MECHANICS AND ENERGETICS OF STEP-TO-STEP TRANSITIONS

by

Caroline Hooi-Nee Soo
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Name: **Caroline Hooi-Nee Soo**

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Examining Committee:

Chair:

Dr. James Wakeling
Assistant Professor – Department of Biomedical Physiology and Kinesiology

Dr. James Maxwell (Max) Donelan
Senior Supervisor
Assistant Professor – Department of Biomedical Physiology and Kinesiology

Dr. Stephen Robinovitch
Supervisor
Associate Professor – Department of Biomedical Physiology and Kinesiology

Dr. Janice J. Eng
External Examiner
Professor – Department of Physical Therapy
University of British Columbia

Date Defended/Approved: August 7, 2009

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ABSTRACT

The major determinant of walking's metabolic cost is the work required to redirect the centre of mass velocity during step-to-step transitions. My first aim was to isolate transitions from other contributors to walking mechanics. The results demonstrated that sagittal plane rocking reproduced the important characteristics of walking's transitions including a strong dependence of work on step length and a proportional increase in metabolic cost. My second aim was to use rocking to gain insight into pathological gait's elevated cost. Physics-based mathematical models predict sub-optimal transitions occur when one or both legs are unable to generate mechanical power with the optimal timing and magnitude, requiring a greater magnitude of total work and an increase in metabolic cost. I tested this prediction by immobilising the ankle joints of healthy subjects to simulate sub-optimal transitions and found that joint immobilization indeed caused sub-optimal transitions thereby increasing transition work and metabolic cost.

Keywords: mechanics; metabolic cost; step-to-step transition; rocking; ankle; immobilization

To my parents

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CHAPTER 1: INTRODUCTION

The metabolic cost of pathological walking is more expensive than healthy walking. Pathological walking increases the energy requirement and decreases the quality of life of patients (Thijssen *et al.*, 2007; Waters & Mulroy, 1999). The dominant paradigm in the rehabilitation community, the six determinant of gait paradigm (Saunders *et al.*, 1953), which tries to link some biomechanical features of walking to the cost of walking has been questioned (Gard & Childress, 1997, 1999; Kerrigan *et al.*, 2001). The work of my thesis explores predictions made by physics-based mathematical models (Kuo *et al.*, 2005; Ruina *et al.*, 2005) which may provide an alternative explanation for the elevated metabolic cost of pathological walking.

Step-to-step transitions, roughly corresponding to the double support phase of walking, are a major determinant of the metabolic cost of walking (Kuo et al., 2005). Physics-based walking models predict that there is an optimal way to perform transitions – when the two legs perform equal, but opposite amount of mechanical work simultaneously (Kuo et al., 2005; Ruina et al., 2005). Healthy humans appear to walk optimally at all walking speeds (Donelan *et al.*, 2002a). In sub-optimal transitions, the magnitudes of the mechanical work performed by the two legs are not equal and nor are they performed simultaneously. These sub-optimal transitions increase the total mechanical work resulting in increased metabolic cost. To better understand the effects of sub-optimal transitions on the metabolic cost of walking, I conducted two studies. I first isolated step-to-step transitions because, while transition work make up about 60-70% of the total energy of walking, other contributors such as leg swing may confound our measures of work. In the second experiment, I tested whether sub-optimal transitions are associated with increased metabolic cost. The findings of the proposed research support the view that one way to minimize the metabolic cost of walking is to focus on the timing and the magnitude of mechanical work perform by each leg during transitions.

1.1 Current paradigm in walking rehabilitation

A common goal of walking rehabilitation treatment is “normalizing” pathological gait. For example, in amputee walking rehabilitation, “…the primary goal of the clinician is to create prostheses and orthoses facilitating movements that approximate the kinematics and energetic of normal walking”(Michael, 2006). The current goal of normalizing of gait reflects the six determinant of gait paradigm, which suggests that metabolic cost of walking is minimized by minimizing the centre-of-mass (COM) motion by using a set of six walking characteristics found in healthy walking.

