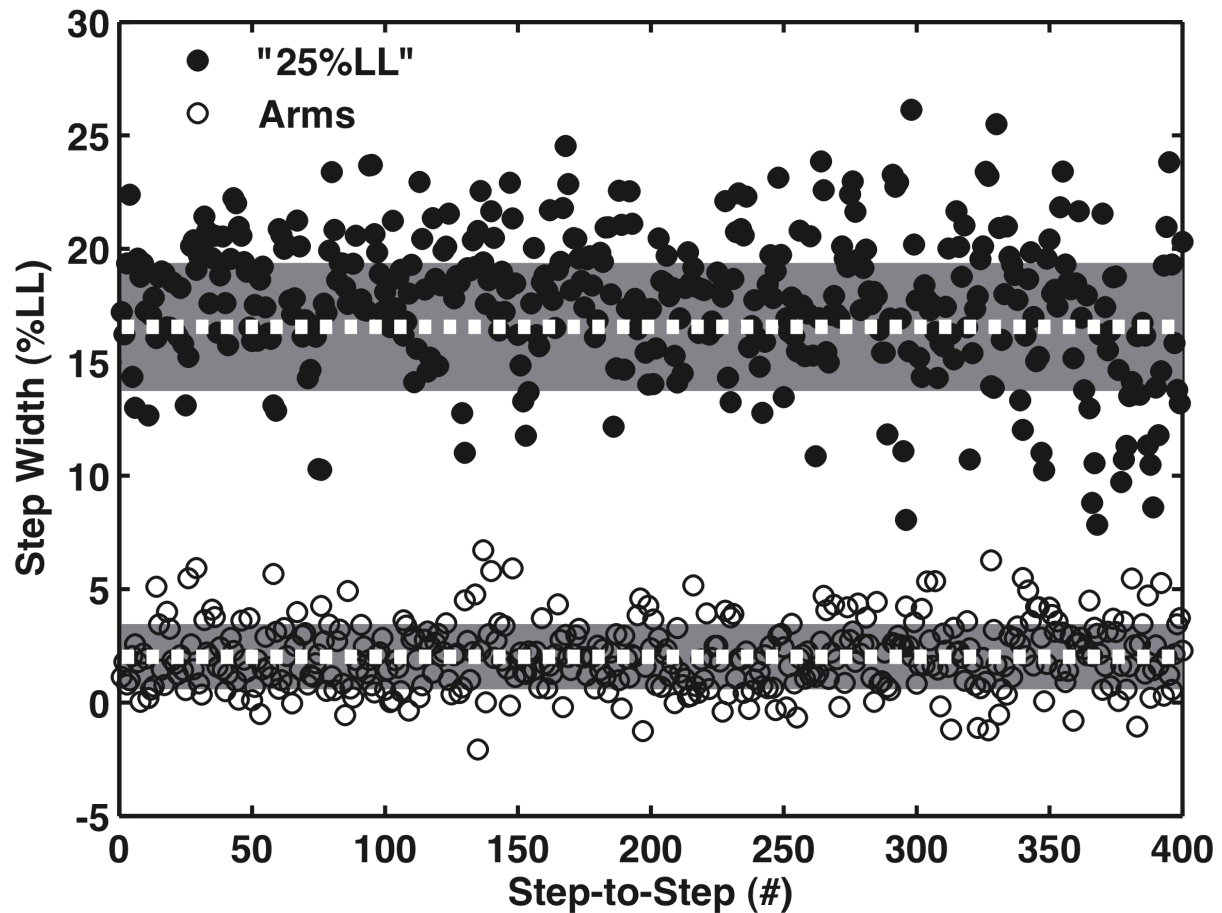


Figure 2.4 Step width (%LL) from step-to-step (#) during the preferred (Arms, open circles) and 25 %LL condition (filled circles) for a single subject. The broken lines (white) represent the average step width and the shaded regions (gray) represent the ($\pm$) standard deviation about the average step width. Note the larger standard deviation in the 25%LL condition as compared to the preferred condition. For clarity, the other target step width conditions are not shown.

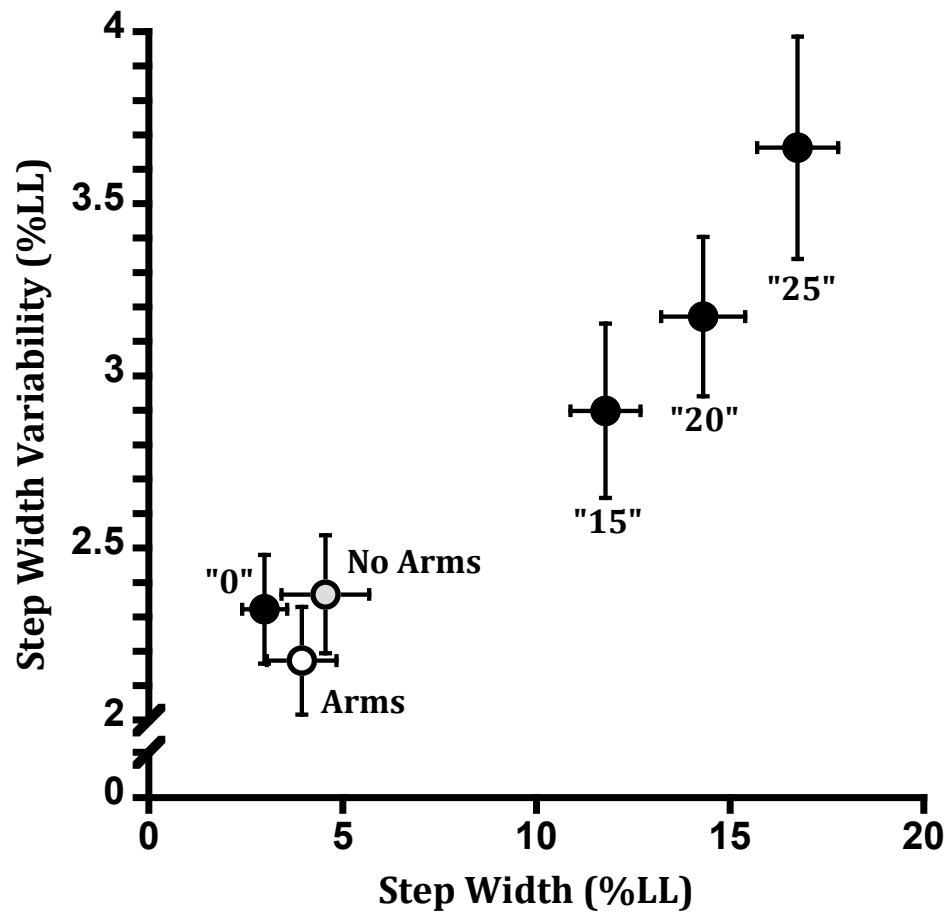


Figure 2.5 Step width variability ($n = 10$; mean $\pm$ SEM) versus step width (% LL). The data demonstrate that running with step width other than the preferred step width increases step width variability. Similarly, running without arm swing increases step width variability indicating that arm swing improves lateral balance.

E. Discussion

In support of our first and second hypotheses, metabolic power and step width variability increased when running with step widths other than preferred. In support of our third and fourth hypotheses, metabolic power and step width variability increased when running without arm swing. These results support our general idea that a narrow step width and arm swing minimize energetic cost and improves lateral balance during running.

Our data demonstrate that running with step widths greater or less than preferred is energetically more expensive. Subjects preferred a step width of only 3.95 %LL (3.6 ± 2.56 cm; mean $\pm$ s.d.) while running with arm swing. Our findings demonstrate a fundamental difference between walking and running, i.e. humans prefer to walk with a moderate step width (8-13 %LL; Donelan et al., 2001; Ortega et al., 2008) but prefer to run with a step width near zero.

Why do humans prefer to run with a step width near zero? McClay and Cavanagh (1994) speculated that a narrow step width has two important biomechanical functions in human running. First, narrow step widths minimize the M-L ground reaction forces generated from step-to-step since foot placement is along the midline of the body. The target step widths (15, 20, 25%LL) in this study reflect a “wide-based” gait and would involve greater kinetic energy fluctuations in the M-L direction. Running with relatively wide steps is mechanically and energetically wasteful since the goal of running is to move the body in the forward direction. Second, narrow step widths minimize the moment generated about the A-P axis, thus reducing the muscular effort required to counteract this moment (Cavanagh, 1987).

Human walking experiments demonstrate that M-L foot placement is an important mechanism for maintaining lateral balance. A small portion ($\sim 3\text{--}6\%$) of the net energetic cost of walking is dedicated toward active control of lateral balance (Donelan et al., 2001; Ortega et al.,

2005). Further, when an external device reduces the need for the active control of lateral balance, humans walk with much narrower step widths and with reduced step width variability (Donelan et al., 2004). Thus, humans maintain lateral balance by walking with a moderate step width but this step width incurs a slight but significant energetic cost. In contrast, humans prefer to run with a step width near zero with minimal step width variability, which suggests there is little need for active control of lateral balance. If this is the case, then an important question remains: Is there any energetic cost to maintaining lateral balance in human running? We hypothesize that there is not a significant energetic cost to maintaining lateral balance.

However, M-L foot placement is not the only mechanism by which humans maintain lateral balance. Running without arm swing decreases lateral balance as indicated by a 9% increase in step width variability compared to running with arm swing. Furthermore, running without arm swing increases the net energetic cost by 8%. Overall, we conclude that arm swing plays an integral part in human running by reducing energetic cost and improving lateral balance. Our findings are in disagreement with Pontzer et al. (2009), who concluded that running without arm swing did not significantly affect energetic cost or lateral balance. Some limitations of their study may explain the different outcomes. First, their metabolic measurements were obtained for a relatively small sample size (six subjects) resulting in low statistical power. In our study, 10 out of 10 subjects exhibited a greater net energetic cost when running without arm swing. Second, Pontzer et al. computed step width and step width variability measures from only 8 consecutive steps, which falls short of the number of steps (~ 400) needed for an accurate measure of step width and step width variability (Owings and Grabiner, 2003).

More generally, Pontzer et al. (2009) proposed a passive arm swing hypothesis, suggesting that the forward and backward motion of the upper arms is derived from the

mechanical energy generated by the swinging legs. If the upper body is modeled as a “passive mass-damped system”, the primary role of the upper arms is to act as a supplementary mass that effectively damps torso and head rotation. The authors presented three lines of evidence in support of this hypothesis. While running with normal arm swing, (1) there was co-activation of the anterior and posterior portions of the deltoid muscle and (2) torso and shoulder rotations appeared to induce arm swing motion. While running without arm swing, (3) there were no significant changes in energetic cost or step width variability. While our data refute the third point, arm swing during running may be partly passive and deserves further investigation.

A limitation of our study was that we were unable to perform a trial with visual feedback of the preferred step width. With a separation distance of up to 13 %LL, the two virtual lines were indistinguishable from a single virtual line. Thus, we chose our minimum target step width to be 15 %LL. As such, it may be considered inappropriate to compare the preferred step width condition (without visual feedback) to the target step width conditions (with visual feedback). However, the preferred step width (3.6 ± 2.6 cm; mean $\pm$ s.d.) in our study is close to zero. The overall interpretation of our results does not change if we compare the 0% LL condition to the other target step width conditions (15, 20, and 25% LL). An experimental condition with a negative step width, i.e. a cross-over gait, might reveal a more definitive U-shape relationship between net metabolic power and step width. A second limitation could be that we did not control for step frequency across running conditions. However, the increase in the energetic cost of running above or below the preferred step width cannot be explained by changes in step frequency since step frequency was similar across conditions (Table 2.1). Compared to running with arm swing, subjects increased step frequency by 2.5% when running without arm swing.

Previous evidence (Cavanagh and Williams, 1982) indicates that a 2.5% increase in step frequency increases $\dot{V}O_2$ by less than 0.5%.

In summary, our data reveal that humans utilize two fundamental mechanisms that minimize energetic cost and improve lateral balance during running: M-L foot placement (i.e. running with narrow step widths) and arm swing. This study is the first to demonstrate that arm swing not only reduces energetic cost but also improves lateral balance by reducing step width variability. An underlying principle that emerges from our results is that there exist U-shaped relationships not only between energetic cost and step width but also lateral balance and step width. In conclusion, humans appear to choose their step width and swing their arms so as to minimize energetic cost and optimize for lateral balance.

3. THE ENERGETIC COST OF MAINTAINING LATERAL BALANCE DURING HUMAN RUNNING

A. Abstract

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 as compared to 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 utilize step width adjustments as the primary mechanism to maintain lateral balance during running.

B. Introduction

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 (Bauby and Kuo, 2000; Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008). In walking, external lateral stabilization (LS) reduces the need for the active control of maintaining lateral balance and thus reduces energetic cost (Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008). Donelan et al. (2004) 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 (Bauby and Kuo, 2000; Donelan et al., 2004) and can therefore explain the reduction in energetic cost. Due to the design of Donelan et al.'s external LS apparatus (Donelan et al., 2004), subjects were required to walk without arm swing, which may exact a greater energetic cost to maintain lateral balance. However, by eliminating arm swing, Donelan's et al. experiment provided important insights into how humans utilize 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% (Collins et al., 2009; Ortega et al., 2008; Umberger, 2008). This increase in energetic cost may be in part due to an increase in the cost to maintain lateral balance. Ortega et al. (2008) found that when walking without arm swing, external LS reduces energetic cost by 6%, similar to the finding of Donelan et al. (2004). On the contrary, Ortega et al. (2008) 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. (2004) was in part 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 (Bruijn et al., 2010; Ortega et al., 2008). As Ortega et al. (2008) 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 (Bruijn et al., 2010), we can presume that arm swing during walking assists with some aspect of lateral balance and based on the study of Ortega et al. (2008), 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) utilized in human walking are also important for maintaining lateral balance during human running. Our previous study (Arellano and Kram, 2011) demonstrated a fundamental difference in how humans utilize step width adjustments to maintain lateral balance in walking versus running. For example, humans prefer to walk with a moderate step width [8-13 %LL; (Donelan et al., 2004; Ortega et al., 2008)] but prefer to run with a step width near zero (Arellano and Kram, 2011). 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 (Arellano and Kram, 2011). 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 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? and 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 as compared to 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 two-fold. First, humans prefer to run at a step width that is not significantly different from zero (Arellano and Kram, 2011). Secondly, we have previously found that when humans run without arm swing, they compensate by increasing their step width (Arellano and Kram, 2011). 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 (Chang and Kram, 1999; Modica and Kram, 2005; Teunissen et al., 2007). Quantifying the energetic cost maintaining lateral balance will help to complete our overall analysis.

C. Methods

Twelve subjects participated in this study (9 males and 3 females, age = 25.8 ± 3.9 years, mass = 67.1 ± 10.0 kg, leg length (LL) = 95.1 ± 5.2 cm; mean $\pm$ sd). Prior to data collection, all subjects read and signed an informed consent document 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 minutes) during which subjects practiced walking (2 trials) and running (2 trials) on a force measuring treadmill (Kram et al., 1998) 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, Salt Lake City, Utah). As described previously (Arellano and Kram, 2011), we placed reflective markers on the left and right heel, dorsum of the 2nd toe, and lateral mid-foot of each shoe to provide real-time visual feedback (Motion Analysis Corporation, Santa Rosa, CA) of foot placement during each trial (Figure 3.1). 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 (1000 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 (Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008). 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 Figure 1B, a separate piece of nylon rope ran from the other end of a rope ratchet (Carolina North Mfg., 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 (Donelan et al., 2004;

Ortega et al., 2008). To accommodate differences in subject height, we adjusted the height of the pulley to ensure that we applied forces horizontally in the lateral direction. Based on 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 $\sim 2,200$ N/m and was most comfortable for walking and running.

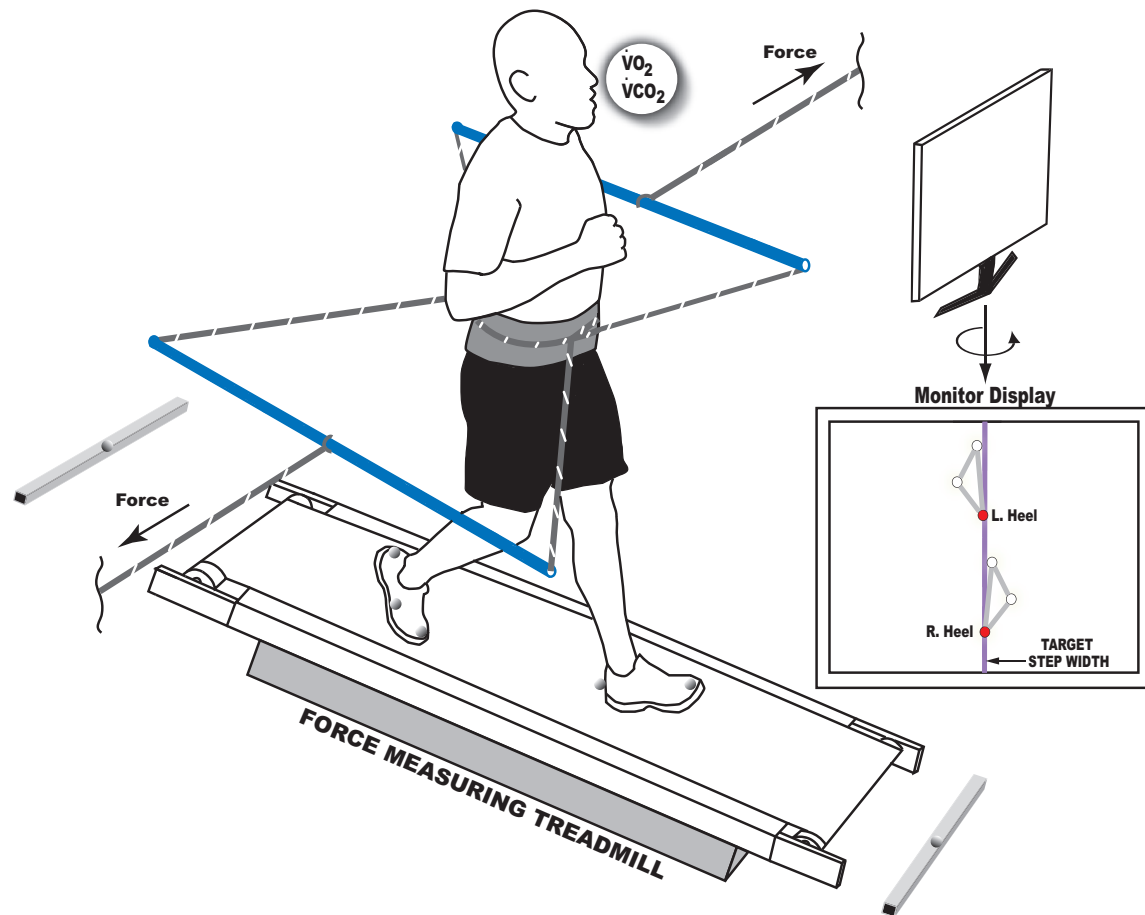


Figure 3.1 We provided external LS and real-time visual feedback of foot placement (monitor providing a top-down view) during all walking and running trials. We applied lateral forces about the waist via lightweight carbon fiber poles (in blue). The poles served two purposes. First, we could apply a 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 subject's 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 ¼ inch rope ratchet (Carolina North Mfg, 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 x 47 cm²) 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).