This paradigm stems from the classical paper by Saunders, Inman and Eberhardt’s paper (1953). The authors claim that the six determinants of gait: pelvic rotation, pelvic tilt, knee flexion in stance, foot mechanisms, knee mechanisms, and lateral pelvic displacement, found in healthy walking, decrease the vertical COM displacement amplitude. Furthermore, they claim that decreasing COM deviation decrease the metabolic cost of walking. The authors claim patients that cannot exhibit these six movements suffer the consequence of increased metabolic cost. The original paper did not have a testable hypothesis nor any data to support their contentions. For almost 50 years, this paradigm went unchallenged (Childress & Gard, 2006). As a result, the goal of minimizing the COM deviation has been very entrenched in the field of walking rehabilitation and is still accepted as a goal of walking (Orendurff *et al.*, 2004).

Current approaches for the rehabilitation of gait pathologies assume that the best way to minimize the metabolic cost of walking is to exhibit the six determinants of gait (Childress & Gard, 2006). Recently, researchers have questioned the validity of the first three determinants of gait to lowering the centre of mass deviation (Gard & Childress, 1997, 1999; Kerrigan et al., 2001). Researchers have also questioned Saunders and colleagues' fundamental claim that a decrease in the centre of mass deviation decreases the metabolic cost. When subjects flattened the centre of mass displacement while walking by adopting a "Groucho" walking pattern, they increased their metabolic cost of walking by two-fold compared to their regular walking pattern (Gordon *et al.*, 2003; Ortega & Farley, 2005). The fundamental goal of this paradigm is undesirable because: 1) the metabolic cost of walking actually increased when subject attempted to flatten out their vertical centre of mass movement and 2) half of the determinants don't accomplish their proposed function. In light of the shortcomings of this paradigm, a new approach to investigate the underlying biomechanical determinants of the metabolic cost of healthy and pathological walking is required.

1.2 Walking models and step-to-step transitions

1.2.1 Inverted pendulum model

The inverted pendulum walking model captures key elements of walking. This model approximates the stance leg as a mass-less strut and the body as a point mass. The key feature in this model is the conservation of energy of the centre of mass. The inverted pendulum mechanism allows for the transfer between kinetic energy and potential energy in an energy conservative manner (Cavagna *et al.*, 1977; Cavagna & Kaneko, 1977). While this model is sufficient to capture the energy dynamics of the centre of mass, this model is unable to make predictions about the biomechanical determinants of the metabolic cost of walking.

1.2.2 Dynamic walking model

Dynamic walking models are simple models that require minimal or no control and are completely reliant upon gravitational and inertial forces to maintain motion (McGeer, 1990). The essence underlying these models is the concept of exploiting the dynamics of the system to capture energy from the environment, thus, limiting the amount of energy the system needs to generate. Tad McGeer (1990) first introduced these models and have successfully built simple robots, that have two struts approximating the legs joined together at the hip, that can walk down a shallow slope without energy input or control. Even though dynamic walking models are different from human walking, these models offer us insight into a possible energy saving mechanisms.

1.2.3 Step-to-step transitions

To study human walking, I use a dynamic walking model made up of two struts, which approximate each leg, joined together with a frictionless joint (i.e. the hip joint) where a point mass approximates the COM of the body (Donelan et al., 2002a, 2002b). The walking cycle alternates between single support or stance phase and double support phases. During single stance, when only one foot is on the ground, the centre of mass follows the pendular arc trajectory and requires no additional energy. During the double support phase, the leading leg collides with the ground and is redirected from the arc of the trailing leg to the arc of the leading leg. A consequence of the stance leg behaving like an inverted pendulum is the need to transition the COM velocity from the inverted pendulum of the trailing leg to the inverted pendulum of the leading leg (Donelan et al., 2002b; Kuo et al., 2005). Since negative (collision) work is required to redirect the COM velocity between inverted pendulum-like single support phases, the positive (push-off) work done must be performed to replace the dissipated energy and maintain a steady walking speed (Donelan et al., 2002a, 2002b; Kuo et al., 2005). Healthy human perform optimal transition at all walking speeds (Donelan et al., 2002a).