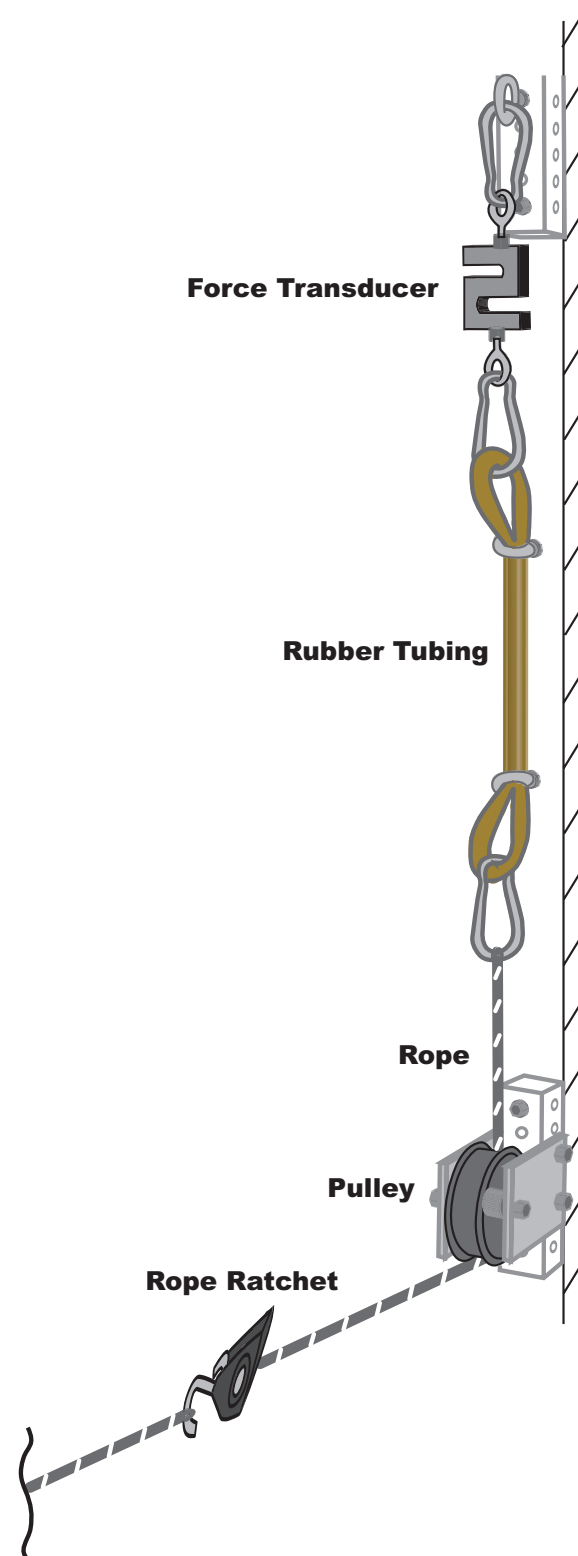


Figure 3.2 From the middle of each pole, we connected a separate rope to one end of a $\frac{1}{4}$ inch rope ratchet in series with a pulley, a piece of latex rubber tubing (resting length ~ 0.20 m, outside diameter ~ 0.02 m, inside diameter ~ 0.01 m, wall thickness ~ 0.003 m), and a force transducer mounted to the wall. We hung the force transducer and rubber tubing vertically to reduce any inertial effects of the external LS system.

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 subject's waist. Using the monitor display (Figure 3.1), 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 seconds. During this time, we recorded the medio-lateral position of the sacrum marker and changes in the lateral forces detected by the left and right force transducers (LabView, National Instruments, Austin, Texas). From the medio-lateral position of the sacrum marker and the net force acting on the subject's waist, we calculated the slope of the net force vs. medio-lateral position curve to yield the effective stiffness of the external LS system (Figure 3.3). Subjects repeated this procedure twice before beginning each walking or running trial to ensure that the appropriate stiffness was applied (Table 3.1). 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 utilized by previous lateral stabilizing human walking experiments (Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008). We provide a detailed description of the logarithmic decrement method in APPENDIX B.

Table 3.1 The effective stiffness (mean $\pm$ SD) of the external LS system and intraclass R values measured for trial 1 and trial 2.

	Prior to Walking (no arm swing)		Prior to Running (arm swing)		Prior to Running (no arm swing)	
Trial	1	2	1	2	1	2
Stiffness (N/m)	2164 (261)	2194 (291)	2250 (270)	2301 (301)	2243 (232)	2281 (223)
intraclass R	0.97 [•]		0.97 ^{••}		0.97 ^{•••}	

intraclass R statistics

•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.

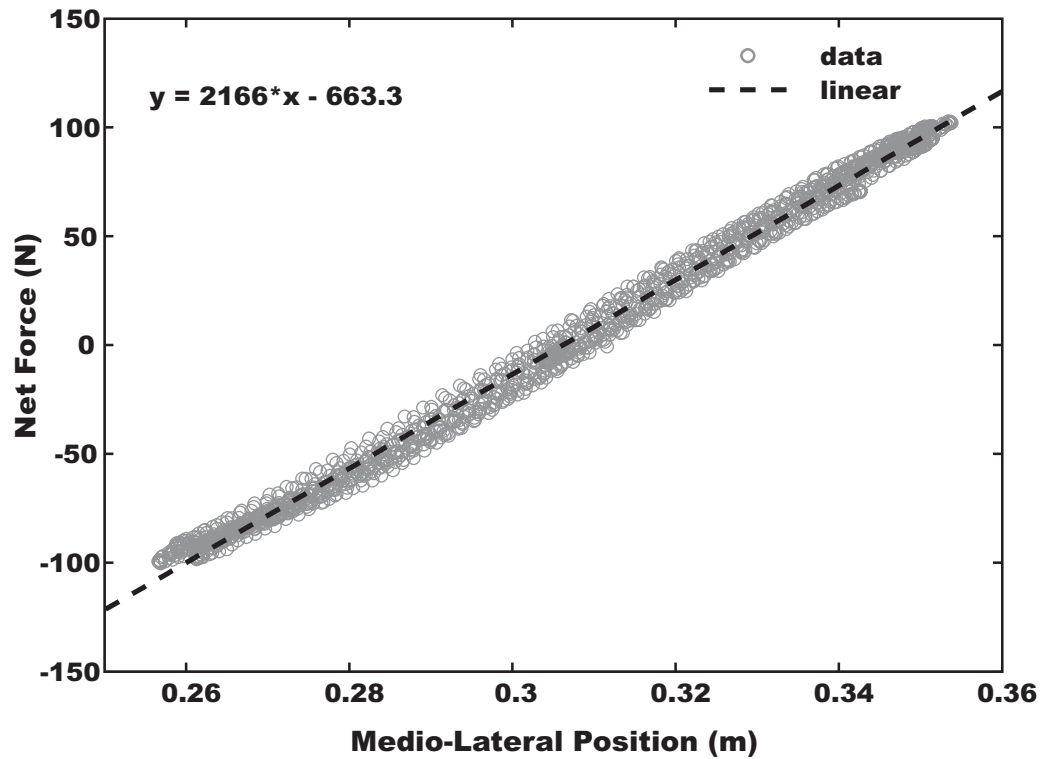


Figure 3.3 Net Force (N) versus medio-lateral position (m) for a single subject swaying about a virtual line projected on the monitor display. We simultaneously recorded the changes in net force and the position of the sacrum marker for 15 seconds (open circles in gray). To estimate the effective stiffness of the external LS system, we applied a least squares linear regression model to the data (dashed line). The slope ($\sim 2,166$ N/m) of the linear regression equation represents the effective stiffness of the external LS system.

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 three minutes (Brockway, 1987). 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 (Arellano and Kram, 2011), we calculated the average step width, step width variability, and step frequency during the last 401 consecutive steps that occurred during the last three minutes for each running trial. Step width was defined as the medio-lateral (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 (Arellano and Kram, 2011; Bauby and Kuo, 2000; Dean et al., 2007; Donelan et al., 2001; Ortega et al., 2008). 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 lateral stabilization 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 lateral stabilization, 2) a within-subjects main effect for arm swing, and 3) an external lateral stabilization-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 Inc., Chicago, IL). All values are reported as mean $\pm$ SEM unless noted otherwise.

D. 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, $F(1,10) = 6.168$; $p = 0.032$; Figure 3.4A) 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, $F(1,10) = 12.578$; $p = 0.005$; Figure 3.4C). 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, $F(1,10) = 0.295$; $p = 0.599$; Figure 3.4B) or step frequency (2.93 ± 0.07 Hz without external LS vs. 2.92 ± 0.06 Hz with external LS, $F(1,10) = 0.041$; $p = 0.843$; Figure 3.4D).

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, $F(1,10) = 34.186$; $p < 0.001$; Figure 3.4A) but did not significantly affect step width (1.90 ± 0.30 %LL without external LS vs. 1.83 ± 0.28 %LL with external LS, $F(1,10) = 0.323$; $p = 0.582$; Figure 3.4B) or step width variability (1.80 ± 0.15 %LL without external LS vs. 1.80 ± 0.13 %LL with external LS, $F(1,10) = 0.001$; $p = 0.975$; Figure 3.4C). 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, $F(1,10) = 6.419$; $p = 0.030$; Figure 3.4D).

External LS-by-Arm Swing Interaction Effect

There were no significant interaction effects for net metabolic power ($F(1,10) = 0.002$; $p = 0.964$; Figure 3.4A), step width ($F(1,10) = 0.276$; $p = 0.611$; Figure 3.4B), or step width variability ($F(1,10) = 0.005$; $p = 0.943$; Figure 3.4C) 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 (Figure 3.4D), 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 $\pm$ sd). RER values less than 1.0 indicate that metabolic energy was provided primarily by aerobic metabolism (Brooks et al., 2004).

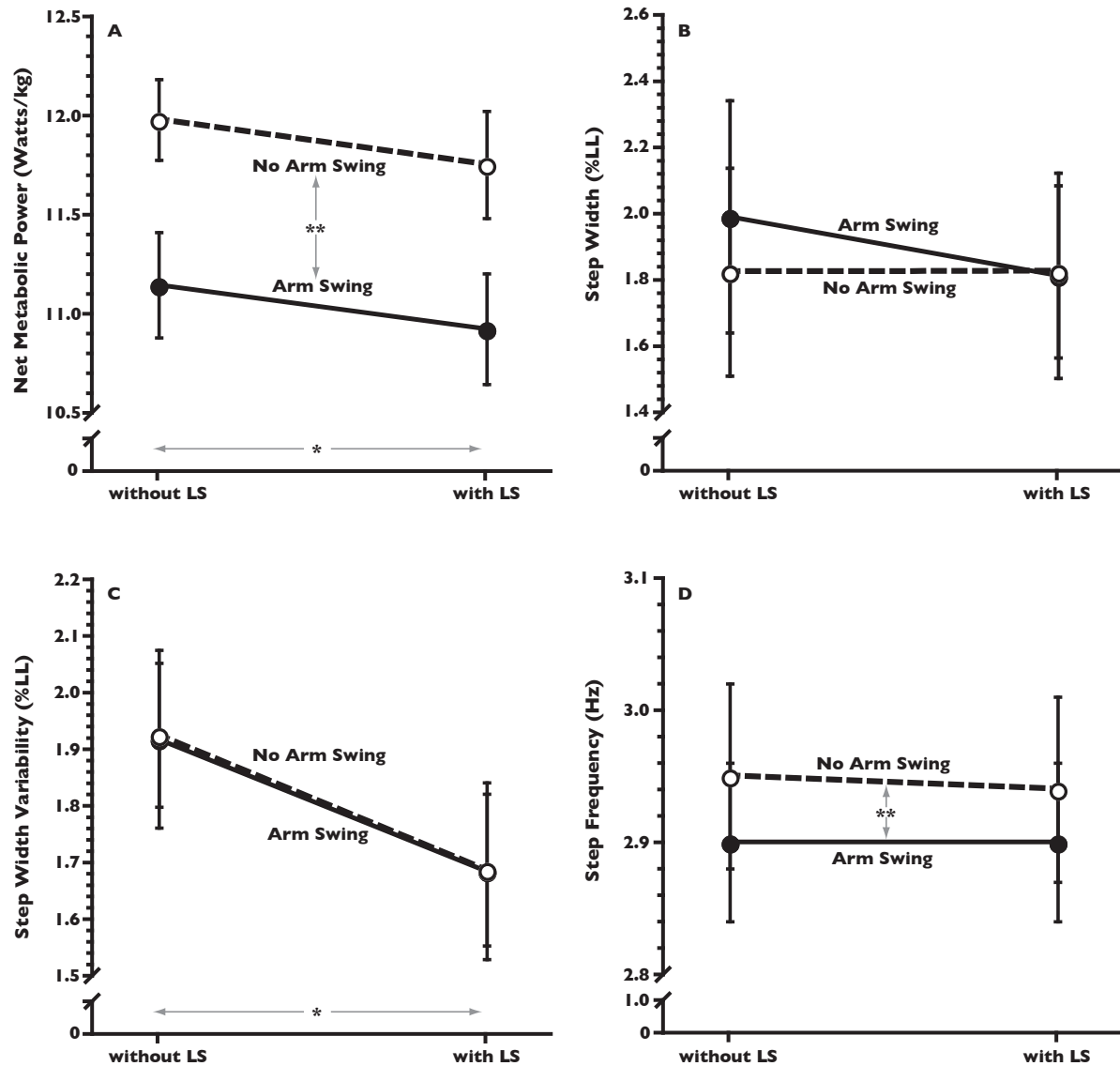


Figure 3.4 Net metabolic power (A), step width (B), step width variability (C), and step frequency (D) while running without and with external LS, both with and without arm swing ($n = 11$; mean $\pm$ SEM). When provided with external LS (solid and dashed lines), subjects significantly reduced net metabolic 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, open circles; Arm Swing, filled circles), 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.

E. 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 W/kg and 0.23 W/kg, respectively. We also found that external LS reduced step width variability by 0.24 %LL 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 (Dean et al., 2007; Donelan et al., 2004). 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 (Bruijn et al., 2008; Hamner et al., 2010; Hinrichs, 1987). When considering the angular momentum about the vertical axis, Hinrichs (Hinrichs, 1987) has shown 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. (Miller et al., 2009) 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 (Miller et al., 2009). 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 (Arellano and Kram, 2011), 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 (Arellano and Kram, 2011).

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 (Figure 3.1).

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 (Cavanagh and Williams, 1982) indicates that a 1.7% increase in step frequency would increase the rate of oxygen consumption by less than 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 (Arellano and Kram, 2011), 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 utilize 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 and without arm swing so that we could compare our findings to previous walking experiments (Dean et al., 2007; Donelan et al., 2004). Similar to Donelan et al. (2004), 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 (Arellano et al., 2009). We defined this instant in time as the beginning of each step and extracted the medio-lateral 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 three minutes 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 3.2).

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

Similar to previous walking experiments (Dean et al., 2007; Donelan et al., 2004), 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. (2004) and Dean et al. (2007), 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. (2007), 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. and interpret these values based on the classification scheme (small: $d = 0.20$; medium: $d = 0.50$; large: $d = 0.80$) presented by Cohen (Cohen, 1988).

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. (2004) and Dean et al. (2007), 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. (2007). Thus, we conclude that our external LS system represents a medium-to-large effect size, comparable to those reported by Donelan et al. (2004) and Dean et al. (2007).

Finally, we found a small, but significant increase in step frequency (1.7%) while walking with external LS as compared to walking without external LS. Based on the findings from Umberger and Martin (2007), 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.

Table 3.2 Net metabolic power, respiratory exchange ratio, step width, step width variability, and step frequency (mean $\pm$ SEM).

	Walking (no arm swing)	
	without LS	with LS
Net Metabolic Power (W/kg)	3.13±0.07	2.96±0.08
	<i>p</i> = 0.018	
step width (% LL)	3.83±0.33	3.70±0.445
	<i>p</i> = 0.568	
step width variability (% LL)	1.16±0.05	1.00±0.06
	<i>p</i> = 0.007	
step frequency (Hz)	1.78±0.04	1.81±0.04
	<i>p</i> = 0.017	
<i>paired t-tests statistics</i>		
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

Similar to previous experiments (Dean et al., 2007; Donelan et al., 2004; Ortega et al., 2008), we modeled our external LS system as a 2nd 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 (Figure 3.5A).

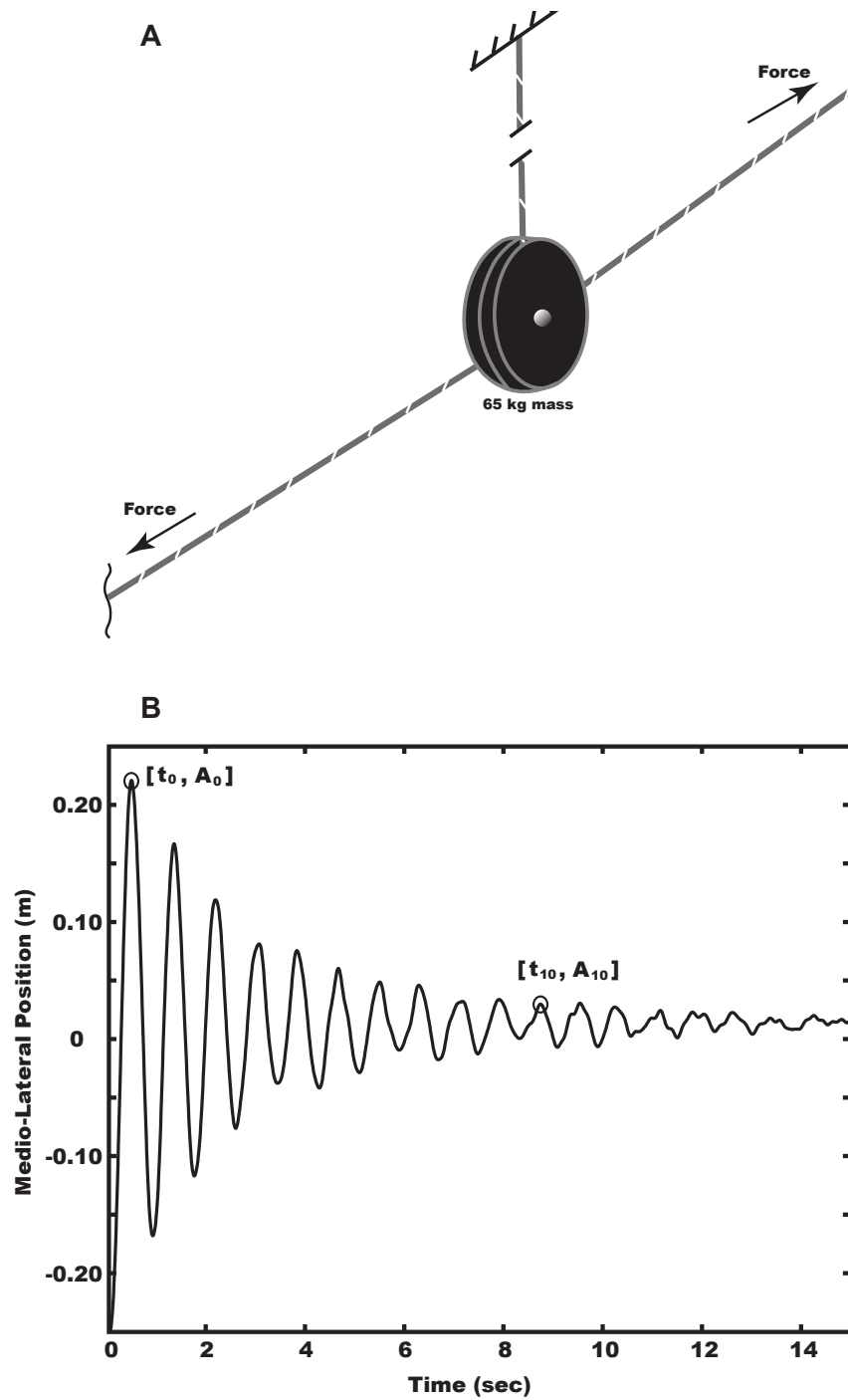


Figure 3.5 Experimental set-up (A) and the medio-lateral decaying oscillation of a known mass attached to the external LS system (B). Assuming a one-degree of freedom oscillation, we can compute the effective stiffness, damping ratio, and damping of the system (see text).

The equation for the 2nd order damped oscillator model is defined as follows:

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

where

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

$c = \text{damping}$, $m = \text{mass}$, $k = \text{stiffness}$

From the logarithmic decrement method, we defined δ as

$$\delta = \frac{1}{10} \ln\left(\frac{A_0}{A_{10}}\right)$$

and the natural period (T_d) between 10 cycles as

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

From our data (Figure 3.5B), we computed the changes in amplitude (A_0 and A_{10}) and the natural period (T_d) as follows:

$$t_0 = 0.5 \text{ and } A_0 = 0.0221$$

$$t_{10} = 8.75 \text{ and } A_{10} = 0.0030$$

Our custom Matlab program yielded the following values:

$$k = 3818.72 \text{ (N/m)}$$

$$c = 31.84 \text{ (N}\cdot\text{s/m)}$$

$$\text{damping ratio} = 0.032$$

As demonstrated, the logarithmic decrement method yielded an effective stiffness value of ~ 3800 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.