The same walking models predict that there is an optimal way to perform transitions (Ruina et al., 2005). The least amount of mechanical work is required when the two legs perform equal, but opposite amounts of mechanical work simultaneously. The inability to perform optimal transitions with the right amount of mechanical work or with the correct timing causes elevated total mechanical work and metabolic cost. While we are starting to gain insight into optimal transitions, predictions about the results of sub-optimal transitions have not been systematically tested; the goal of this thesis is to test these predictions.

1.3 Mechanical work

Walking requires mechanical work. Moving a body, such as walking, requires changes in the energy of the system, which results from a force being applied over a given displacement. If the force is applied in the same direction as the displacement, then positive work, in Joules, is performed on the body. Conversely, if the force and distance are in the opposite directions, then negative work is performed. Positive mechanical work is performed by muscles that are active and shortening. Negative mechanical work is performed by muscles which are active and lengthening (McMahon, 1984). While positive work is limited to an active concentric muscle source, negative work can also be performed simultaneously by passive structures such as soft tissues and tendons (Alexander & Bennet-Clark, 1977; Fukunaga *et al.*, 2001; Pain, 2001, 2002; Roberts *et al.*, 1997; Stefanyshyn, 2000; Wakeling *et al.*, 2003). Mechanical power, in Watts, is another way of quantifying the changes in mechanical energy and is defined as the mechanical work over time (work rate) or the dot product of force and velocity.

In the context of healthy walking step-to-step transitions, negative mechanical work is performed during the collision phases by the leading leg starting at heel strike to limit the progression of the COM velocity from the pendular arc of the trailing leg. The trailing leg performs positive push-off work to redirect the COM velocity to the pendular arc of the leading leg. During level walking at constant velocity, no net work is performed. This zero net work can be obtained in two ways. One, no work is being performed by either limb, or two, equal amounts of positive and negative work are performed.

1.3.1 Measuring muscle mechanical work

Work can be measured using direct or indirect measurements. Direct measurements of work by embedding miniature strain gauges into the muscle are possible (Komi, 1990; Roberts et al., 1997). Due to the invasive nature of direct measurements, I will not be considering direct methods for our study. Indirect measurements of work are easier to perform. However, they are all estimates of mechanical work. Every indirect measurement method has individual strengths and weakness and is limited by the assumptions of each model that underlie the method. While all these methods define work in the same way, inherent simplifications of the components of the model such as the origin of force, the structure of the bone and the location of joint centres yield different mechanical work estimation. Since these methods cannot detect agonist-antagonistic co-contraction, the mechanical work estimates may underestimate the actual work performed. Another common limitation is the origin of force. While net work is calculated from the resultant forces, the composition of the net work by the total positive work and total negative work is underestimated. The resultant force may be made up of simultaneous but opposing work of multiple muscles or from different levels of muscle organization such as cross bridges, sarcomeres or motor units.

In this thesis, two non-invasive methods of estimating mechanical work will be used: the individual limb method (ILM) (Donelan *et al.*, 2001, 2002b) and the inverse dynamics or joint power method (JPM) (Winter, 1990). The ILM approximates the body as a point mass with two massless legs. Having two forces directed along each leg would be more representative of human anatomy because the individual forces represent the individual legs. With the ILM, I can measure all the work done on the COM but cannot attribute the work to any particular source. The JPM model uses multiple rigid link segments (i.e. feet, shanks, thighs) and assumes those joint torques are caused by muscles spanning the joint. Because this method involves multiple segments, the researchers must make assumptions about the modelled segments such as the anthropometry of the link segments and the location of joint (de Leva, 1996). Unlike the ILM, the JPM work can be attributed to a specific joint – but only the joints that are modelled. In our study, I will be using these two methods for estimating work because of how they complement each other. The total work measured by the ILM can be attributed to specific JPM joint work.