4. LATERAL BALANCE AND FOOT PLACEMENT ACROSS RUNNING SPEEDS: A COMPARISON OF SPRINTERS WITH AND WITHOUT TRANS-TIBIAL AMPUTATIONS

A. Abstract

Passive-elastic running-specific prostheses help to partially restore the sagittal-plane, spring-like behavior of the legs during running, but how do running-specific prostheses affect lateral balance? To address this question, we studied 12 non-amputee sprinters and 7 Paralympic sprinters with trans-tibial amputations (6 unilateral/1 bilateral) running across a range of speeds up to maximum sprint speed. We compared step width and its variability between groups as well as medio-lateral (M-L) foot placement relative to the midline of the body and its variability between individual legs. We interpret greater variability as a decrease in lateral balance. Compared to non-amputees, sprinters with unilateral amputations ran with 40% greater step width and 11% greater step width variability. The sprinter with bilateral amputations ran with a step width similar to sprinters with unilateral amputations, but exhibited 50% greater step width variability. M-L foot placement and its variability were symmetric between the biological legs of non-amputees, but were asymmetric between the affected (AL) and unaffected (UL) legs of sprinters with unilateral amputations — the AL exhibited 18% greater M-L foot placement variability. Compared to the AL of sprinters with unilateral amputations, both ALs of the sprinter with bilateral amputations exhibited 125% greater M-L foot placement variability. At faster running speeds, variability of both step width and M-L foot placement increased in all sprinters, indicating progressive decreases in lateral balance. Overall, we find that 1) sprinters with trans-tibial amputations have greater challenges with maintaining lateral balance and 2) maintaining lateral balance is more challenging at faster running speeds.

B. Introduction

Anecdotally, individuals with trans-tibial amputations who use passive-elastic running-specific prostheses find it very difficult to maintain “balance” while running (Sokolove, 2012). Our recent studies of runners with biological legs demonstrate that adjusting step width from step-to-step is an effective strategy for maintaining lateral balance (Arellano and Kram, 2011, 2012). Although passive elastic running-specific prostheses are mechanically designed to mimic the sagittal plane, spring-like behavior of biological legs (Nolan, 2008), they may not meet the biomechanical demands required to overcome instabilities in the medio-lateral (M-L) direction, potentially disrupting lateral balance. Here, we investigated the control of foot placement and lateral balance in sprinters using running-specific prostheses.

We also investigated how increases in running speed affect lateral balance. Our previous experiments of lateral balance focused on only a single, modest speed of running (3.0 m/s). Faster running speeds, up to maximum sprint speed, entail greater peak vertical ground reaction forces, decreases in ground contact time, steady increases in step frequency, and more rapid leg swing (Grabowski et al., 2010; Mero et al., 1992; Weyand et al., 2009). Changes in any of these factors could make it more difficult to maintain lateral balance. Further, maintaining lateral balance at faster running speeds could prove especially difficult for sprinters using running-specific prostheses because these prostheses are designed to restore sagittal plane leg mechanics, but may be ineffective in the frontal plane.

We compared how sprinters with and without trans-tibial amputations maintain lateral balance by measuring step width and its variability. We also measured M-L placement of each foot relative to the midline of the body (McClay and Cavanagh, 1994) and its variability. Although step width provides a general idea of foot placement, it is insensitive to foot placement

asymmetries that may exist between the legs of the same individual. Because individuals with unilateral trans-tibial amputations have been shown to exhibit a range of inter-limb asymmetries while running and sprinting (Buckley, 1999; Enoka et al., 1982; Grabowski et al., 2010; Prince et al., 1992; Sanderson and Martin, 1996), we compared M-L foot placement between the affected leg (AL) and unaffected leg (UL) of sprinters with unilateral trans-tibial amputations as well as the right leg (RL) and left leg (LL) of non-amputee sprinters. We also compared the right affected leg (RAL) and left affected leg (LAL) of a sprinter with bilateral trans-tibial amputations.

We hypothesized that: 1) step width and step width variability would be greater in sprinters with trans-tibial amputations as compared to non-amputee sprinters and 2) step width variability would increase across running speed up to maximum speed in all sprinters. We interpret greater step width variability as indicating a decrease in lateral balance (Arellano and Kram, 2011; Bauby and Kuo, 2000). We further hypothesized that 3) M-L foot placement and its variability would be similar between the RL and LL of non-amputee sprinters and greater in the AL as compared to the UL of sprinters with unilateral trans-tibial amputations. Finally, we hypothesized that 4) M-L foot placement variability would increase across running speed up to maximum speed in all sprinters, again with greater variability indicating a decrease in lateral balance.

C. Methods

Twelve non-amputee sprinters and seven elite Paralympic sprinters with trans-tibial amputations (6 unilateral and 1 bilateral) volunteered and provided informed consent as per the

Intermountain Healthcare Institutional Review Board. All data were collected at the Biomechanics Laboratory of the Orthopedic Specialty Hospital (Murray, Utah).

Anthropometric Measurements

We measured the height, body mass, and leg length of each subject. For subjects with trans-tibial amputations, body mass included their running-specific prosthesis and socket mass (Table 4.1). Leg length was measured as the distance from the greater trochanter to the floor while standing. For each subject with a trans-tibial amputation, we placed a wooden block underneath the contralateral leg to unload the AL while standing.

Table 4.1 Anthropometric and biomechanical characteristics for each unilateral (UL), bilateral (BL), and non-amputee (NA) sprinter. For sprinters with unilateral and bilateral trans-tibial amputations, total mass includes prosthesis mass.

	sex (M/F)	age (years)	height (m)	total mass (kg)	UL (or RL) length (m)	AL (or LL) length (m)	RSP (model)	RSP mass (kg)	top speed (m/s)
1UL	F	35	1.68	66.1	0.955	0.985	Cheetah	1.7	7.0
2UL	F	23	1.69	62.9	0.895	0.950	Sprinter	1.2	8.4
3UL	M	36	1.84	79.9	0.985	1.030	Cheetah	1.6	9.0
4UL	M	27	1.75	69.3	0.930	0.970	Cheetah	1.5	9.3
5UL	M	29	1.87	109.1	0.985	1.065	C-Sprint	1.4	9.5
6UL	M	25	1.85	71.0	1.005	1.025	Cheetah	1.2	9.7
1BL	M	20	1.75	67.7	0.995	1.005	Cheetah	1.7	8.9
1NA	F	16	1.70	59.5	0.900	0.900	-	-	7.0
2NA	F	18	1.70	56.4	0.885	0.890	-	-	7.6
3NA	F	28	1.74	60.9	0.952	0.945	-	-	8.2
4NA	M	40	1.69	88.6	0.885	0.890	-	-	9.0
5NA	M	16	1.75	69.1	0.920	0.935	-	-	9.0
6NA	M	16	1.82	70.7	0.950	0.955	-	-	9.0
7NA	M	19	1.88	89.1	0.990	1.000	-	-	9.0
8NA	M	21	1.84	75.0	1.000	1.000	-	-	9.0
9NA	M	18	1.79	76.8	0.910	0.915	-	-	9.1
10NA	M	17	1.80	78.2	0.960	0.940	-	-	9.2
11NA	M	18	1.80	73.2	0.905	0.915	-	-	9.3
12NA	M	18	1.85	84.1	0.985	0.980	-	-	9.4

Experimental Protocol

After a 5-10 minute warm up period at a comfortable running speed, each subject performed brief running trials (~10-30 strides) on a high-speed 3D force-sensing motorized treadmill (Figure 4.1A; Athletic Republic, Park City, UT). During each trial, we simultaneously collected ground reaction forces (2400 Hz) and whole body kinematics from the 3D positions of reflective markers placed on the body (200 Hz; Motion Analysis Corporation, Santa Rosa, CA). A full-body marker set defined the position of the subject's head, trunk, arms, legs, and running-specific prosthesis. Each subject began the series of trials at 3 m/s and we incremented the speed by 1 m/s until approaching maximum speed, defined as the fastest speed at which the subject could maintain the same position on the treadmill for at least 8 strides (Weyand et al., 2009).

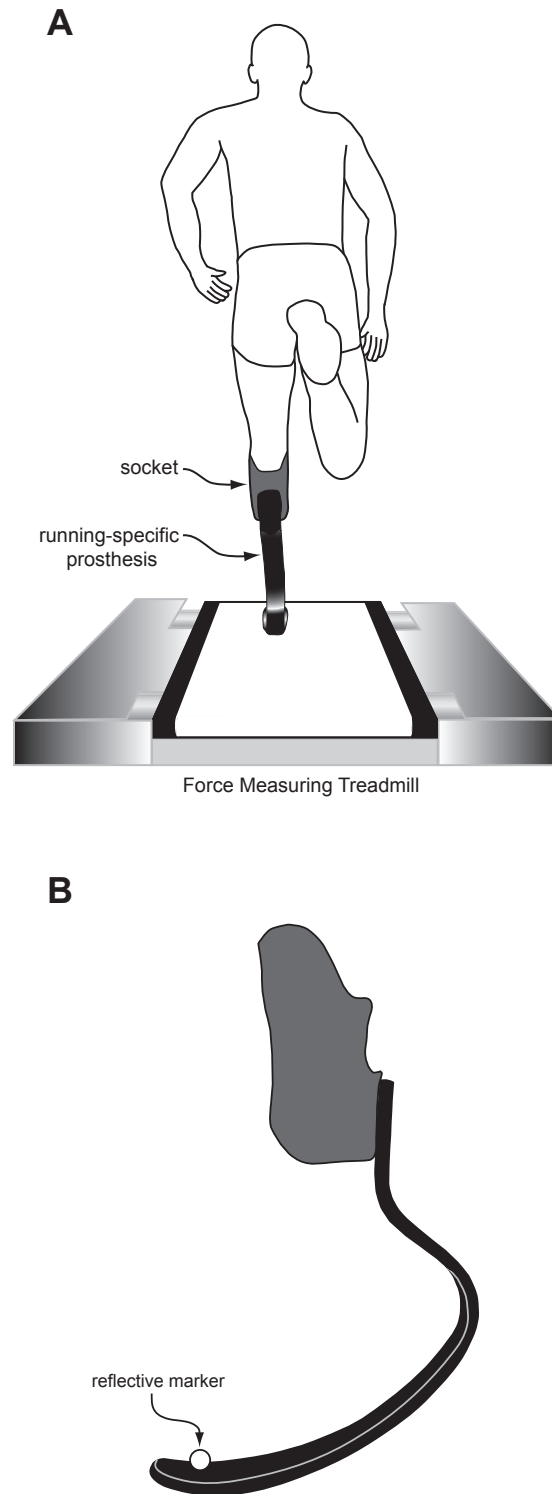


Figure 4.1 (A) Rear view of a sprinter with a left-side trans-tibial amputation running on a force-measuring treadmill. (B) Side view of the running-specific prosthesis. For measurements of step width and M-L foot placement relative to the midline, we placed reflective markers at the distal end of the running-specific prosthesis(es) and heel of the shoe for the UL.

Data Analysis

We measured step width, M-L foot placement relative to midline, and their variability using the position data from reflective markers placed on the left and right heel of each shoe for non-amputee sprinters, on the heel of the shoe for the UL and distal end of the running-specific prosthesis of the AL for sprinters with unilateral amputations, and on the distal ends of the running-specific prosthesis for the RAL and LAL for the sprinter with bilateral amputations (Figure 4.2). For all subjects, we treated the pelvis center of mass (COM) as the midline of the body. We modeled the pelvis as an elliptical cylinder (Hanavan, 1964) with dimensions defined by markers placed on bilateral iliac crests, anterior-superior iliac spines, greater trochanters, posterior-superior iliac spines, and sacrum. The location of the pelvis COM was defined as the geometric center of the cylinder (Visual 3D, C-Motion Inc., Germantown MD.).

For each running trial, we calculated average step width and step width variability (Arellano and Kram, 2011, 2012). Step width was defined as the medio-lateral (M-L) distance between the heel (or prosthesis(es)) markers during successive instants of initial contact. Step width variability was defined as the standard deviation about the average step width. We identified initial contact as the instant when the vertical ground reaction force reached a 40 N threshold at the beginning of each step. At each instant of initial contact, we also defined foot placement relative to the midline as the M-L distance between the heel (or prosthesis) marker and the pelvis COM. Foot placement variability was defined as the standard deviation about the average foot placement relative to the midline (methods are illustrated and described in Figure 4.3).

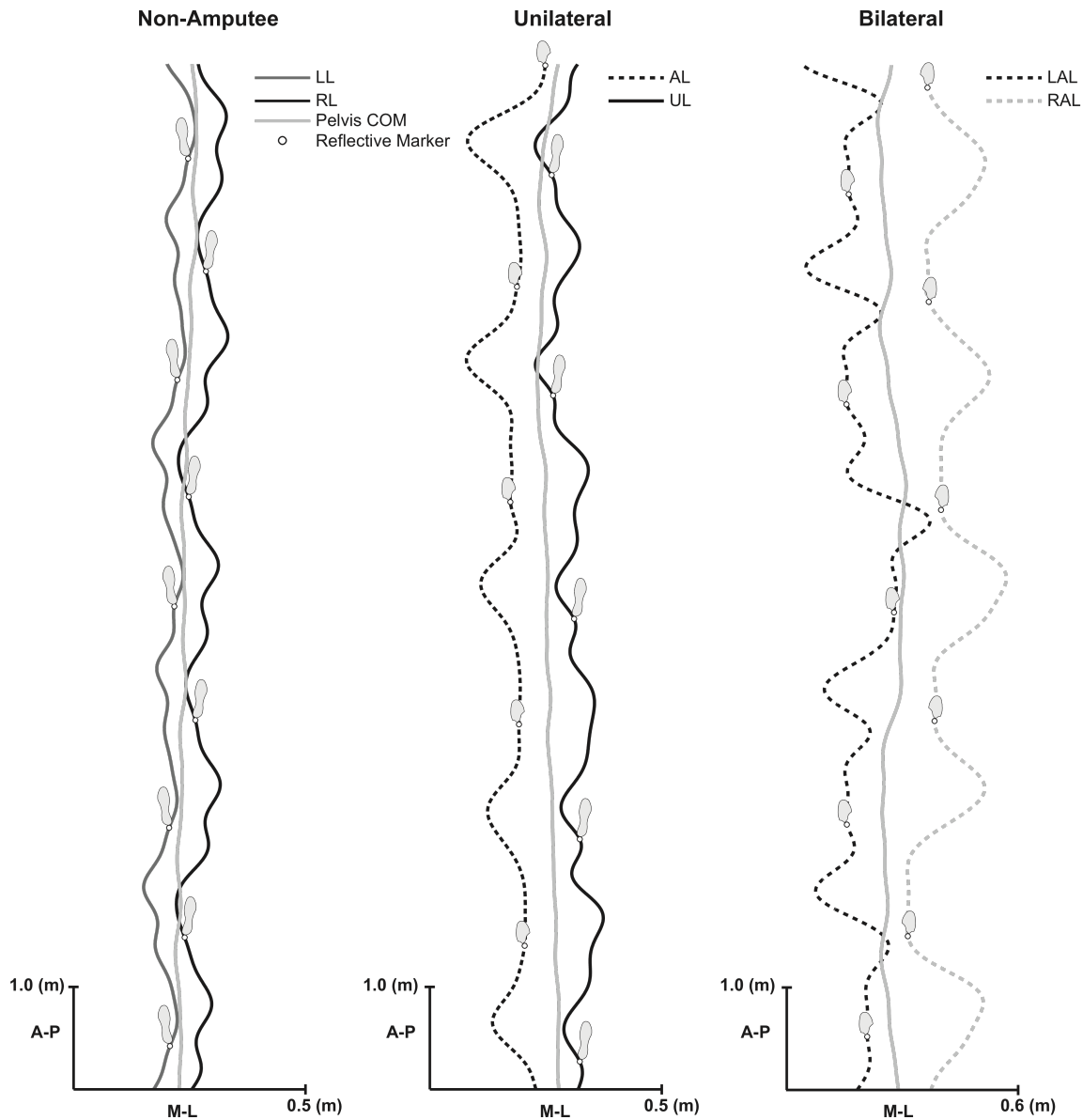


Figure 4.2 Representative foot placement trajectories while running at 3 m/s. For a sprinter without trans-tibial amputations (Non-Amputee), a sprinter with a unilateral trans-tibial amputation (Unilateral), and a sprinter with bilateral trans-tibial amputations (Bilateral), the position of the reflective markers (open circles) indicates the instances of initial contact for each step while running. Note that foot placement of the unilateral (AL) and bilateral (LAL and RAL) sprinter are depicted by the “foot print” based on the distal portion of a running shoe sole bonded to the bottom of the running-specific prosthesis. We depict foot placement of the biological legs (LL, RL, and UL) by the outline of the entire sole of a running shoe (not drawn to scale).

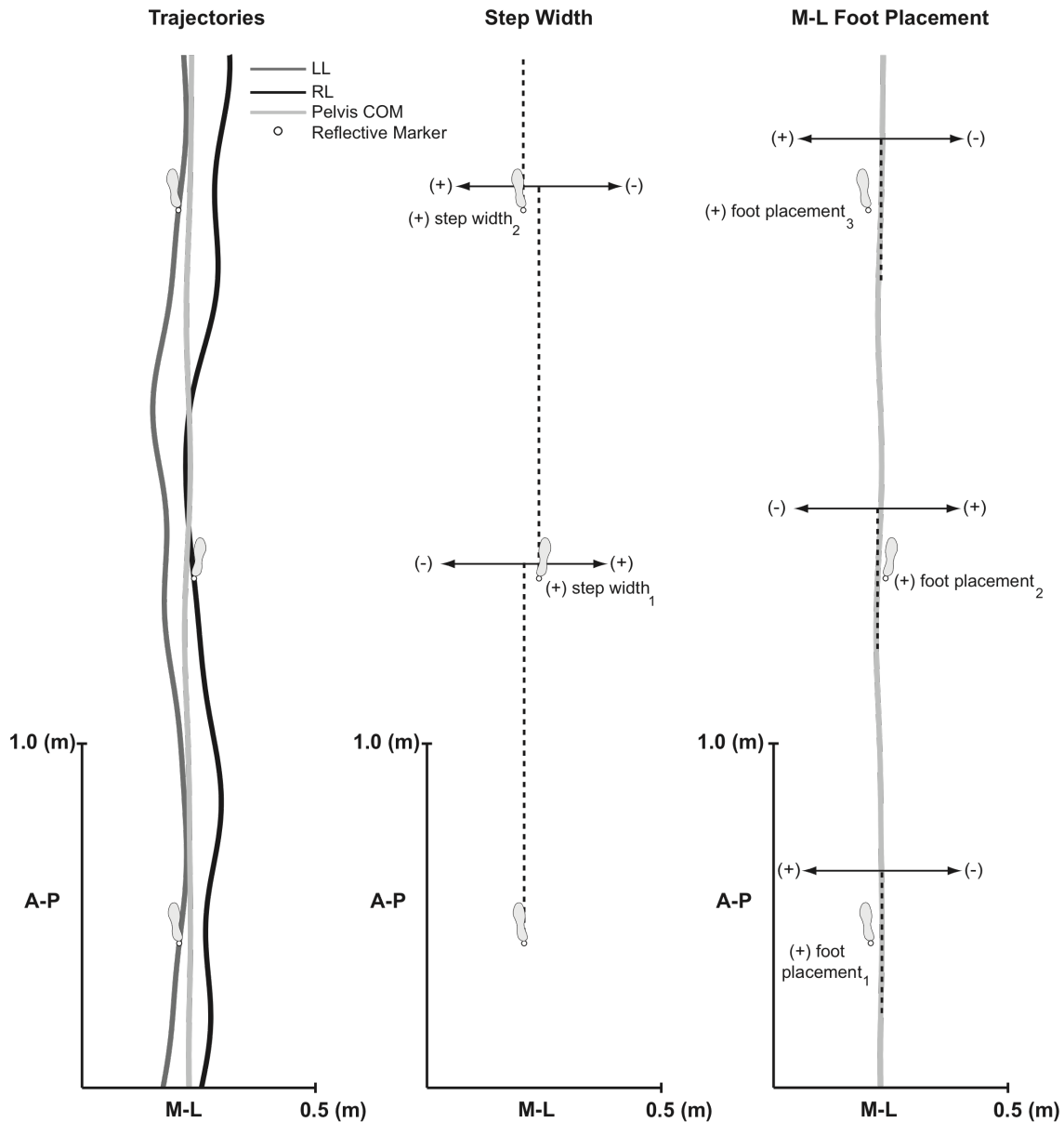


Figure 4.3 (Left column) To measure the step width and M-L foot placement relative to the midline of the body, we extracted the position of the reflective markers (open circles) at the instances of initial contact for each step while running. (Middle column) We defined step width as the M-L distance between markers during successive instants of initial contact. In this example, the non-amputee sprinter made initial contact with the left foot first. The position of the left heel marker defines the neutral axis (dashed reference line) for the position of the right heel marker at initial contact. Placing the right foot to the right of the line is considered a positive step width, to the left of the line a negative step width, and placement on the line as zero. For the subsequent step, the sign convention changes. (Right column) We defined M-L foot placement relative to the midline of the body as the distance between the heel marker and the pelvis COM. Following a similar convention as McClay and Cavanagh (1994), the position of the pelvis COM defines the midline of the body (dashed reference line). Placing the left foot to the left of the line is considered a positive foot placement, to the right of the line a negative foot placement, and placement on the line as zero. For the subsequent step, the sign convention changes.

Statistical Analysis

For analyses comparing between sprinters with unilateral amputations and non-amputee sprinters and between the legs of the same individual, we collapsed data across speed since every subject achieved a different maximum speed (Table 4.1). We examined differences in step width and step width variability using a MANOVA with group (sprinters with unilateral amputations vs. non-amputee sprinters) as a between subjects factor. With respect to M-L foot placement and its variability, we performed separate repeated measures MANOVAs to compare between the UL and AL of the group with unilateral amputations and the RL and LL of the non-amputee group. Here, we defined the leg as the within subjects fixed factor. To determine the direction and strength of the relationship between variables, we followed the MANOVA with separate correlation and linear regression analyses for each dependent variable (i.e. step width, M-L foot placement, and its variability) with speed normalized to maximum speed as the independent variable. Correlation and linear regression analyses were performed separately for sprinters with unilateral amputations ($n = 6$) and non-amputee sprinters ($n = 12$). Statistical significance for all analyses was set at an α level = 0.05 (SPSS Inc., Chicago, IL).

For the sprinter with bilateral trans-tibial amputations ($n = 1$), we report average data along with regression lines to demonstrate the relation between the dependent variables and speed normalized to maximum speed. We recognize that this sprinter represents a special case and therefore we did not attempt statistical inference.

D. Results

Step width

Compared to non-amputee sprinters, sprinters with unilateral trans-tibial amputations ran with a 40% greater step width and 11% greater step width variability (Table 4.2). Across the 2.3

fold increase in running speed (Figure 4.4a,d), non-amputee sprinters decreased step width by 72% ($r = -0.65$, $p < 0.001$) but exhibited 77% greater step width variability ($r = 0.50$, $p < 0.001$). On the contrary, sprinters with unilateral trans-tibial amputations ran with a similar step width ($r = -0.06$, $p = 0.34$) but exhibited a 19% increase in step width variability ($r = 0.20$, $p = 0.08$, Figure 4.4b,e), although not significant at the 0.05 level.

The sprinter with bilateral trans-tibial amputations ran with an average step width of 9.6 ± 2.9 cm, which falls within the range of step widths exhibited by sprinters with unilateral trans-tibial amputations (Figure 4.4c). However, when compared to sprinters with a unilateral amputation, he ran with 50% greater step width variability (6.2 ± 1.8 cm). Regression analysis did not reveal a trend for step width across speed ($r = -0.06$), however, when running from his slowest to maximum speed, step width variability increased by 99% ($r = 0.74$, Figure 4.4f).

Table 4.2 Mean, standard deviation, and MANOVA results for comparisons of step width and step width variability (data collapsed across speed).

	Non-Amputee (<i>n</i> = 12)		Unilateral (<i>n</i> = 6)		Univariate ANOVA	<i>p</i> value
	M	SD	M	SD		
step width (cm)	5.6	3.6	9.3	5.3	$F_{1,141} = 24.53$	< 0.001
step width variability (cm)	2.7	0.9	3.0	0.8	$F_{1,141} = 5.22$	0.002

A one-way MANOVA revealed a significant main effect for group on both step width and step width variability: $F_{2,140} = 20.77$, $p < 0.001$, Wilk's $\lambda = 0.77$. Since significance was detected, we followed up with univariate ANOVAs.

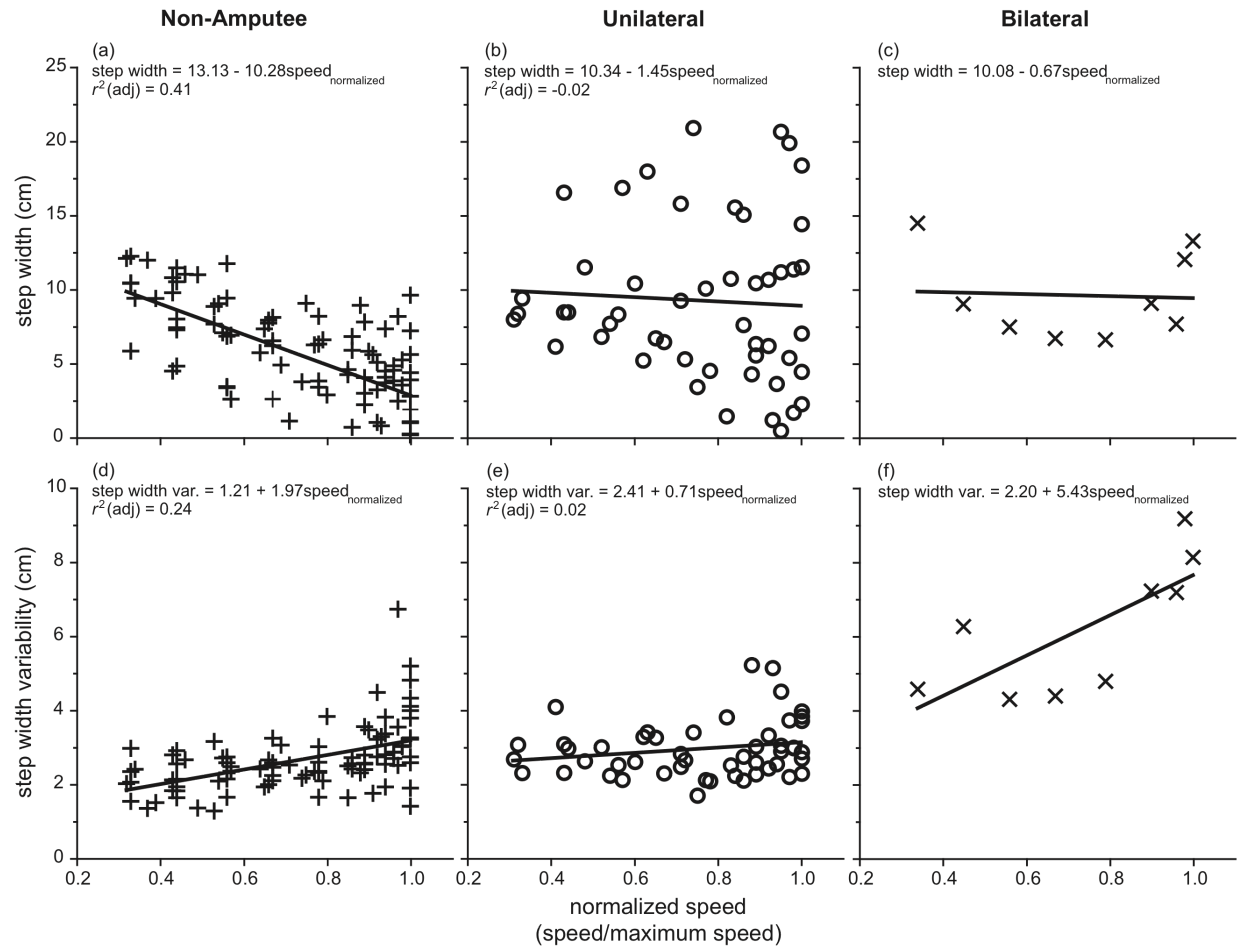


Figure 4.4 Step width and step width variability across running speeds up to maximum sprint speed. (a) As non-amputee sprinters approached maximum speed, they decreased their step width to near zero (b and c) while sprinters with trans-tibial amputations did not show a strong tendency to decrease their step width as they approached maximum speed. As speed increased up to maximum sprint speed, step width variability increased in all sprinters (d, e, and f), indicating progressive decreases in lateral balance. For each variable, we present the equation and adjusted r^2 values from a least-squares linear regression analysis.

M-L Foot Placement

For non-amputee sprinters, M-L foot placement and its variability were symmetrical between the RL and LL (Table 4.3). Across the 2.3 fold increase in speed, the average difference in M-L foot placement and its variability between the RL and LL was less than 0.5 cm and 0.3 cm, respectively. Regression analysis revealed that as non-amputee sprinters ran faster, they placed each foot closer to the midline of the body ($r = -0.45$, $p < 0.001$ and $r = -0.43$, $p < 0.001$, respectively). At the slowest speed, non-amputee sprinters placed their foot 5.4 cm lateral to the midline. At maximum speed, they placed their foot 2.9 cm lateral to the midline, a 46% reduction (Figure 4.5a). In addition, M-L foot placement variability of the RL and LL increased by 133% at maximum speed as compared to the slowest speed ($r = 0.52$, $p < 0.001$ and $r = 0.56$, $p < 0.001$, respectively; Figure 4.5d).

For sprinters with unilateral trans-tibial amputations, M-L foot placement and its variability were significantly different between the AL and UL. For their AL, subjects placed the distal portion of the prosthesis ~1.6 cm closer to the midline as compared to the UL (Table 4.3). Compared to the UL, the AL exhibited 18% greater M-L foot placement variability. Regression analyses showed no significant relationships for M-L foot placement of the UL or AL across speeds ($r = 0.03$, $p = 0.424$ and $r = -0.09$, $p = 0.269$, respectively; Figure 4.5b). Across the 2.3 fold increase in speed, the UL and AL exhibited a 55% and 69% increase in M-L foot placement variability, respectively ($r = 0.40$, $p = 0.002$ and $r = 0.34$, $p = 0.007$, respectively; Figure 4.5e).

For the sprinter with bilateral trans-tibial amputations, M-L foot placement of the LAL (3.7 ± 1.9 cm) and RAL (4.1 ± 1.6 cm) were similar across speed and fell within the range observed in sprinters with unilateral trans-tibial amputations (Figure 4.5c). M-L foot placement variability was also similar between the RAL (4.8 ± 2.5 cm) and LAL (4.2 ± 1.1 cm) but was on

average 125% greater than the AL of unilateral sprinters. Regression analyses did not reveal strong relationships between M-L foot placement of the RAL or LAL and speed ($r = -0.02$ and $r = 0.25$, respectively; Figure 4.5c). When compared to 3 m/s, at his maximum speed, the bilateral sprinter exhibited a 285% increase in M-L foot placement variability for both the RAL and LAL ($r = 0.79$ and $r = 0.85$, respectively; Figure 4.5f).

Table 4.3 Mean, standard deviation, and MANOVA results for comparisons of M-L foot placement and M-L foot placement variability (data collapsed across speed).

	Non-Amputee (<i>n</i> = 12)					
	RL		LL			
	M	SD	M	SD	Univariate ANOVA	<i>p</i> value
M-L foot placement (cm)	4.1	1.7	3.7	1.9	-	-
M-L foot placement variability (cm)	1.7	0.7	1.7	0.7	-	-
A one-way MANOVA revealed no significant main effect for M-L foot placement and M-L foot placement variability between the RL and LL: $F_{2,91} = 3.08$, $p = 0.051$, Wilk's $\lambda = 0.94$. Since significance was not detected, we did not follow up with univariate ANOVAs.						
	Unilateral (<i>n</i> = 6)					
	AL		UL			
	M	SD	M	SD	Univariate ANOVA	<i>p</i> value
M-L foot placement (cm)	4.4	1.9	6.0	3.6	$F_{1,49} = 16.74$	< 0.001
M-L foot placement variability (cm)	2.0	0.9	1.7	0.5	$F_{1,49} = 10.03$	0.003
A one-way MANOVA revealed a significant main effect for M-L foot placement and M-L foot placement between the AL and UL: $F_{2,48} = 11.99$, $p < 0.001$, Wilk's $\lambda = 0.67$. Since significance was detected, we followed up with univariate ANOVAs.						

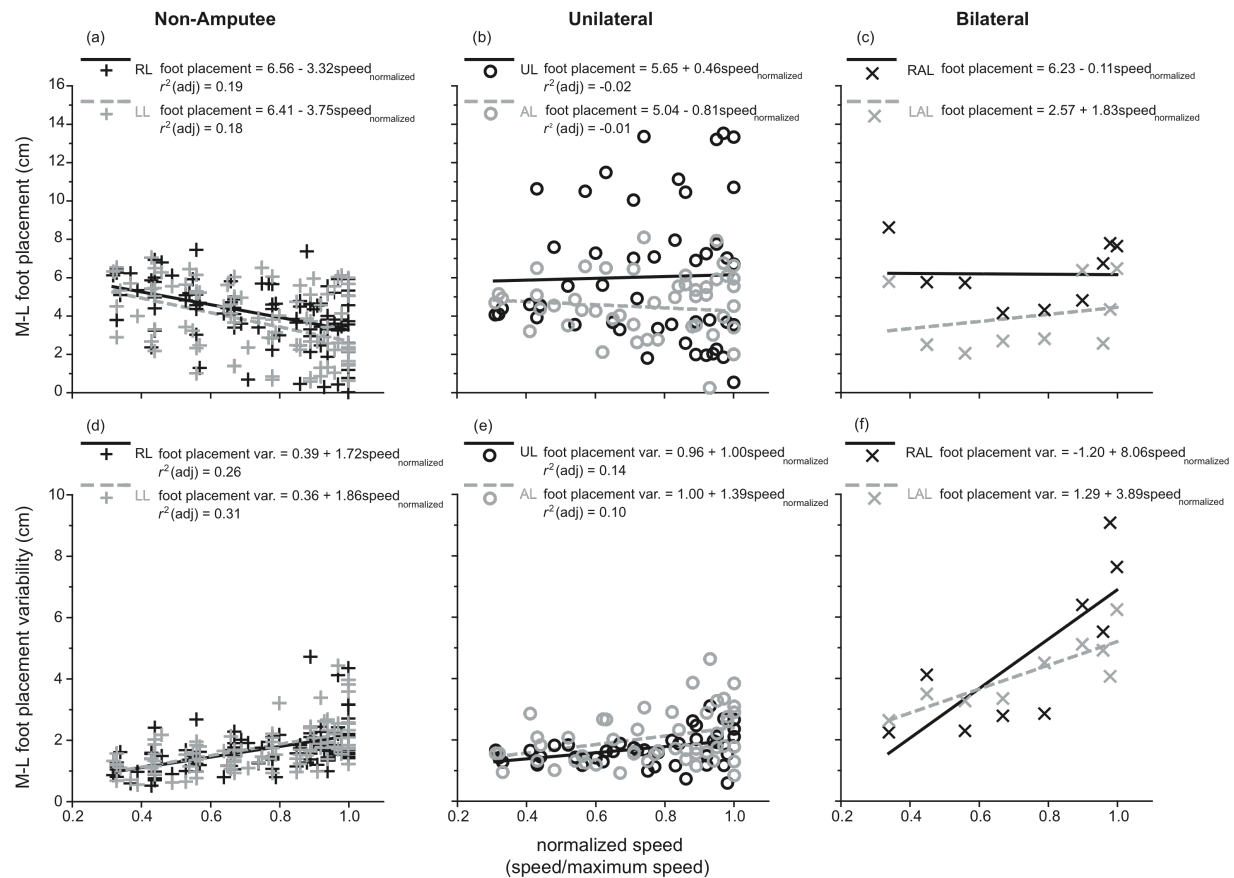


Figure 4.5 M-L foot placement and M-L foot placement variability across running speeds up to maximum sprint speed. (a and d) Across running speeds, non-amputees exhibited symmetrical foot placement patterns between their RL and LL. As non-amputee sprinters approached maximum speed, they placed their foot closer to the midline of the body; however, each leg exhibited greater M-L foot placement variability. (b and e) On the contrary, sprinters with unilateral trans-tibial amputations exhibited asymmetrical foot placement patterns between the AL and UL. They tended to place the AL closer to the midline as compared to the UL, but the AL exhibited greater M-L foot placement variability. (c and f) The sprinter with bilateral trans-tibial amputation also exhibited asymmetrical foot placement patterns. Although he tended to place the LAL closer to the midline of the body as compared to the RAL, M-L foot placement was similar across speed. On the contrary, both the LAL and RAL exhibited a large increase in M-L foot placement variability at faster running speeds. For each variable, we present the equation and adjusted r^2 values from a least-squares linear regression analysis.

E. Discussion

In support of our 1st and 2nd hypotheses, we found that step width and its variability were greater in sprinters with trans-tibial amputations as compared to non-amputee sprinters and that step width variability was greater at faster speeds in all sprinters. In support of our 3rd hypothesis, we found that M-L foot placement and its variability were similar in the RL and LL of non-amputee sprinters, indicating a high degree of symmetry between legs. Although the sprinters with unilateral trans-tibial amputations placed their AL slightly closer to the midline as compared to their UL, the AL exhibited greater M-L foot placement variability. Finally, in support of our 4th hypothesis, M-L foot placement variability was greater at faster running speeds in all sprinters.

At the slowest running speed (3.0 m/s), non-amputee sprinters adopted a relatively wide step width of ~ 10 cm; much greater than that of recreational, distance runners who preferred an average step width of ~ 3.6 cm (Arellano and Kram, 2011). This relatively wide step width could be due to subtle differences in the running form exhibited by those trained to sprint as opposed to those trained to run long distances. Nonetheless, as non-amputee sprinters approached maximum speed, their step width decreased to near zero, which coincided with steady increases step width variability, indicating progressive decreases in lateral balance. These trends were similar for M-L foot placement: non-amputee sprinters placed the foot of their RL and LL closer to the midline as they approached maximum speed.

On the contrary, sprinters with unilateral or bilateral trans-tibial amputations did not show a strong tendency to decrease their step width or to place their foot closer to the midline as they approached maximum speed. Sprinters with unilateral amputations may place each foot slightly away from the midline to compensate for large inertial and ground force asymmetries

that exist between the AL and UL (Grabowski et al., 2010). Another possibility is that sprinters with unilateral amputations lose fine muscular control due to the loss of the foot and ankle as well as the loss of proprioception in the AL. As demonstrated by the sprinter with bilateral trans-tibial amputations (Figure 4.5c and f), M-L foot placement and its variability between the RAL and LAL were highly asymmetric across running speed. It seems reasonable that the loss of proprioception and fine muscular control in both legs would make it more difficult to control foot placement from step-to-step. Nonetheless, placing the foot further away from the midline may be a compensatory foot placement strategy for maintaining lateral balance in sprinters using running-specific prostheses.

As expected, we found similar changes in M-L foot placement and its variability between the RL and LL across speed, reflecting symmetry between the biological legs of non-amputee sprinters. However, M-L foot placement and its variability were significantly asymmetric between the AL and UL of sprinters with unilateral trans-tibial amputations. Most notably, at faster speeds up to maximum speed, the variability of the AL increased to a greater extent than the UL (Figure 4.5e), indicating greater difficulty with the active control of lateral balance when using the AL. Other scientists, who have studied individuals with trans-tibial amputations, have highlighted important asymmetries in sagittal plane kinematics and kinetics while running and sprinting (Buckley, 1999; Grabowski et al., 2010; McGowan et al., 2012; Prince et al., 1992; Sanderson and Martin, 1996). Here, we demonstrate that sprinters with unilateral trans-tibial amputations also exhibit asymmetries in the frontal plane.

Although we analyzed step width and M-L foot placement relative to the midline, we also noticed that sprinters with unilateral and bilateral trans-tibial amputations exhibited a greater amount of lateral circumduction of their affected legs during the swing phase, based on their foot

trajectories (Figure 4.2). Some sprinters using running-specific prostheses have a limited ability to flex the knee (Buckley, 1999) and do not have the ability to dorsiflex the “foot”, which are important mechanisms for clearing the leg during the swing phase of running and sprinting. Furthermore, the length of the AL was slightly longer than the UL of sprinters with unilateral trans-tibial amputations, perhaps making the action of clearing the distal portion of the running-specific prosthesis even more difficult. Circumducting the AL more laterally during the swing phase may be a compensatory strategy for sprinters with trans-tibial amputations who cannot easily flex their knee, but this could potentially lead to inaccuracies and greater variability in foot placement from step-to-step.

In our most intriguing case, the sprinter with bilateral trans-tibial amputations exhibited the greatest increase in step width and M-L foot placement variability across running speed up to maximum speed. When compared to the other sprinters, our data suggest that he had the greatest difficulty with maintaining lateral balance. However, we interpret these findings with caution because we collected data on just one elite sprinter with bilateral trans-tibial amputations and at the time, he had only 2 years of experience using running-specific prostheses. In our future work, we hope to investigate lateral balance in additional sprinters with bilateral trans-tibial amputations and to determine if more experience with the use of running-specific prostheses improves the ability to maintain lateral balance.