1.4 Metabolic cost of gait

Minimizing the metabolic cost of walking appears to be a fundamental goal of locomotion (Alexander, 1989; Cavagna et al., 1977). In healthy walking, there are many determinants of the metabolic cost of walking such as step length and step frequency. However, we choose to walk in a manner that minimizes the metabolic cost (Alexander, 2002; Atzler & Herbst, 1927). Furthermore, when one determinant is constrained, we modify other determinants to walk in a way that reduce the cost of walking. For example, when we are constrained to walk at a fixed velocity, we choose to walk at the step frequency that result in the least energy expenditure (Alexander, 2002). In healthy walking, walking at a comfortable speeds requires more than three times the energy of standing (Waters & Mulroy, 1999). The metabolic cost of pathological walking is even more expensive - twice the amount as speed-matched able-bodied walking (Fisher & Gullickson, 1978; Waters & Mulroy, 1999). An often-stated functional goal of walking rehabilitation is to reduce the metabolic cost of walking (Platts 2006, Thijssen, 2007). Minimizing the metabolic cost of walking is important because "…higher energy costs predisposes…survivors to a sedentary life…" which may affect the quality of daily activities of patients (Thijssen et al., 2007).

1.4.1 Measuring metabolic cost

Muscles transform chemical energy from food into mechanical work. Metabolic cost is defined as the energy required to perform a given amount of work. To fuel the muscles and perform work, we need to ingest food, which is then broken up into the three basic components: carbohydrates, fats and proteins. Through different metabolic pathways these components are used as substrates to produce adenosine triphosphate (ATP) - the energetic currency that allows muscles to contract (Brooks *et al.*, 2000). Contracting muscles produces work across joints, which allow movement. During aerobic pathways, heat is given off as a by-product when each food component undergoes chemical reactions by interacting with oxygen and giving off carbon dioxide, water and energy (Brooks et al., 2000). The relative amounts of oxygen required, or carbon dioxide and water released will depend on the stoichiometric ratios of the chemical reaction of each of the three basic components. To estimate the metabolic cost, I use a well-established predictive equation that takes into account the contributions of the three basic food components. This equation estimates aerobic metabolic cost from the average oxygen consumed and the carbon dioxide produced during the steady state of a given activity (Brockway, 1987).

1.4.2 Efficiency

Efficiency is the ratio of the mechanical work to metabolic cost. A helpful way to think about efficiency is how much of the food energy is transformed into mechanical work while the reminder energy dissipates as heat. The sources of estimates of maximum positive and negative work efficiencies range from isolated muscle studies (Fenn, 1923; Hill, 1938) to human whole body walking experiments (Margaria, 1963). Efficiencies for human whole body experiments come from sloped walking. During uphill walking, the muscles of the leg overcome gravity by undergoing concentric contraction and generating positive mechanical work. Conversely, during downhill walking, the muscles are predominantly undergoing eccentric contraction and generating negative work against gravity. The maximum efficiency for positive work and negative work is approximately 25% and -120%, respectively (McMahon, 1984).

During level walking at constant velocity, no net work is performed. This zero net work can be obtained in two ways. One, no work is being performed – this is unlikely because your muscles are active. The second possibility is when equal amounts of positive and negative work are performed, as is the case with optimal transitions. For example, when positive and negative work are combined to perform 1 J positive mechanical work, the maximum efficiency of positive work is 1J of positive mechanical work divided by (4J +0.83J) of metabolic cost. 1J of mechanical work divided by 4.83J metabolic cost yields a maximum efficiency of 21%. In human subjects, Donelan et al. (Donelan et al., 2002b) estimated that the efficiency of transition alone is approximately 10%. In my attempt to isolate step-to-step transitions and to study the effects of sub-optimal transitions on the metabolic cost, I will be expecting efficiencies around 10%.

1.4.3 Determinants of the metabolic cost of walking

Step-to-step transitions are a major determinant of the metabolic cost of walking, making up about 60-70% of the total energy of walking (Kuo et al., 2005). However, there are other important contributors such as leg swing (Doke *et al.*, 2005; Gottschall & Kram, 2005) and active lateral stabilization (Donelan *et al.*, 2004). For example, the cost of stabilizing gait can increase the metabolic cost of walking by approximately 10% (Donelan et al., 2004). The second largest known determinant, leg swing, may consume up to 1/3 of the total cost of the metabolic cost of walking (Doke et al., 2005) and thus could confound any conclusions made about step-to-step transition while using a walking paradigm. To limit the effect of leg swing, I will test whether sagittal plane rocking, a gait-like activity with the swing phase removed, isolates and retains the mechanical and energetic characteristics of transitions during walking.

CHAPTER 2: SAGITTAL PLANE ROCKING APPEARS TO ISOLATE STEP-TO-STEP TRANSITIONS

2.1 Introduction

The single support phase of walking is characterized by centre-of-mass (COM) motion similar to that of an inverted pendulum—the stance limb behaves much like a rigid strut allowing kinetic energy to be stored as gravitational potential energy and then returned in a nearly conservative manner (Cavagna & Kaneko, 1977). The COM moves along an arc dictated by the stance limb and the COM velocity is approximately perpendicular to the long axis of the limb. Consequently, each transition to a new stance limb (step-to-step transition) requires redirection of the COM velocity from one inverted pendulum arc to the next. Since ground reaction forces are directed approximately along each leg (Biewener, 1990), this redirection of the COM velocity requires negative work by the leading leg during the collision phase, which begins at heel contact.

To maintain a steady walking speed, positive work is needed to replace the leading leg negative collision work (Figure 2-1 A). While the positive push-off work could be performed at any time during a stride, dynamic walking models predict that the least COM work is required when positive work by the trailing leg and negative work by the leading leg are equal, opposite, and performed simultaneously (Kuo *et al.*, 2005; Ruina *et al.*, 2005). If the trailing leg does not push-off and perform an equal amount of positive work at the same time the leading leg performs negative work, up to four times more energy is lost in redirecting the COM velocity. This is because pushing off with the trailing leg partially redirects the COM velocity and reduces the amount of negative work required from the leading leg to complete the redirection. Measurements of the COM work performed by the trailing and leading legs during walking have demonstrated that the positive COM work performed by the trailing leg during double support is about equal in magnitude to the negative work performed by the leading leg during the same period (Donelan et al., 2002b).

Transition work depends strongly on step length during walking (Figure 2-1 B). This is because the work required to redirect the COM velocity increases with both the angle of redirection as well as the velocity of the COM. Dynamic walking models predict that transition work increases with the fourth power of step length when step frequency is fixed. This prediction was supported by empirical experiments on healthy humans—COM work rate increased with the fourth power of step length during length modulated walking (Donelan *et al.*, 2002a). Transition work in these experiments appeared to exact a proportional metabolic cost and accounted for approximately 60 -70% of metabolic cost during walking at a comfortable speed (Kuo et al., 2005).

While step-to-step transitions appear to be important, there are other essential sub-tasks to walking. For example, people may actively swing their legs to modulate their step frequency (Doke & Kuo, 2007). Because walking involves multiple sub-tasks, it can be difficult to attribute aspects of walking mechanics to any particular sub-task—both step-to-step transitions and leg swing contribute to

COM work and exact a metabolic cost (Doke et al., 2005; Donelan et al., 2002a). To study mechanics in isolation, it is desirable to separate transitions from other contributors to walking mechanics. However, this is difficult to do in walking—leg swing has a contribution to COM work even during length modulated walking where step frequency is kept constant.

The objective of this study was to determine whether a cyclical rocking task designed to eliminate the need to swing the legs or progress forward can be used to isolate step-to-step transitions. In this rocking task, subjects were instructed to rock backwards and forwards in the sagittal plane, matching the beat of a metronome, while I measured COM work and metabolic cost. To be a useful model of step-to-step transitions during walking, I hypothesize that rocking would need to capture the following characteristics (Donelan et al., 2002a, 2002b): a) during each transition, the leading leg would perform primarily negative work, while the trailing leg would perform primarily positive work; b) the majority of the work performed during the rocking cycle would be performed during double support, which approximates the transition phase, rather than single support, which approximates the inverted pendulum phase; c) work would increase strongly with step length defined as the distance between the feet during the transition, and d) metabolic cost would increase proportional to increases in COM work rate.