Apparent limitations of this study include: the relatively small sample size of sprinters with unilateral trans-tibial amputations, the number of steps used for our foot placement analysis, and using the pelvis COM as a proxy for the midline of the body. Although our sample size of sprinters with trans-tibial amputations was small, these individuals were all elite Paralympic sprinters using running-specific prostheses and were able to achieve maximum speeds

comparable to elite non-amputee sprinters. Our analyses were based on a small number of steps (~10-30 strides), which may limit the accuracy of our measurements (Arellano and Kram, 2011). However, our measurements were based on the most feasible approach given that we studied maximum sprinting speeds, which can only be maintained for a short period of time. Since our analysis was based on a similar number of steps for non-amputee sprinters and sprinters with trans-tibial amputations, it seems reasonable that this limitation would not affect our overall conclusions. Although we used the pelvis COM to represent the midline of the body, we would ideally measure M-L foot placement relative to the whole body COM. While there may be differences between the two, we feel that they are likely small and consistent in the M-L direction, thereby having little influence on our results.

Overall, our key findings demonstrate that 1) sprinters with trans-tibial amputations have greater challenges with maintaining lateral balance as compared to non-amputee sprinters and 2) for all runners, maintaining lateral balance is more challenging at faster running speeds up to maximum sprint speed. Furthermore, the apparent asymmetries in foot placement and its variability suggest that the use of running-specific prostheses results in a compensatory foot placement strategy for maintaining lateral balance.

5. ARM SWING DURING HUMAN WALKING: ACTIVE AND PASSIVE CONTRIBUTIONS TO A HYBRID SYSTEM

A. Abstract

Is arm swing during human walking actively driven by shoulder muscle forces, a passive pendulum-like response to body accelerations, or some combination? We quantified arm swing amplitudes and shoulder muscle activity while subjects walked at a range of step frequencies with: 1) their biological arms and 2) with free-swinging, anthropomorphic passive mechanical arms. Muscle activity measurements show that at preferred step frequencies, the backward swing of biological arms is caused by posterior deltoid muscle actions while the forward swing is driven by gravity and passive pendulum dynamics. Passive mechanical arm swing resembled the resonance behavior of a horizontally driven pendulum, reaching its largest amplitude as step frequency approached the biological arms natural frequency. However, the swinging amplitudes of passive mechanical arms were much less than the biological arms. Our findings demonstrate that arm swing during human walking is a hybrid system of active muscular actuation and passive pendulum dynamics.

B. Introduction

During the course of evolution, our ancestors transitioned from quadrupedal to bipedal locomotion. Although the arms were relieved from the functions of body weight support and propulsion, there remains a natural tendency for our arms to swing back and forth while walking. Classic and recent experiments have shown that the primary function of arm swing during bipedal walking is to improve balance by minimizing whole-body rotations (Collins et al., 2009; Elftman, 1938; Herr and Popovic, 2008; Hinrichs, 1990; Li et al., 2001). However, controversy still remains as to the underlying mechanism that produces arm swing motion during walking. Is arm swing actively driven by muscle forces, a passive pendulum response to the accelerations of the body, or some combination?

Scientific interest in arm swing dates back to the 1800's. Gerdy (Gerdy, 1829) and Weber and Weber (Weber and Weber, 1836) originally described human arm swing during walking as passive pendulums (noted in (Ballesteros et al., 1965)). This idea persisted for many years (Morton and Fuller, 1952) and was later endorsed by prominent biomechanists (Chapman and Ralston, 1964; Inman et al., 1981). For example, Chapman and Ralston (1964) observed that the metabolic cost of walking was similar when the arms were free to swing and when the arms were bound to the side of the body. Since arm swing did not appear to demand metabolic energy by the muscles, they concluded that “the arms behave as passive compound pendulums activated by movements at their point of attachment to the shoulder”.

In contrast to the idea that the arms act as passive pendulums, others have argued that arm swing is an active movement caused directly by the activation of the shoulder muscles. Based on a mechanical analysis of arm swing, Elftman (1938) posited that the shoulder muscles produce a large amount of torque, leading him to conclude that “the swinging of the arms is not

an example of pendulum action, but is brought about in large part by the action of muscles”. Later, Ballesteros et al. (1965) and Hogue (Hogue, 1969) measured the electromyographic (EMG) activity of various muscles of the arm and trunk and found that the posterior deltoid muscle at the shoulder plays the most important role in the backward swinging of the arm as well as decelerating the forward swing at the end of the flexion phase. In particular, Ballesteros et al. found that EMG activity of the shoulder flexors, such as the anterior deltoid, remains silent during the forward swing suggesting that passive forces (e.g. gravity) are responsible for the forward swinging motion of the arm. Furthermore, Ballesteros et al. found that the backward swinging motion of the arm was “abolished” in a patient with a paralyzed posterior deltoid muscle. These data indicate that swinging the arms during walking requires muscular actuation.

One would think that the EMG data would have resolved this issue long ago, but the controversy surrounding the underlying mechanism of human arm swing has resurfaced. Recently, Pontzer et al. (2009) proposed a passive arm swing hypothesis whereby arm swing is derived by the mechanical energy from the swinging legs, which is transferred through the pelvis and upward through the torso and shoulder girdle. In contrast to Ballesteros et al., they reported simultaneous activation of the anterior and posterior deltoid; leading them to surmise that these muscles act to stabilize the shoulder, not drive arm swing. They also found that restricting arm swing did not affect the metabolic cost of walking, an observation analogous to Chapman and Ralston (1964). These findings led Pontzer et al. to hypothesize that “the power for arm swing is ultimately derived from the swinging legs”.

In another recent study, Collins et al. (2009) used a three-dimensional passive dynamic walking computer model to show that purely passive arms could swing back and forth like biological arms. They also built passive mechanical arms made of wood or thick rope and

provided physical demonstrations of passive arm swinging during human walking. In contrast to Pontzer et al., Collins et al. found that restricting arm swing increased the metabolic cost of walking. As such, they proposed a “cost-benefit” hypothesis whereby shoulder muscles incur a direct metabolic cost to swing the arms but arm swing indirectly reduces the overall cost of walking by reducing the ground moment about the vertical axis. They further reasoned that the direct cost of swinging the arms is negligible since both their walking model and artificial arms could generate passive arm swinging dynamics. These results are very intriguing, however, no attempt was made to match the pendulum characteristics of the passive mechanical arms to those of biological arms. Comparing passive arms that are mechanically similar to biological arms could provide key insights into whether human arm swing could be accomplished by passive dynamics. Although the hypothesis proposed by Collins et al. is fundamentally different from that of Pontzer et al., an underlying theme in both is that arm swing can arise primarily from passive dynamics with little to no muscular effort to drive the arms.

Motivated by these recent findings, we sought to determine if biological arm swing during walking could arise from passive dynamics alone. In line with the physics of forced, lightly damped pendulums (Baker and Blackburn, 2005; Den Hartog, 1985), we reasoned that the arms could be made to swing in the presence of a cyclical horizontal force applied to the shoulder joint, the pivot point of attachment of the arm. We studied people walking 1) normally with their biological arms and 2) with passive mechanical arms that matched the pendulum characteristics of their biological arms (Figure 5.1). Subjects walked on a treadmill at a constant speed (1.25 m/s) across a range of slow and fast metronome enforced step frequencies. As our first test of the passive arm swing hypothesis, we measured anterior and posterior deltoid muscular activity and the amplitude of biological arm swing at 70, 80, 90, 100, 110, 120, and

130% of each subject's preferred step frequency (Figure 5.1A). If biological arm swing is the result of purely passive pendulum motion, then we would expect that the force generated against the ground during each step would accelerate the shoulder joint and excite the passive motion of the biological arms, causing the arms to swing without the need for shoulder muscular activity. Alternatively, if arm swing is not purely passive, then biological arm swing should require some degree of muscular activity to drive the arms.

As our second test of the passive arm swing hypothesis, we measured the swinging amplitude of anthropomorphic passive mechanical arms while subjects walked at the range of step frequencies noted above (Figure 5.1B). During these experiments, subjects crossed their biological arms across their chest so that the passive mechanical arms were free to swing. If biological arm swing is the result of purely passive pendulum motion, then we would expect that the force generated against the ground during each step would accelerate the mechanical pivot and excite the motion of the pendulum-like passive mechanical arms, causing the swing amplitudes across step frequencies to be similar to biological arms. Alternatively, if arm swing is not purely passive, then the passive mechanical arms should swing with amplitudes that are less than the swing amplitudes of biological arms.

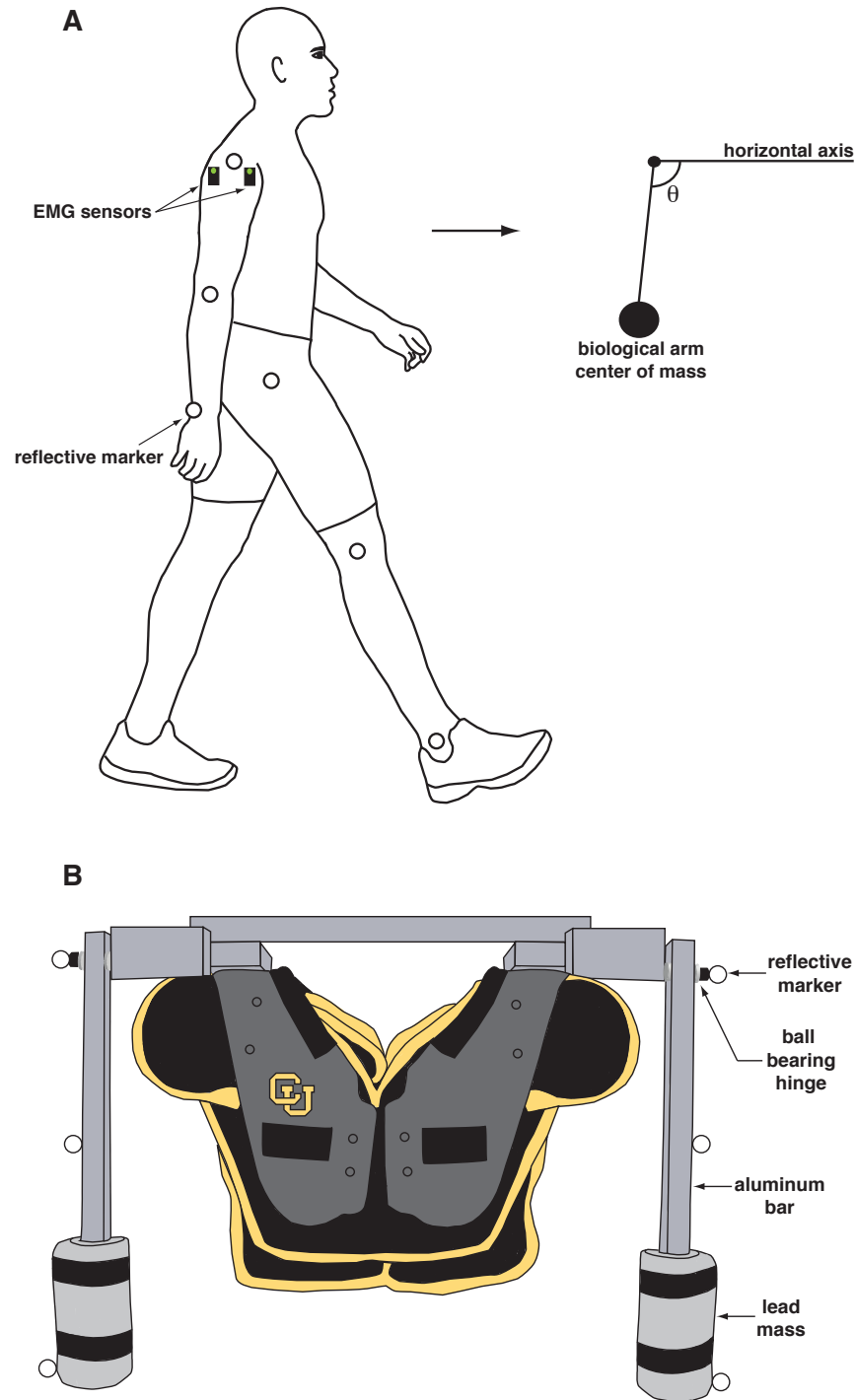


Figure 5.1 (A) Reflective marker and EMG sensor placement for measurements of biological arm swing, leg swing, and deltoid muscle activity. From segment lengths and masses, we calculated the position of the biological arm's center of mass relative to the shoulder and computed the angle with respect to the horizontal axis in the sagittal plane. (B) During passive mechanical arm swinging, subjects wore our custom-built device, consisting of a set of football shoulder pads linked to aluminum bars that hung vertically from their pivot point. Reflective markers were placed at the locations of the ball bearing hinge, middle of the bar, and at the end of the lead mass. Damping of the passive mechanical arms was very small (~ 1.05 Ns/m) and equivalent to only 3.0% of the critical damping (~ 35.59 Ns/m) that would cause them not to swing (described in Appendix C).

C. Methods

Ten male subjects participated in this study (age = 25.1 ± 4.5 years, mass = 77.0 ± 7.1 kg, height = 1.83 ± 0.1 cm; mean $\pm$ sd). Prior to the experimental data collection, all subjects read and signed an informed consent document that was approved by the University of Colorado's Institutional Review Board. Subjects wore their own shoes, were experienced with treadmill walking, healthy, and injury-free.

Experimental Design

The experiment consisted of two randomized sessions that included 1) walking with their biological arms and 2) walking with pendulum-like passive mechanical arms. For each session, subjects walked on a treadmill for randomized conditions of 70, 80, 90, 100, 110, 120, and 130% of their normal step frequency while speed was held fixed at 1.25 m/s. At the beginning of the experiment, we asked subjects to walk on the treadmill at 1.25 m/s and measured their stride frequency by determining the time taken to complete 10 strides using a stopwatch. We divided stride frequency by 2 to yield step frequency and then enforced step frequency during each condition using a metronome (Boss Corp., DB-66). To begin each condition, we set the frequency of the metronome to the enforced step frequency and allowed the subject adequate time to match their step frequency to the frequency of the metronome. On average, subjects preferred a step frequency of 1.74 ± 0.1 Hz (mean $\pm$ sd).

Biological Arm Swing

We placed reflective markers on the following landmarks of the left and right side of the body (Motion Analysis Corporation, Santa Rosa, CA): (1) acromion process; (2) lateral aspect of

the elbow joint; (3) styloid process of the ulna; (4) greater trochanter; (5) lateral aspect of the knee joint; (6) lateral malleolus. The surface EMG of the left and right shoulder muscles (anterior and posterior deltoid) were recorded with wireless EMG sensors (Delsys Inc., Trigno Wireless System, Boston, MA) with an amplifier gain of 300, a Common Mode Rejection Ratio > 80 dB, and a signal bandwidth of 20-450 Hz. Each EMG sensor consists of four contact sites (5 x 1 mm) made from 99.9% silver with an inter-electrode distance of 10 mm. Prior to sensor placement, we prepared the site by abrading the surface of the skin (shaving followed by rubbing with fine grade sandpaper) and then wiping with isopropyl alcohol. Finally, we placed each EMG sensor at the center of the muscle belly with the orientation parallel to the muscle fibers following the recommendations by Cram and Kasman (Cram and Kasman, 1998). When walking with biological arms, we waited for each subject to achieve steady-state, then simultaneously recorded the three-dimensional motions of the reflective markers (100 Hz) and surface EMG (1000 Hz) for 20 seconds.

Passive Arm Swing Device

For the passive mechanical arm swing trials, we placed reflective markers at the hinge joint, middle and end of the aluminum bar (Figure 5.1B). Reflective markers placed on the leg were similar to that noted above. When walking with our device, subjects were instructed to cross their biological arms in front of their chest while the mechanical arms swung passively. For each trial, we again waited for each subject to achieve steady-state, then recorded the three-dimensional motions of the reflective markers (100 Hz) for 20 seconds.

Kinematic and EMG Analysis

For each trial, we filtered the reflective marker position data using a 9th order, zero-lag low-pass Butterworth filter with a cutoff frequency of 6 Hz. For biological arm swing, we computed the sagittal plane coordinates of the arm center of mass (COM) using reflective marker position data and published anthropometric data tables (Enoka, 2002; Winter, 1990). For passive arm swing, we computed sagittal plane coordinates of the mechanical arms using the position of the reflective marker placed at the hinge joint and at the end of the aluminum bar. For both conditions, arm swing was measured from an angle formed between a horizontal reference axis and a vector defining the orientation of the arm. We defined one arm swing cycle as the maximum angle swept by the arm's center of mass, i.e. successive instants of when the arm reached its maximum swing in the backwards direction.

To process the surface EMG signals, we subtracted the baseline offset from the raw EMG so that the mean raw EMG signal was equal to zero. Following the procedure of Yang and Winter (Yang and Winter, 1984), the EMG signal was full-wave rectified followed by a 2nd order, zero-lag low pass Butterworth filter with a cutoff of 3 Hz to obtain a linear envelope. We time normalized each individual cycle of angle and EMG data to 1001 data points, representing the arm swing cycle from 0-100%. For each subject, we then averaged arm swing amplitude (in degrees) and EMG amplitude (in Volts) over 10 arm swing cycles that were free from movement artifact. In the case of passive mechanical arm swing, we time normalized the angle data only. To reduce inter-subject variability, we normalized the average EMG amplitude at the preferred step frequency by setting this value to 1.0 for each subject.

Statistical Analysis

As a preliminary analysis for biological and passive mechanical arm swing, we used a general linear model (GLM) repeated-measures analyses with two within subject fixed factors (step frequency and arm, i.e. left or right) to determine whether there were any significant differences in arm swing amplitude between the right and left arm across step frequency conditions. Our preliminary analyses revealed a significant main effect for step frequency (biological arms: $p < 0.001$; passive mechanical arms: $p < 0.001$), no significant main effect for arm (biological arms: $p = 0.637$; passive mechanical arms: $p = 0.852$), and no significant interaction effect between step frequency and arm (biological arms: $p = 0.595$; passive mechanical arms: $p = 0.129$). These results indicate that the arm swinging amplitude across step frequency conditions was similar between the left and right arm. Thus, for both the biological and passive mechanical arm swing trials, we averaged the arm swing amplitude across the left and right arm.

We also used a GLM repeated-measures analyses with two within subject fixed factors (step frequency and arm, i.e. left or right) to determine whether there were any significant differences in normalized EMG amplitude during biological arm swing. Our preliminary analysis revealed a significant main effect for step frequency (anterior deltoid: $p < 0.001$; posterior deltoid: $p < 0.001$), no significant difference for arm (anterior deltoid: $p = 0.283$; posterior deltoid: $p = 0.130$), and no significant interaction effect between step frequency and arm (anterior deltoid: $p = 0.332$; posterior deltoid: $p = 0.389$). Thus, we averaged the normalized EMG data across the left and right arm.

As a final analysis, we used a GLM repeated-measures analyses with step frequency as a within subjects fixed factor to evaluate a main effect for biological and passive mechanical arm

swinging amplitude. We also used a GLM repeated measures analyses with step frequency as a within subjects fixed factor to evaluate a main effect for normalized EMG amplitude. For all statistical tests, significance was set at an α level = 0.05 (SPSS Inc., Chicago, IL). If Mauchly's test of sphericity was violated at the 0.05 level, we adjusted the degrees of freedom (Huynh-Feldt) to test for significance. If significant main effects were detected for step frequency, we followed the GLM repeated-measures analysis with *a priori* comparisons between the control (100% of preferred step frequency) and the other step frequency conditions (70, 80, 90, 110, 120, and 130% of preferred step frequency) using the simple contrast method in SPSS. All values are reported as mean $\pm$ SEM unless noted otherwise.

D. Results

Step Frequency Main Effect for Arm Swing Amplitude

Our analysis revealed a significant main effect for both biological ($p < 0.001$) and passive mechanical arm swing ($p < 0.001$). When walking at the preferred step frequency, the amplitudes of biological and passive mechanical arm swing were 27 and 12 degrees, respectively (Table 5.1). Compared to the preferred step frequency (control), the amplitude of biological arm swing significantly increased by 11%, 30%, and 62% when walking at slower step frequencies of 90%, 80%, and 70%, respectively. In contrast, the amplitude of biological arm swing significantly decreased by 14%, 25%, and 30% when walking at faster step frequencies of 110%, 120%, and 130%, respectively. Compared to the preferred step frequency (control), the amplitude of passive mechanical arm swing significantly increased by 54% and 156% when walking at slower step frequencies of 80% and 70%, respectively. In contrast, the amplitude of passive mechanical arm

swing was similar when walking at a slower step frequency of 90% or at faster step frequencies of 110%, 120%, or 130% (all p -values are provided in Table 5.1).