2.2 Materials and methods

2.2.1 Experimental Procedures

I measured the COM work and metabolic costs of rocking as a function of step length in 10 healthy subjects (9 males and 1 female; age = 28.7 ± 3.7 years; leg length = 0.92 ± 0.04 metres; mean ± standard deviation). Rocking consists of a series of rocking cycles with the motion restricted primarily to the sagittal plane. Each rocking cycle is comprised of a forward and a backward half-cycle (Figure 2-2 A). The forward half-cycle begins as the body reverses direction from moving backwards to moving forwards. As the body moves forwards, the front leg contacts the ground and the subject transitions from back leg single stance to front leg single stance. The body continues to move forward but, unlike walking, it ultimately reverses direction, thus beginning the backward rocking half-cycle. As the body moves backwards, the back leg contacts the ground and the subject transitions from front leg single stance to back leg single stance. The body continues to move backward until it reverses direction beginning the next rocking cycle. Each rocking cycle has two transitions with the front leg leading during the forward half-cycle and the back leg leading during the backward half-cycle. While rocking, our subject's arms were held stationary with hands on the hips. For convenience, I defined each new cycle as beginning at foot contact of the front leg. Prior to testing, all subjects gave their written informed consent and the experimental protocol was approved by the Simon Fraser University Office of Research Ethics.

To constrain the rocking task, subjects rocked at five different step lengths keeping rocking frequency constant. To account for difference in body size, subjects rocked at lengths equal to 60, 70, 80, 90 and 100% of their leg length. Each length was enforced by asking subjects to contact two pieces of tape (Figure 2-2 B), placed the correct distance apart at the front foot heel and back foot toe. I enforced rocking frequency by asking subjects to match their foot contacts to the beat of a metronome during the forward and backward half-cycles, forcing the half-cycles to have equal duration. To account for differences in body size, I asked subjects to rock at fractions of their natural or dimensionless frequency, $\sqrt{g/l}$, where l is leg length and gravity is g. The metronome was set to a dimensionless frequency of 0.50 equating to a frequency of 0.25 for the full rocking cycle. For an average subject, this equated to 50 rocking cycles per minute. To cue the subjects to change rocking directions, the metronome sounded at 50% and at 100% of each rocking cycle. I explored other rocking frequencies in pilot experiments and settled on this frequency because it produced large amplitude rocks without requiring the subjects to be stationary during support phases. On a day prior to the day of testing, I familiarized the subjects with rocking at different step lengths while keeping rocking frequency fixed. I randomized the order of step lengths and subjects switched between having their left or right leg as the front leg after each trial. Post-hoc data analyses demonstrated that subjects rocked at the desired step lengths and frequencies and rocking half-cycles were of equal duration.

For each subject and at each step length treatment, I measured the ground reaction forces, joint kinematics and the metabolic cost of rocking. The individual limb ground reaction forces (Figure 2-3 A and B) and moments were measured using two force plates—one under the front leg and another under the back leg (Bertec Corp., Columbus, Ohio). Signals were collected from both force platforms simultaneously at 960 Hz per channel. I used a 4th order, recursive, zero-phase-shift low-pass Butterworth filter with a cut-off frequency of 25 Hz to condition the ground reaction force signals. For kinematic data, I recorded the motion of 12 markers at 120 Hz with an 8-camera motion capture system (Vicon Motion Systems, Oxford Metrics Limited, Oxford, England). Markers were placed bilaterally on the fifth metatarsal of the foot, the lateral malleoli, the lateral epicondyles of the knee, the greater trochanters, and on the sacrum. Additionally, a secondary marker was placed on each thigh and shank segment to facilitate marker identification. I used a 4th order, recursive, zero-phase-shift low-pass Butterworth filter with a cut-off frequency of 6 Hz to condition the kinematic data. To determine metabolic cost, I measured and calculated average oxygen consumption ($\dot{V}_{O2}$) and carbon dioxide production ($\dot{V}_{CO2}$) using a metabolic analysis system (Vmax Encore, SensorMedics Corp., Yorba Linda, California). The metabolic cost was estimated by using the standard equation (Brockway, 1987):