Step Frequency Main Effect for normalized EMG amplitude

Our analysis revealed significant main effects for both the posterior deltoid ($p < 0.001$) and anterior deltoid ($p < 0.001$). Compared to the preferred step frequency (control), the normalized EMG amplitude of the posterior deltoid significantly increased by 42%, 84%, and 157% when walking at slower step frequencies of 90%, 80%, and 70%, respectively. Similarly, the normalized EMG amplitude of the posterior deltoid significantly increased by 40%, 63%, and 93% when walking at faster step frequencies of 110%, 120%, and 130%. Compared to the preferred step frequency (control), the normalized EMG amplitude of the anterior deltoid significantly increased by 28%, 125%, and 350% when walking at slower step frequencies of 90%, 80%, and 70%, respectively. In contrast, the normalized EMG amplitude of the anterior deltoid was similar when walking at faster step frequencies of 110%, 120% and 130% (all p -values are provided in Table 5.2).

Table 5.1 Amplitude of biological and passive mechanical arm swing at each step frequency condition ($n = 10$; mean $\pm$ SEM).

Step Frequency (%)	Biological Arms (deg.)	Passive Mechanical Arms (deg.)
70	43.2 $\pm$ 4.8 ($p = 0.002$)	30.3 $\pm$ 3.4 ($p < 0.001$)
80	34.5 $\pm$ 3.4 ($p = 0.021$)	18.2 $\pm$ 1.9 ($p = 0.016$)
90	29.7 $\pm$ 2.0 ($p = 0.007$)	13.3 $\pm$ 1.4 ($p = 0.441$)
100 (control)	26.6 $\pm$ 2.2	11.8 $\pm$ 2.2
110	23.0 $\pm$ 2.2 ($p = 0.014$)	9.4 $\pm$ 1.0 ($p = 0.148$)
120	20.0 $\pm$ 2.0 ($p = 0.002$)	9.3 $\pm$ 0.8 ($p = 0.215$)
130	19.0 $\pm$ 2.0 ($p = 0.003$)	9.9 $\pm$ 1.4 ($p = 0.354$)
simple contrast statistics		
For each step frequency comparison that yielded statistical significance against the control, p values < 0.05 are denoted in bold.		

Table 5.2 Normalized EMG amplitude of the posterior and anterior deltoid while swinging the biological arms at each step frequency condition ($n = 10$; mean $\pm$ SEM).

Step Frequency (%)	Posterior Deltoid (V/V)	Anterior Deltoid (V/V)
70	2.57 $\pm$ 0.22 ($p < 0.001$)	3.50 $\pm$ 0.35 ($p < 0.001$)
80	1.84 $\pm$ 0.11 ($p < 0.001$)	2.25 $\pm$ 0.21 ($p < 0.001$)
90	1.42 $\pm$ 0.08 ($p < 0.001$)	1.28 $\pm$ 0.09 ($p = 0.016$)
100 (control)	1.00	1.00
110	1.40 $\pm$ 0.13 ($p < 0.016$)	0.94 $\pm$ 0.04 ($p = 0.114$)
120	1.63 $\pm$ 0.13 ($p < 0.001$)	1.01 $\pm$ 0.04 ($p = 0.759$)
130	1.93 $\pm$ 0.19 ($p < 0.001$)	1.03 $\pm$ 0.05 ($p = 0.603$)
simple contrast statistics		
For each step frequency comparison that yielded statistical significance against the control, p values < 0.05 are denoted in bold.		

E. Discussion

While walking at the preferred step frequency (100%), biological arm swing clearly required muscular actuation for retraction (i.e. backwards motion), which was initiated by the posterior deltoid, the primary extensor muscle at the shoulder (Figure 5.2). Similar to the findings of Ballesteros et al. (1965), the posterior deltoid showed a burst of EMG activity lasting from just before until just after the biological arm reached its maximum forward position during arm swing, acting to slow down the forward motion and initiate retraction of the arm. The subsequent swinging of the biological arm in the forward direction occurred in the absence of any significant anterior deltoid EMG activity, the primary flexor muscle at the shoulder. Every subject ($n = 10$) showed this general behavior in the EMG activity and amplitude of biological arm swing. Our EMG data provide evidence that when walking normally, biological arm swing is not a purely passive process but rather resembles a hybrid system whereby the backward swing is driven by active retraction by the posterior deltoid muscle and the forward swing is driven by passive pendular motion.

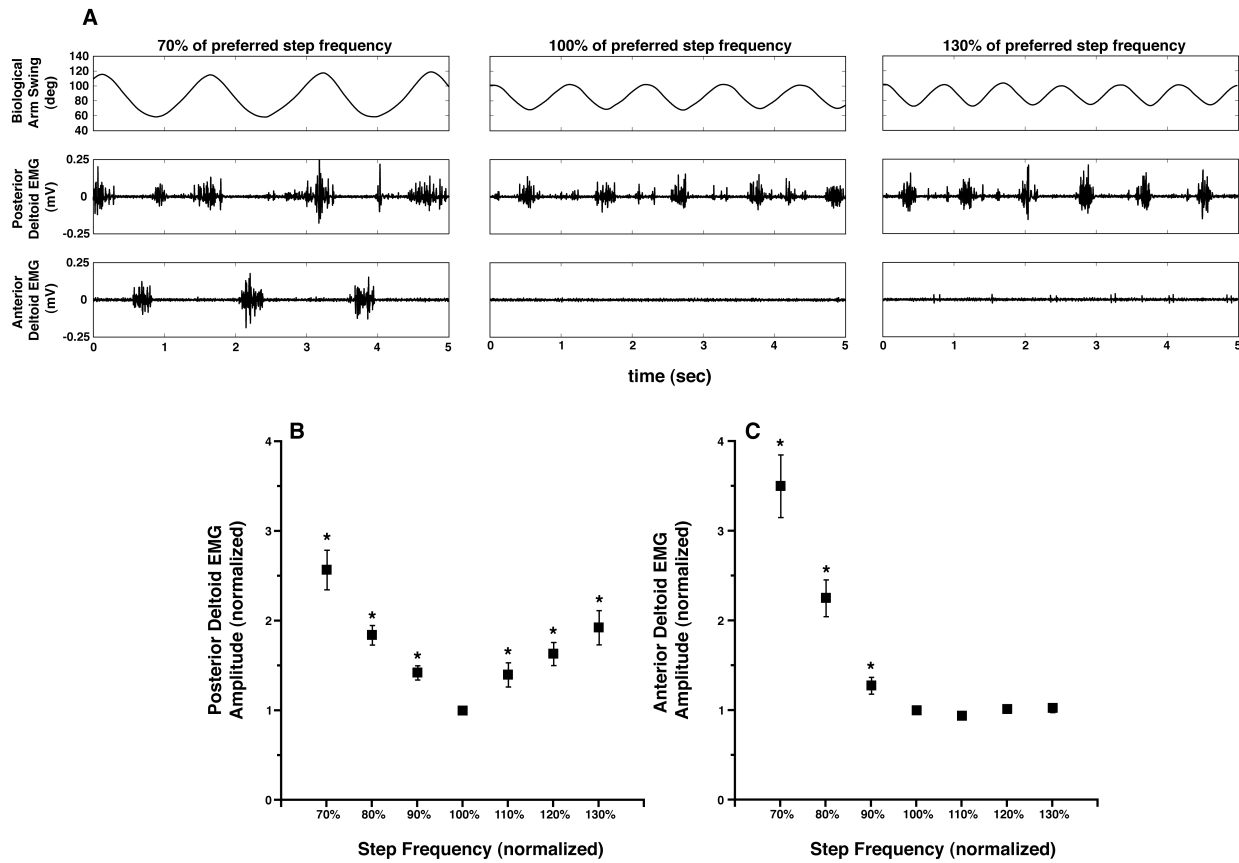


Figure 5.2 (A) Biological arm swing and raw EMG activity for a subject walking at 70%, 100%, and 130% of their preferred step frequency (data for right arm only). Across step frequencies, the average EMG amplitude for the (B) posterior and (C) anterior deltoid resembled V-shaped and L-shaped trends, respectively ($n = 10$). Subjects significantly increased posterior deltoid activity when walking at slower and faster step frequencies (when compared to 100% preferred step frequency, * denotes $p < 0.05$). In contrast, subjects only increased anterior deltoid activity at slower step frequencies (when compared to 100%, * denotes $p < 0.05$). The anterior deltoid remained nearly silent at preferred and faster step frequencies. Values are means $\pm$ SEM.

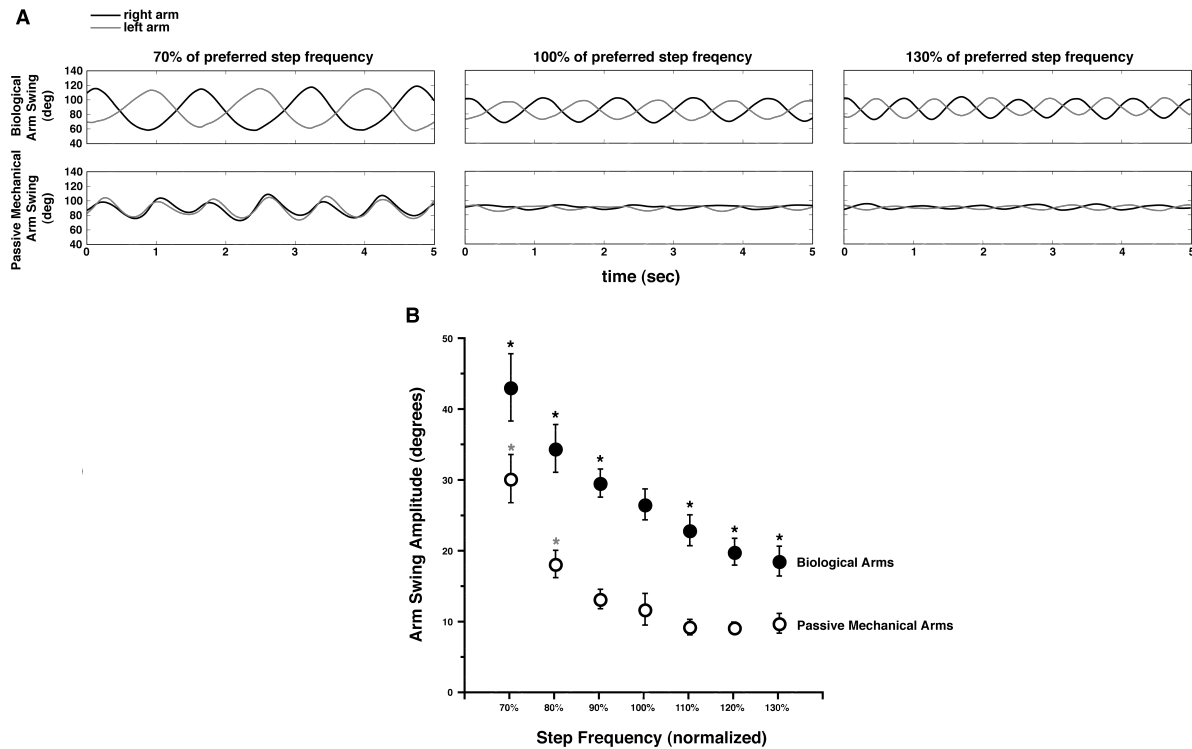


Figure 5.3 (A) Amplitude of biological and passive mechanical arm swing for a representative subject walking at 70%, 100%, and 130% of preferred step frequency. (B) Average swing amplitude of biological (filled) and passive mechanical arms (open) while walking across normalized step frequencies (mean $\pm$ SEM; $n = 10$). At faster step frequencies (110-130%), the amplitude of biological arm swing decreased by 30% while passive mechanical arm swing stayed the same. At slower step frequencies (70%-90%), both biological and passive mechanical arm swinging amplitude increased steadily. The amplitude of passive mechanical arm swinging was clearly much less than that of biological arm swinging. Further, while biological right and left arm swinging was always found to be out-of-phase, passive right and left mechanical arm swinging exhibited a range of in-phase to out-of-phase oscillations. Note, when compared to 100% preferred step frequency, the black * denotes $p < 0.05$ for biological arm swing and the gray * denotes $p < 0.05$ for passive mechanical arm swing.

We further explored the passive arm swing hypothesis using the established theory of forced mechanical vibrations (Baker and Blackburn, 2005; Den Hartog, 1985). Every simple pendulum has a natural frequency of oscillation ($\omega_{natural} = \sqrt{g/l}$) that depends on gravity ($g = 9.81 \text{ m/s}^2$) and the length of the pendulum ($l = \text{length}$). A pendulum can be induced to swing back and forth when the pivot is subjected to a driving force, due to a sinusoidal, horizontal displacement (Franz-Josef, 1998). This phenomenon is illustrated by the resonance diagram in Figure 5.4A and demonstrates three important effects of the forcing frequency, ω_{force} . At point a, when ω_{force} approaches zero, the forcing frequency applied to the pivot becomes extremely slow, thus the amplitude of the swinging pendulum approaches zero. At point c, at very high forcing frequencies ($\omega_{force}/\omega_{natural} \gg 1$), the forcing frequency oscillates so fast that the swinging pendulum cannot follow, thus the swinging amplitude is very small. Point b is the most interesting and represents resonance whereby the forcing frequency coincides with the natural frequency of the pendulum, causing it to swing at infinitely large amplitudes. Using this theoretical framework, we treated step frequency as the forcing frequency and compared changes in the amplitude of biological and passive arm swing to the amplitude predicted from a horizontally driven pendulum. Since the legs generate force against the ground during each step and assuming the force can be transferred to the upper body, we reasoned that walking at a fixed speed (1.25 m/s) but at different step frequencies would change the forcing frequency applied to the pivot point and therefore excite the swinging amplitude of the pendulum-like passive mechanical arms.

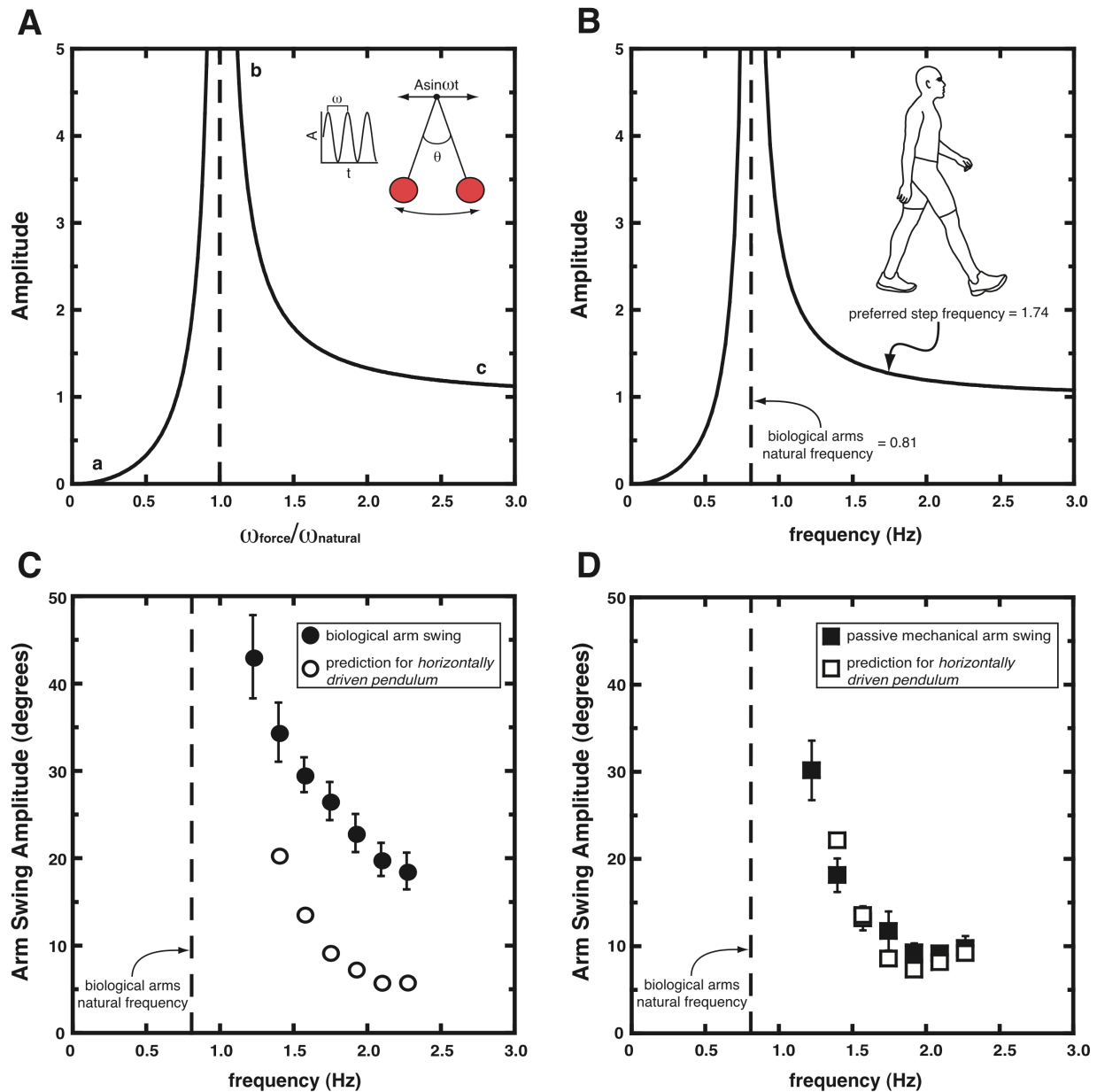


Figure 5.4 (A) Resonance diagram for a horizontally driven pendulum subjected to different frequencies. (B) Resonance diagram comparing the biological arms natural frequency (0.81 Hz) to the preferred step frequency adopted during walking (1.74 Hz). (C) Biological arm swing amplitude measured experimentally (closed circles) and predicted (open circles) for a horizontally driven anthropomorphic pendulum subjected to different step frequencies during walking (described in Appendix D). Across step frequencies, the predicted arm swing amplitude underestimates the swinging amplitudes of the biological arms. (D) Passive mechanical arm swing amplitude measured experimentally (closed squares) and predicted (open squares) for a horizontally driven anthropomorphic pendulum subjected to different step frequencies during walking. Across step frequencies, the arm swing amplitudes predicted for a horizontally driven pendulum matches well with the swinging amplitudes of the passive mechanical arms. At the slowest step frequency (1.22 Hz = 70%), passive mechanical arm swinging reached the largest amplitude when approaching resonance. Note, the linearized approximation for a horizontally driven pendulum is not valid for angles much greater than 20 degrees, thus we do not include the predicted amplitude for the slowest step frequency.

While walking at the preferred step frequency (100%), biological arm swing swept an average angle of 27 degrees. In contrast, passive mechanical arm swing at the preferred step frequency swept an angle of only 12 degrees, less than half the amplitude of biological arm swing. The amplitude of biological arm swing increased at slower step frequencies but decreased at fast step frequencies (Figure 5.3A,B). At the slowest step frequency (70%), biological arm swing reached an amplitude of 43 degrees, 62% greater than the preferred step frequency. At the fastest step frequency (130%), the amplitude only reached 19 degrees, 30% less than the preferred step frequency. The changes in the amplitude of biological arm swing were accompanied by changes in the muscular activation of the posterior and anterior deltoid (Figure 5.2B,C). At the slowest step frequency, the EMG amplitude of the anterior and posterior deltoid muscle increased by 160% and 250%, respectively. At the fastest step frequency (130%), the EMG amplitude of the posterior deltoid increased by 92% while the anterior deltoid remained silent. When compared to the preferred step frequency, no significant changes in anterior deltoid EMG amplitude were detected at 110, 120, or 130%, indicating that the forward swinging of the biological arm was achieved passively.

In general, the amplitude of purely passive mechanical arm swing was much less than biological arm swing across the range of step frequencies (Figure 5.3B). At the slowest (70%) and fastest (130%) step frequency, biological arm swing amplitude exceeded that of passive arm swing by 13 and 9 degrees, respectively. Nonetheless, the amplitude of passive mechanical arm swing was excited by steady decreases in step frequency, similar to what occurs when an external force excites a damped pendulum when applied at variable frequencies. In contrast, walking at faster step frequencies did not change the amplitude of passive mechanical arm swing, indicating that step frequencies greater than preferred did not further excite the amplitude

of the passive arms. Our data demonstrate that amplitudes of biological arm swing cannot be accomplished by purely passive dynamics; shoulder muscular actuation is also necessary to drive the arms.

As demonstrated in Figure 5.4B, we find that the preferred step frequency during walking (1.74 Hz) was much greater than the biological arms natural frequency (0.81 Hz). In short, the resonance diagram suggests that the preferred step frequency is too fast ($\omega_{force}/\omega_{natural} \gg 1$) to generate large amplitudes of passive mechanical arm swing. The preferred step frequency is 2.15 times greater than the biological arm's natural frequency, suggesting that little mechanical energy can be harnessed to drive biological arm swing. Indeed, we find that the data predicted by a horizontally driven pendulum matches well with the amplitude of the passive mechanical arms but not the biological arms (Figure 5.4C,D). Note that across the step frequencies studied here, the changes in the amplitude of biological arm swing were much greater than the amplitude predicted from a horizontally driven pendulum. Overall, our data are consistent with the established theory of forced, lightly damped pendulums and provides a reasonable explanation as to why passive mechanical arm swing does not match the amplitude of biological arm swing. In conclusion, we show that human arm swing is not primarily passive but can be characterized as a hybrid system of active muscular actuation and passive pendulum dynamics.

APPENDIX C

Matching the Inertial Characteristics of Passive Mechanical Arms to Biological Arms

The mass, inertia, and resonant frequency of the passive mechanical arms were matched to the mass, inertia, and resonant frequency of each subject's biological arm. Based on the subjects body mass and height, we estimated the total mass, length, and inertia of the arm using published anthropometric data (Enoka, 2002). From these data, we treated the biological arm as a simple pendulum by computing the position of the biological arms center of mass from the shoulder joint. Then, we set out to match the mechanical characteristics of the passive arms to that of the biological arms by manipulating the inertia and distance of the lead mass from its proximal point of attachment at the hinge joint. By accounting for the inertial characteristics of the aluminum bar, we calculated the values required to match the inertial characteristics of the passive mechanical arm to that of the biological arm (Eq. 1):

$$\begin{aligned}
 I_{lead\ mass} + I_{aluminum\ bar} &= I_{passive\ mechanical\ arm} \\
 I_{passive\ mechanical\ arm} &= I_{biological\ arm}
 \end{aligned}
 \tag{Equation 1}$$

To ensure that the inertia of passive mechanical and biological arms were similar, we measured the natural resonant frequency of the right and left passive mechanical arms by measuring the time for them to swing for 10 cycles. We found the time for the left and right arm to swing for 10 cycles to be similar, thus we used the average time to calculate the natural resonant frequency. As seen in Table 5.3, there was no significant difference between the natural resonant frequency of the passive mechanical arms and biological arms.

Table 5.3 Natural resonant frequency of biological and passive mechanical arms (mean $\pm$ SD).

Subject	Biological Arms (Hz)	Passive Mechanical Arms (Hz)
1	0.82	0.82
2	0.79	0.79
3	0.80	0.81
4	0.82	0.82
5	0.81	0.81
6	0.82	0.83
7	0.80	0.81
8	0.82	0.82
9	0.82	0.82
10	0.82	0.82
Average	0.81 $\pm$ 0.01	0.81 $\pm$ 0.01
An independent t-test detected no significant difference in the natural resonant frequency of the biological and passive mechanical arms ($p = 0.522$).		

Damping of Passive Mechanical Arms

As described in our previously published paper (see Appendix B), we used the logarithmic decrement method to calculate the damping characteristics of our passive mechanical arms. We displaced the passive mechanical arm from its vertical equilibrium position and measured the decaying oscillation of the arm swing angle using the reflective markers placed on the arm and data recorded from our motion capture system. Since the damping was similar for the left and right arm, we report the average damping (~ 1.05 Ns/m) and critical damping (~ 35.59 Ns/m) values in the main text. The average mass of the passive mechanical arms was 3.63 kg, yielding a damping constant equal to 0.29 sec^{-1} .

Potential Effects of Initial Conditions

We did not study the effects of initial conditions on the amplitude of passive mechanical arm swing. Collins et al. (2009) noted that their passive arms (made of light wood or thick rope) could swing out-of-phase or in-phase when starting with the proper initial conditions. One might reason that starting at different initial conditions during the various walking trials could also change the swinging amplitude of the passive mechanical arms that were observed in this study. However, this reasoning is not supported by the periodic oscillations predicted by the equation of motion for a forced, lightly damped pendulum (Baker and Blackburn, 2005). Due to the combination of energy dissipation due to friction and external cyclical forcing, the general solution to the differential equation contains a “transient” and a “steady state” solution [see Eq. 3.19 through 3.22 in (Baker and Blackburn, 2005)]. The transient solution is the short-term periodic behavior that decays quickly due to damping and only depends on initial conditions. The steady state solution is independent of initial conditions and predicts the long-term periodic

behavior. Ultimately, the amplitude of a forced, lightly damped pendulum depends on the magnitude of damping, which we found to be very small in our passive mechanical arms. In addition, the predictions of the steady state solution are in line with the observations of Jackson et al. (Jackson et al., 1978) who developed a mathematical model of arm swing during human walking. They found that changing initial conditions only altered the transient motion, which died away after the 2nd or 3rd cycle. After the initial transients died away, the arm swinging motion predicted by the model settled to a “stable mode of oscillation”. Although we found different modes of oscillation (e.g. in-phase and out-of-phase) during passive mechanical arm swing (Figure 5.3A), it is unlikely that different initial conditions would have a major effect on its amplitude. Indeed, we found that the swinging amplitudes of the right and left passive mechanical arm were not significantly different (see *Statistical Analysis*). Even if the left and right passive mechanical arms started with different initial conditions, our data show that they reached similar arm swing amplitudes.

We must note that the linearized approximation of a forced, lightly damped pendulum only holds for relatively small angles where $\sin \theta \approx \theta$, perhaps up to 20 degrees. Thus, more complicated models would be required to predict the periodic behavior of a force, lightly damped pendulum when the linearized approximation is no longer valid (Baker and Blackburn, 2005). The case where the linearized approximation does not hold in this study is when walking at 70% of preferred step frequency, where the amplitude of the passive mechanical arms reached ~30 degrees. However, during this condition, we found that passive mechanical arm swinging tended to settle down to a constant amplitude (Figure 5.2).

APPENDIX D

Resonance Curves and Arm Swing Amplitude Predictions

Following the linearized equation of motion for a horizontally driven pendulum at its pivot point (Franz-Josef, 1998), the relationship between the swinging amplitude, the amplitude of the driving force, and forcing frequency is as follows:

$$\theta(A, \omega_{force}) = \frac{A}{\sqrt{(\omega_n^2 - \omega_{force}^2)^2 + (c\omega_{force})^2}} \times \left(\frac{180}{\pi} \right) \quad \text{Equation (2)}$$

Here, ω_n^2 is the pendulum's natural frequency and c is the damping constant expressed in units of sec^{-1} . The term A is as follows:

$$A = \omega_{force}^2 (x/l),$$

ω_{force}^2 = forcing frequency ($\text{radians}^2/\text{second}^2$)
 x = horizontal amplitude of driving force (meters)
 l = length of pendulum (meters)

To calculate the value of A across step frequencies, we computed x — the horizontal amplitude of the driving force — from the horizontal motion (1) of the reflective markers placed at the shoulder joint of the biological arms and from the horizontal motion (2) of the reflective markers placed at the pivot point of the passive mechanical arms. For each subject and step frequency condition, we calculated the average amplitude of the horizontal motion over 10 arm swing cycles. For both the biological and passive mechanical arm swing trials, we averaged the horizontal amplitude of the reflective marker across the left and right arm (Figure 5.4).

Table 5.4 Amplitude of the horizontal motion of the reflective marker placed the shoulder joint and pivot point of the biological and passive mechanical arms during each step frequency condition (mean $\pm$ SEM).

Step Frequency (Hz)	Biological Arms (m)	Passive Mechanical Arms (m)
1.22	0.149 $\pm$ 0.011	0.132 $\pm$ 0.011
1.39	0.089 $\pm$ 0.008	0.078 $\pm$ 0.007
1.57	0.066 $\pm$ 0.005	0.053 $\pm$ 0.006
1.74 (preferred)	0.049 $\pm$ 0.004	0.036 $\pm$ 0.002
1.91	0.040 $\pm$ 0.003	0.032 $\pm$ 0.002
2.09	0.034 $\pm$ 0.002	0.037 $\pm$ 0.002
2.26	0.034 $\pm$ 0.003	0.044 $\pm$ 0.006

The average pendulum length for the biological and passive mechanical arms was 0.376 m and 0.304 m, respectively. We measured the pendulum length as the distance from the reflective marker positioned at either the shoulder joint or the pivot point to the position of the arm's center of mass. Using these values, we computed the arm swing amplitude predicted from a horizontally driven pendulum at its pivot point using Equation 2 (Table 5.5). As seen in Figure 5.4C, these predicted values are compared to the swinging amplitudes of the biological and passive mechanical arms that were quantified experimentally.

Table 5.5 Amplitude of biological and passive mechanical arm swinging as predicted from a horizontally driven pendulum subjected to variable step frequencies during walking.

Step Frequency (Hz)	Biological Arms (deg.)	Passive Mechanical Arms (deg.)
1.22	-	-
1.39	20.48	22.24
1.57	13.72	13.65
1.74 (preferred)	9.53	8.67
1.91	7.42	7.38
2.09	6.10	8.22
2.26	5.94	9.31

Note, the linearized approximation for a horizontally driven pendulum is not valid for angles much greater than 20 degrees, thus we do not include the predicted amplitude for the slowest step frequency (1.22 Hz). For both cases, we use a damping constant of 0.29 sec^{-1} as calculated for the passive mechanical arms (see *Damping of Mechanical Arms*). The damping ratio of the biological arms cannot be feasibly calculated, thus we resorted to using a damping constant of 0.29 sec^{-1} as a conservative estimate.

6. SUMMARY

To gain insight into my general hypothesis that humans not only minimize energetic cost but also optimize for balance during locomotion, I studied the role of step width and arm swing as the primary balance control mechanisms during running. I also studied human walking to address the long-standing controversy as to the underlying mechanism(s) that drive arm swing during walking. Below, I provide a short summary of each study as well as highlight my most significant findings.

1. The effects of step width and arm swing on energetic cost and lateral balance during running

Insights from human walking experiments suggested that step width and arm swing are the primary balance control mechanisms for maintaining lateral balance. Thus, I wanted to determine if both step width and arm swing not only help minimize energetic cost, but also improve lateral balance during human *running*. I found that when humans ran with a step width other than their preferred narrow step width, both metabolic power demand and step width variability increased, indicating a decrease in lateral balance. Furthermore, when running without arm swing, both net metabolic power demand and step width variability increased as compared to running with arm swing. My main conclusion from this study is that humans prefer to run with a narrow step width and swing their arms so as to minimize energetic cost and improve lateral balance.

2. The energetic cost of maintaining lateral balance during human running

Based on the insights from my first study, I wanted to establish a stronger link between energetic cost and lateral balance. I reasoned that if perturbing lateral balance increases the energetic cost of running (i.e. by asking human subjects to run with relatively wide step widths

and without arm swing), then I should also be able to aid lateral balance by stabilizing subjects about the waist, thus reducing the energetic cost of maintaining lateral balance. In addition, I reasoned that the energetic cost of maintaining must be relatively small since humans prefer to run with a narrow step width. Thus, I applied external lateral stabilization (LS) while humans ran with and without arm swing and quantified changes in energetic cost and step width variability. When provided with external LS, both net metabolic power demand (by ~2.0%) and step width variability (by ~12.0%) decreased to a similar extent when running with or without arm swing. While eliminating arm swing increased net metabolic power demand by ~8%, external LS did not decrease net metabolic power to a further extent when running without arm swing. This suggests that the primary role of arm swing is not to assist with lateral balance. From these data, I infer that the 2.0% reduction in energetic cost with external LS reflects the net energetic cost of maintaining lateral balance. In addition, I conclude that humans use step width adjustments as the primary mechanism to maintain lateral balance during running.

3. Lateral balance and foot placement across running speeds: A comparison of sprinters with and without trans-tibial amputations

During the time of my dissertation, there was a growing controversy as to whether the use of running-specific prostheses provided an advantage to competitive sprinters with unilateral and bilateral trans-tibial amputations. Several studies had compared the mechanics of sprint-running between non-amputee sprinters and sprinters with unilateral and bilateral trans-tibial amputations, however, anecdotal evidence suggested that sprinters using running-specific prostheses find it very difficult to maintain “balance”. I reasoned that the use of running-specific prostheses do not optimally facilitate lateral balance because they are designed to mimic the sagittal plane, spring-like behavior of biological legs.

Therefore, I wanted to determine how the use of running-specific prosthesis affects lateral balance. In addition, I also investigated how increases in running speed up to maximum sprint speed affect lateral balance. I compared step width and its variability between sprinters with trans-tibial amputations and non-amputee sprinters. To study potential asymmetries in lateral balance, I also compared M-L foot placement relative to the midline of the body and its variability between the legs of the same individual. Across the range of running speeds, sprinters with unilateral trans-tibial amputations ran with a 40% greater step width and 11% step width variability as compared to non-amputee sprinters. Although the sprinter with bilateral amputations ran with a step width similar to sprinters with unilateral amputations, he exhibited 50% greater step width variability. When comparing individual legs, M-L foot placement and its variability were similar between the right leg and left leg of non-amputee sprinters; however, the affected leg (AL) of sprinters with unilateral trans-tibial amputations exhibited 18% greater M-L foot placement variability as compared to the unaffected leg (UL). Both ALs of the sprinter with bilateral trans-tibial amputations exhibited 125% greater M-L foot placement variability as compared to the AL of sprinters with unilateral trans-tibial amputations. Overall, I observed that 1) sprinters with trans-tibial amputations have greater challenges with maintaining lateral balance and 2) maintaining lateral balance is more challenging at faster running speeds.

4. Arm swing during human walking: Active and passive contributions to a hybrid-system

This study was motivated by a long-standing debate about the underlying mechanism(s) that produces arm swing motion during human walking. I addressed the following question: Is arm swing during human walking actively driven by shoulder muscle forces, a passive pendulum-like response to body accelerations, or some combination? I quantified arm swing amplitudes and shoulder muscle activity while subjects walked at fixed speed of 1.25 m/s, but at

a range of step frequencies with: 1) their biological arms and 2) with free-swinging, anthropomorphic passive mechanical arms. The passive mechanical arms were matched to the mechanical characteristics of each subject's biological arms. While walking at their preferred step frequency, muscle activity measurements demonstrate the backward swing of biological arms is caused by posterior deltoid muscle actions while the forward swing is driven by gravity and passive pendulum dynamics. Walking at step frequencies slower or faster than preferred required greater actuation from the posterior and anterior deltoid muscles, which coincided with greater biological arm swinging amplitudes. Walking at step frequencies slower than the preferred step frequency excited the swinging amplitude of the passive mechanical arms, resembling the resonance behavior of a horizontally driven pendulum, reaching its largest amplitude as step frequency approached the biological arms natural frequency. However, the swinging amplitudes of passive mechanical arms were much less than the biological arms. From these findings, I conclude that arm swing during human walking comprises a hybrid system of active muscular actuation and passive pendulum dynamics.

Future Directions

Since the underlying theme of my dissertation is focused on human running, I discuss here some future directions for research based on my first three studies. The results from my first two studies provide support for my general hypothesis that humans not only minimize energetic cost but also optimize for balance during locomotion. I focused on the link between energetic cost and lateral balance by investigating the roles of step width and arm swing during human running. I found that 1) step width adjustments are the primary mechanism for maintaining lateral balance and 2) arm swing does not assist with lateral balance. My findings

lead me to believe that arm swing must assist with other aspects of balance control, specifically motions about the vertical axis.

Therefore, my future efforts will focus on the effects of arm swing on upper body rotation. Just as I studied step width and its variability as indicators of lateral balance, I put forward the idea that studying upper body rotation and its variability can serve as indicators of “rotational” balance. This rationale is based on my experimental observations of human subjects running without arm swing. When arm swing was eliminated, I observed a tendency for subjects to modify torso rotation, most likely reflecting a compensatory strategy to maintain rotational balance by counteracting the energy from the swinging legs. Studying the role of arm swing and its affect on rotational balance may help to identify the underlying mechanism(s) that are responsible for the increase in the energetic cost of running when arm swing is eliminated.

The findings from my third study provide motivation for future research on the use of running-specific prostheses during sprint-running. For example, I found that M-L foot placement and its variability were significantly asymmetric between the unaffected (UL) and affected (AL) leg of sprinters with unilateral trans-tibial amputations. Across the 2.3 fold increase in running speed, the variability of the AL increased to a greater extent than the UL, indicating greater difficulty with the control of lateral balance when using the AL. Thus, it appears that the current design of passive-elastic running-specific prostheses disrupts the control of lateral balance. An area for future research is to improve the current design of running-specific prostheses so that they meet the demands required to overcome instabilities in the M-L direction, potentially improving foot placement symmetry between the AL and UL and overall lateral balance.

7. REFERENCES

- Alexander, R.M., 1984. Elastic energy stores in running vertebrates. *American Zoologist* 24, 85-94.
- Alexander, R.M., 1991. Energy-saving mechanisms in walking and running. *Journal of Experimental Biology* 160, 55-69.
- Alexander, R.M., Bennetclark, H.C., 1977. Storage of elastic strain-energy in muscle and other tissues. *Nature* 265, 114-117.
- Arellano, C.J., Kram, R., 2011. The effects of step width and arm swing on energetic cost and lateral balance during running. *Journal of Biomechanics* 44, 1291-1295.
- Arellano, C.J., Kram, R., 2012. The energetic cost of maintaining lateral balance during human running. *Journal of Applied Physiology* 112, 427-434.
- Arellano, C.J., O'Connor, D.P., Layne, C., Kurz, M.J., 2009. The independent effect of added mass on the stability of the sagittal plane leg kinematics during steady-state human walking. *Journal of Experimental Biology* 212, 1965-1970.
- Baker, G.L., Blackburn, J.A., 2005. *The Pendulum: a case study in physics*. Oxford University Press, New York, pp. 8-66.
- Ballesteros, M.L., Buchthal, F., Rosenfalck, P., 1965. The pattern of muscular activity during the arm swing of natural walking. *Acta Physiologica Scandinavica* 63, 296-310.
- Bauby, C.E., Kuo, A.D., 2000. Active control of lateral balance in human walking. *Journal of Biomechanics* 33, 1433-1440.
- Blickhan, R., 1989. The spring mass model for running and hopping. *Journal of Biomechanics* 22, 1217-1227.
- Bobbert, A.C., 1960. Energy expenditure in level and grade walking. *Journal of Applied Physiology* 15, 1015-1021.
- Bøje, O., 1944. Energy production, pulmonary ventilation, and length of steps in well-trained runners working on a treadmill. *Acta Physiologica Scandinavica*.
- Brockway, J.M., 1987. Derivation of formulae used to calculate energy expenditure in man. *Human Nutrition: Clinical Nutrition* 41, 463-471.
- Brooks, G.A., Fahey, T.D., Baldwin, K.M., 2004. *Exercise physiology: Human bioenergetics and its applications*. McGraw-Hill, London.
- Bruijn, S.M., Meijer, O.G., Beek, P.J., van Dieen, J.H., 2010. The effects of arm swing on human gait stability. *Journal of Experimental Biology* 213, 3945-3952.
- Bruijn, S.M., Meijer, O.G., van Dieen, J.H., Kingma, I., Lamothe, C.J., 2008. Coordination of leg swing, thorax rotations, and pelvis rotations during gait: the organisation of total body angular momentum. *Gait & Posture* 27, 455-462.
- Buckley, J.G., 1999. Sprint kinematics of athletes with lower-limb amputations. *Archives of Physical Medicine and Rehabilitation* 80, 501-508.
- Cavagna, G.A., Heglund, N.C., Taylor, C.R., 1977. Mechanical work in terrestrial locomotion: two basic mechanisms for minimizing energy-expenditure. *The American Journal of Physiology - Regulatory, Integrative and Comparative Physiology* 233, R243-R261.
- Cavagna, G.A., Kaneko, M., 1977. Mechanical work and efficiency in level walking and running. *Journal of Physiology-London* 268, 467-481.
- Cavagna, G.A., Margaria, R., 1966. Mechanics of walking. *Journal of Applied Physiology* 21, 271-278.

- Cavagna, G.A., Margaria, R., Saibene, F.P., 1963. External work in walking. *Journal of Applied Physiology* 18, 1-9.
- Cavagna, G.A., Saibene, F.P., Margaria, R., 1964. Mechanical work in running. *Journal of Applied Physiology* 19, 249-256.
- Cavagna, G.A., Thys, H., Zamboni, A., 1976. Sources of external work in level walking and running. *Journal of Physiology-London* 262, 639-657.
- Cavanagh, P.R., 1987. The biomechanics of lower extremity action in distance running. *Foot & Ankle* 7, 197-217.
- Cavanagh, P.R., Williams, K.R., 1982. The effect of stride length variation on oxygen uptake during distance running. *Medicine and Science in Sports and Exercise* 14, 30-35.
- Chang, Y.H., Kram, R., 1999. Metabolic cost of generating horizontal forces during human running. *Journal of Applied Physiology* 86, 1657-1662.
- Chapman, M.W., Ralston, H.J., 1964. Effect of immobilization of the back and arms on energy expenditure during level walking, Technical Report 52. University of California, San Francisco.
- Cohen, J., 1988. Statistical power analysis for the behavioral sciences. Lawrence Erlbaum Associates, New Jersey.
- Collins, S.H., Adamczyk, P.G., Kuo, A.D., 2009. Dynamic arm swinging in human walking. *Proceedings of the Royal Society B: Biological Sciences* 276, 3679-3688.
- Cotes, J., Meade, F., 1960. The energy-expenditure and mechanical energy demand in walking. *Ergonomics* 3, 97-119.
- Cram, J.R., Kasman, G.S., 1998. Electrode Placements, in: Cram, J.R., Kasman, G.S., Holtz, J. (Eds.), *Introduction to Surface Electromyography*. Aspen, Gaithersburg, MD, pp. 289-293.
- Dean, J.C., Alexander, N.B., Kuo, A.D., 2007. The effect of lateral stabilization on walking in young and old adults. *IEEE Transactions on Biomedical Engineering* 54, 1919-1926.
- Den Hartog, J.P., 1985. *Mechanical Vibrations*. McGraw-Hill, New York, pp. 23-78.
- Doke, J., Donelan, J.M., Kuo, A.D., 2005. Mechanics and energetics of swinging the human leg. *Journal of Experimental Biology* 208, 439-445.
- Doke, J., Kuo, A.D., 2007. Energetic cost of producing cyclic muscle force, rather than work, to swing the human leg. *Journal of Experimental Biology* 210, 2390-2398.
- Donelan, J.M., Kram, R., Kuo, A.D., 2001. Mechanical and metabolic determinants of the preferred step width in human walking. *Proceedings of the Royal Society B: Biological Sciences* 268, 1985-1992.
- Donelan, J.M., Kram, R., Kuo, A.D., 2002a. Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. *Journal of Experimental Biology* 205, 3717-3727.
- Donelan, J.M., Kram, R., Kuo, A.D., 2002b. Simultaneous positive and negative external mechanical work in human walking. *Journal of Biomechanics* 35, 117-124.
- Donelan, J.M., Shipman, D.W., Kram, R., Kuo, A.D., 2004. Mechanical and metabolic requirements for active lateral stabilization in human walking. *Journal of Biomechanics* 37, 827-835.
- Dunnett, C.W., 1955. A multiple comparison procedure for comparing several treatments with a control. *Journal of the American Statistical Association* 50, 1096-1121.
- Dunnett, C.W., 1964. New tables for multiple comparisons with a control. *International Biometric Society* 20, 482-491.

- Egbuonu, M.E., Cavanagh, P.R., Miller, T.A., 1990. Degradation of running economy through changes in running mechanics. *Medicine and Science in Sports and Exercise* 22, S17.
- Elftman, H., 1938. The function of the arms in walking. *Human Biology* 11, 529-535.
- Elftman, H., 1966. Biomechanics of muscle: with particular application to studies of gait. *The Journal of Bone and Joint Surgery*.
- Enoka, R.M., 2002. Movement Forces, *Neuromechanics of human movement*, 3rd ed. Human Kinetics, Champaign, IL, pp. 57-118.
- Enoka, R.M., Miller, D.I., Burgess, E.M., 1982. Below-knee amputee running gait. *American Journal of Physical Medicine and Rehabilitation* 61, 66-84.
- Farley, C.T., Glasheen, J., McMahon, T.A., 1993. Running springs-speed and animal size. *Journal of Experimental Biology* 185, 71-86.
- Farley, C.T., Ferris, D.P., 1998. Biomechanics of walking and running: Center of mass movements to muscle action. *Exercise Sport Science Reviews* 26, 253-285.
- Farley, C.T., Gonzalez, O., 1996. Leg stiffness and stride frequency in human running. *Journal of Biomechanics* 29, 181-186.
- Farley, C.T., McMahon, T.A., 1992. Energetics of walking and running-insights from simulated reduced-gravity experiments. *Journal of Applied Physiology* 73, 2709-2712.
- Fenn, W.O., 1924. The relation between the work performed and the energy liberated in muscular contraction. *Journal of Physiology-London* 58, 0373-0395.
- Fenn, W.O., 1930a. Frictional and kinetic factors in the work of sprint running. *American Journal of Physiology* 92, 583-611.
- Fenn, W.O., 1930b. Work against gravity and work due to velocity changes in running - movements of the center of gravity within the body and foot pressure on the ground. *American Journal of Physiology* 93, 433-462.
- Ferris, D.P., Liang, K., Farley, C.T., 1999. Runners adjust leg stiffness for their first step on a new running surface. *Journal of Biomechanics* 32, 787-794.
- Ferris, D.P., Louie, M., Farley, C.T., 1998. Running in the real world: adjusting leg stiffness for different surfaces. *Proceedings of the Royal Society B: Biological Sciences* 265, 989-994.
- Franz-Josef, E., 1998. The linearized equation of motion for a horizontally driven pendulum are described in "The Pendulum Lab": <http://www.elmer.unibas.ch/pendulum/index.html>. Animations of a horizontally driven pendulum can also be found there.
- Gerdy, P.N., 1829. Memoires sur le mecanisme de la marcha de l'homme. *J Physiol exp path* 9, 1-28.
- Givoni, B., Goldman, R.F., 1971. Predicting metabolic energy cost. *Journal of Applied Physiology* 30, 429-433.
- Gottschall, J., Kram, R., 2003. Energy cost and muscular activity required for propulsion during walking. *Journal of Applied Physiology*, 1766-1772.
- Gottschall, J., Kram, R., 2005. Energy cost and muscular activity required for leg swing during walking. *Journal of Applied Physiology*, 23-30.
- Grabowski, A., Farley, C.T., Kram, R., 2005. Independent metabolic costs of supporting body weight and accelerating body mass during walking. *Journal of Applied Physiology* 98, 579-583.
- Grabowski, A.M., McGowan, C.P., McDermott, W.J., Beale, M.T., Kram, R., Herr, H.M., 2010. Running-specific prostheses limit ground-force during sprinting. *Biology Letters* 6, 201-204.

- Hamner, S.R., Seth, A., Delp, S.L., 2010. Muscle contributions to propulsion and support during running. *Journal of Biomechanics* 43, 2709-2716.
- Hanavan, E.P., 1964. A mathematical model for the human body. Technical Report, Wright-Patterson Air Force Base.
- He, J.P., Kram, R., McMahon, T.A., 1991. Mechanics of running under simulated low gravity. *Journal of Applied Physiology* 71, 863-870.
- Heglund, N.C., Cavagna, G.A., Taylor, C.R., 1982a. Energetics and mechanics of terrestrial locomotion. 3. Energy changes of the center of mass as a function of speed and body size in birds and mammals. *Journal of Experimental Biology* 97, 41-56.
- Heglund, N.C., Fedak, M.A., Taylor, C.R., Cavagna, G.A., 1982b. Energetics and mechanics of terrestrial locomotion. 4. Total mechanical energy changes as a function of speed and body size in birds and mammals. *Journal of Experimental Biology* 97, 57-66.
- Herr, H., Popovic, M., 2008. Angular momentum in human walking. *Journal of Experimental Biology* 211, 467-481.
- Hill, A.V., 1922. The maximum work and mechanical efficiency of human muscles, and their most economical speed. *Journal of Applied Physiology* 56, 19-41.
- Hill, A.V., 1938. The heat of shortening and the dynamic constants of muscle. *Proceedings of the Royal Society B: Biological Sciences* 126, 136-195.
- Hinrichs, R.N., 1987. Upper extremity function in running: II. Angular momentum considerations. *International Journal of Sport Biomechanics* 3, 242-263.
- Hinrichs, R.N., 1990. Whole body movement: coordination of arms and legs in walking and running, in: Winters, J., Woo, S.Y.L. (Eds.), *Multiple muscle systems: biomechanics and movement organization*. Springer-Verlag, New York, pp. 694-705.
- Hinrichs, R.N., Cavanagh, P.R., Williams, K.R., 1987. Upper extremity function in running: I. Center of mass and propulsion considerations. *International Journal of Sport Biomechanics* 3, 222-241.
- Hogberg, P., 1952. How do stride length and stride frequency influence the energy-output during running? *European Journal of Applied Physiology and Occupational Physiology* 14, 437-441.
- Hogue, R.E., 1969. Upper-extremity muscular activity at different cadences and inclines during normal gait. *Physical Therapy* 49, 963-972.
- Holt, K.G., Hamill, J., Andres, R.O., 1991. Predicting the minimal energy costs of human walking. *Medicine and Science in Sports and Exercise* 23, 491-498.
- Inman, V.T., Ralston, H.J., Todd, F., 1981. *Human walking*. Williams & Wilkins, Baltimore.
- Jackson, K.M., Joseph, J., Wyard, S.J., 1978. Mathematical model of arm swing during human locomotion. *Journal of Biomechanics* 11, 277-289.
- Ker, R.F., Bennett, M.B., Bibby, S.R., Kester, R., Alexander, R.M., 1987. The spring in the arch of the human foot. *Nature* 325, 147-149.
- Kerdok, A.E., Biewener, A.A., McMahon, T.A., Weyand, P.G., Herr, H.M., 2002. Energetics and mechanics of human running on surfaces of different stiffnesses. *Journal of Applied Physiology* 92, 469-478.
- Kram, R., Griffin, T.M., Donelan, J.M., Chang, Y.H., 1998. Force treadmill for measuring vertical and horizontal ground reaction forces. *Journal of Applied Physiology* 85, 764-769.
- Kram, R., Taylor, C.R., 1990. Energetics of running: a new perspective. *Nature* 346, 265-267.
- Kuo, A.D., 1999. Stabilization of lateral motion in passive dynamic walking. *International Journal of Robotics Research* 18, 917-930.

- Kuo, A.D., 2001. A simple model of bipedal walking predicts the preferred speed-step length relationship. *Journal of Biomechanical Engineering-Transactions of the ASME* 123, 264-269.
- Kuo, A.D., 2007. The six determinants of gait and the inverted pendulum analogy: A dynamic walking perspective. *Human Movement Sciences* 26, 617-656.
- Kuo, A.D., Donelan, J.M., 2009. Comment on "Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking" (Neptune et al., 2001) and "Muscle mechanical work requirements during normal walking: The energetic cost of raising the body's center-of-mass is significant" (Neptune et al., 2004). *Journal of Biomechanics* 42, 1783-1785.
- Kuo, A.D., Donelan, J.M., Ruina, A., 2005. Energetic consequences of walking like an inverted pendulum: Step-to-step transitions. *Exercise and Sport Science Reviews* 33, 88-97.
- Li, Y., Wang, W.J., Crompton, R.H., Gunther, M.M., 2001. Free vertical moments and transverse forces in human walking and their role in relation to arm-swing. *Journal of Experimental Biology* 204, 47-58.
- Lloyd, B., Zacks, R., 1972. The mechanical efficiency of treadmill running against a horizontal impeding force. *The Journal of Physiology*, 355-363.
- Mann, R., Herman, J., 1985. Kinematic analysis of Olympic sprint performance: men's 200 meters. *International Journal of Sport Biomechanics* 1, 151-162.
- Mann, R.V., 1981. A kinetic analysis of sprinting. *Medicine and Science in Sports and Exercise* 13, 325-328.
- Margaria, R., 1976. *Biomechanics and energetics of muscular exercise*. Clarendon Press, Oxford, United Kingdom.
- Margaria, R., Cerretelli, P., Aghemo, P., Sassi, G., 1963. Energy cost of running. *Journal of Applied Physiology* 18, 367-370.
- Marsh, R.L., Ellerby, D.J., Henry, H.T., Rubenson, J., 2006. The energetic costs of trunk and distal-limb loading during walking and running in guinea fowl *Numida meleagris*: I. Organismal metabolism and biomechanics. *Journal of Experimental Biology* 209, 2050-2063.
- Martin, P.E., Rothstein, D.E., Larish, D.D., 1992. Effects of age and physical-activity status on the speed-aerobic demand relationship of walking. *Journal of Applied Physiology* 73, 200-206.
- McClay, I.S., Cavanagh, P.R., 1994. Relationship between foot placement and mediolateral ground reaction forces during running. *Clinical Biomechanics* 9, 117-123.
- McGeer, T., 1990a. Passive bipedal running. *Proceedings of the Royal Society B: Biological Sciences* 240, 107-134.
- McGeer, T., 1990b. Passive dynamic walking. *International Journal of Robotics Research* 9, 62-82.
- McGowan, C.P., Grabowski, A.M., McDermott, W.J., Herr, H.M., Kram, R., 2012. Leg stiffness of sprinters using running-specific prostheses. *Journal of the Royal Society Interface*.
- McMahon, T.A., 1984. *Muscles, Reflexes, and Locomotion*. Princeton University Press, Princeton, New Jersey.
- McMahon, T.A., Cheng, G.C., 1990. The mechanics of running-How does stiffness couple with speed? *Journal of Biomechanics* 23, 65-78.
- McMahon, T.A., Valiant, G., Frederick, E.C., 1987. Groucho running. *Journal of Applied Physiology* 62, 2326-2337.

- Menier, D.R., Pugh, L.G.C., 1968. Relation of oxygen intake and velocity of walking and running in competition walkers. *Journal of Physiology-London* 197, 717-721.
- Mercier, J., Le Gallais, D., Durand, M., Goudal, C., Micallef, J.P., Prefaut, C., 1994. Energy expenditure and cardiorespiratory responses at the transition between walking and running. *European Journal of Applied Physiology and Occupational Physiology* 69, 525-529.
- Mero, A., Komi, P.V., Gregor, R.J., 1992. Biomechanics of sprint running - A review. *Sports Medicine* 13, 376-392.
- Miller, R.H., Caldwell, G.E., Van Emmerik, R.E., Umberger, B.R., Hamill, J., 2009. Ground reaction forces and lower extremity kinematics when running with suppressed arm swing. *Journal of Biomechanical Engineering* 131, 124502.
- Mochon, S., McMahon, T.A., 1980. Ballistic Walking. *Journal of Biomechanics* 13, 49-57.
- Modica, J.R., Kram, R., 2005. Metabolic energy and muscular activity required for leg swing in running. *Journal of Applied Physiology* 98, 2126-2131.
- Moed, B., Kram, R., 2004. Metabolic costs of forward propulsion and leg swing at different running speeds, ISB XXth Congress-ASB 29th Annual Meeting, Cleveland, Ohio.
- Morton, D.J., Fuller, D.D., 1952. Human locomotion and body form: a study of gravity and man. Williams and Wilkins Company, Baltimore.
- Neptune, R.R., Zajac, F.E., Kautz, S.A., 2004. Muscle mechanical work requirements during normal walking: the energetic cost of raising the body's center-of-mass is significant. *Journal of Biomechanics* 37, 817-825.
- Neptune, R.R., Zajac, F.E., Kautz, S.A., 2009. Comment on "Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking" (Neptune et al., 2001) and "Muscle mechanical work requirements during normal walking: The energetic cost of raising the body's center-of-mass is significant" (Neptune et al., 2004) Response. *Journal of Biomechanics* 42, 1786-1789.
- Nolan, L., 2008. Carbon fibre prostheses and running in amputees: A review. *Foot and Ankle Surgery* 14, 125-129.
- Ortega, J.D., Fehلمان, L.A., Farley, C.T., 2008. Effects of aging and arm swing on the metabolic cost of stability in human walking. *Journal of Biomechanics* 41, 3303-3308.
- Owings, T.M., Grabiner, M.D., 2003. Measuring step kinematic variability on an instrumented treadmill: how many steps are enough? *Journal of Biomechanics* 36, 1215-1218.
- Pontzer, H., Holloway, J.H., Raichlen, D.A., Lieberman, D.E., 2009. Control and function of arm swing in human walking and running. *Journal of Experimental Biology* 212, 523-534.
- Prince, F., Allard, P., Therrien, R.G., Mcfadyen, B.J., 1992. Running gait impulse asymmetries in below-knee amputees. *Prosthetics and Orthotics International* 16, 19-24.
- Ralston, H.J., 1958. Energy-speed relation and optimal speed during level walking. *Internationale Zeitschrift Fur Angewandte Physiologie, Einschliesslich Arbeitsphysiologie* 17, 277-283.
- Ralston, H.J., Lukin, L., 1969. Energy levels of human body segments during level walking. *Ergonomics* 12, 39-46.
- Roberts, T.J., Kram, R., Weyand, P.G., Taylor, C.R., 1998. Energetics of bipedal running I. Metabolic cost of generating force. *Journal of Experimental Biology* 201, 2745-2751.
- Roberts, T.J., Marsh, R.L., Weyand, P.G., Taylor, C.R., 1997. Muscular force in running turkeys: The economy of minimizing work. *Science* 275, 1113-1115.
- Sanderson, D.J., Martin, P.E., 1996. Joint kinetics in unilateral below-knee amputee patients during running. *Archives of Physical Medicine and Rehabilitation* 77, 1279-1285.

- Sokolove, M., 2012. The Fast Life of Oscar Pistorius, NYTimes.com, 18 Jan. <http://www.nytimes.com/2012/01/22/magazine/oscar-pistorius.html?_r=1&pagewanted=1&hpw>
- Taylor, C.R., 1985. Force development during sustained locomotion-A determinant of gait, speed and metabolic power. *Journal of Experimental Biology* 115, 253-262.
- Taylor, C.R., Heglund, N.C., Maloiy, G.M.O., 1982. Energetics and mechanics of terrestrial locomotion. 1. Metabolic energy-consumption as a function of speed and body size in birds and mammals. *Journal of Experimental Biology* 97, 1-21.
- Taylor, C.R., Heglund, N.C., McMahon, T.A., Looney, T.R., 1980. Energetic cost of generating muscular force during running - A comparison of large and small animals. *Journal of Experimental Biology* 86, 9-18.
- Teunissen, L.P., Grabowski, A., Kram, R., 2007. Effects of independently altering body weight and body mass on the metabolic cost of running. *Journal of Experimental Biology* 210, 4418-4427.
- Umberger, B.R., 2008. Effects of suppressing arm swing on kinematics, kinetics, and energetics of human walking. *Journal of Biomechanics* 41, 2575-2580.
- Umberger, B.R., 2010. Stance and swing phase costs in human walking. *Journal of The Royal Society Interface*, 1-13.
- Umberger, B.R., Martin, P.E., 2007. Mechanical power and efficiency of level walking with different stride rates. *Journal of Experimental Biology* 210, 3255-3265.
- van der Walt, W.H., Wyndham, C.H., 1973. An equation for prediction of energy expenditure of walking and running. *Journal of Applied Physiology* 34, 559-563.
- Warddrip, E., Kram, R., 2007. Disintegrating the metabolic cost of human running., University of Colorado. Thesis (Unpublished).
- Weber, W., Weber, E., 1836. *Mechanik der menschlichen Gehwerkzeuge*, Göttingen.
- Weyand, P.G., Bundle, M.W., McGowan, C.P., Grabowski, A., Brown, M.B., Kram, R., Herr, H., 2009. The fastest runner on artificial legs: different limbs, similar function? *Journal of Applied Physiology* 107, 903-911.
- Winter, D.A., 1990. *Biomechanics and motor control of human movement.*, 2nd ed. John Wiley & Sons, Inc.
- Workman, J.M., Armstrong, B.W., 1963. Oxygen cost of treadmill walking. *Journal of Applied Physiology* 18, 798-803.
- Wyndham, C.H., Strydom, N.B., van Graan, C.H., van Rensburg, A.J., Rogers, G.G., Greyson, J.S., van der Walt, W.H., 1971. Walk or jog for health. I. The energy costs of walking or running at different speeds. *South African Medical Journal* 45, 50-53.
- Yang, J.F., Winter, D.A., 1984. Electromyographic amplitude normalization methods: improving their sensitivity as diagnostic tools in gait analysis. *Archives of Physical Medicine and Rehabilitation* 65, 517-521.
- Zacks, R., 1973. The mechanical efficiencies of running and bicycling against a horizontal impeding force. *European Journal of Applied Physiology and Occupational Physiology* 31, 249-258.
- Zarrugh, M.Y., Todd, F.N., Ralston, H.J., 1974. Optimization of energy expenditure during level walking. *European Journal of Applied Physiology and Occupational Physiology* 33, 293-306.