$$\text{Metabolic Cost} = 16.58\,(\dot{V}_{O2}) + 4.51\,(\dot{V}_{CO2}).$$

To ensure that a steady state in rocking was reached, I averaged only the last three minutes of measured respiratory gases and the last one minute of measured ground reaction forces for each six-minute trial. Prior to rocking trials, I determined each subject's resting metabolic cost during standing.

2.2.2 Data Analyses

The ground reaction force data of each rocking trial lasted approximately 60 seconds collected in the last minute of each six-minute trial. These forces were divided up into rocking cycles where front foot contact (i.e. heel strike) represents the start of each cycle. I used the measured ground reaction forces and the individual limbs method (ILM) to calculate the COM work and COM work rate (Donelan et al., 2002b). Briefly, the rate of work performed on the COM by each leg is defined as the dot product of the individual limb ground reaction forces (Figure 2-3 A and B) and the COM velocity (Figure 2-3 C). To determine COM velocity, I first calculated the acceleration of the COM from the vector sum of the individual limb ground reaction forces and gravity and then integrated the COM acceleration with respect to time. Because rocking requires zero displacement over a complete cycle time, the integration constants were determined by forcing the average vertical, fore-aft and mediolateral COM velocity to equal zero for each rocking cycle to control for integration drift. The COM work rate was normalized in the time domain such that the lengths of rocking cycles were the same length. The COM work rate of each rocking cycle was then averaged to obtain variables representative for each subject.

I determined COM work from the cumulative time-integral of the instantaneous COM work rate (Figure 2-3D and E). Positive and negative COM work was determined by restricting the integration to intervals over which the instantaneous COM work rate was positive or negative, respectively. Similarly, I calculated COM work for a full rocking cycle by integrating over the complete cycle and for phases of the rocking cycle by integrating only over those phases (e.g. double support). Average COM work rate was defined as the mean positive COM work divided by the mean rocking cycle period.

I used inverse dynamics, also known as the joint power method (JPM), to determine the ankle, knee and hip joint power for each limb during rocking. Joint power was calculated by taking the product of the angular velocity of that joint and the moment of the segment distal to the joint. For each joint, the angular velocity was determined by differentiating the joint angle with respect to time. Markers on the lateral epicondyle of the knee, the lateral malleoli and the fifth metatarsal formed the ankle joint angle. Markers on the greater trochanter, the lateral epicondyle of the knee and the lateral malleoli, formed the knee joint angle. Markers on the sacrum, the greater trochanter and the lateral malleoli formed the hip joint angle. To determine the segmental moment, I used anthropomorphic tables (de Leva, 1996) to determine the segmental characteristics such as the segmental moment-of-inertia and weight. The moment at the ankle was determined by contributions from the ground reaction forces and the segment weight with respect to the ankle joint. The knee and hip joint moments were determined by contributions from the joint reaction forces distal to the segment and the segment weight relative to the knee and hip joint, respectively. All analyses were performed using custom-written code in Matlab

(MathWorks Inc., Natick, Massachusetts). As the primary plane of motion is the sagittal plane, I only calculated two-dimensional joint power in that plane.

I performed statistical comparisons using paired t-tests and repeated measures analysis of variance, with a level of significance of $p < 0.05$. Based on physics-based mathematical models, the COM work rate increased by length to the fourth power (Donelan et al., 2002a) and the metabolic cost was proportional to the COM work rate (Donelan et al., 2002b). To test whether our results conformed to the prediction and walking results, non-linear regression assessed whether measured COM work rate and metabolic cost increased proportional to the fourth power of step length. I used linear regression to test if metabolic rate increased in proportion to COM work rate, as would be expected if muscle performed this work at constant efficiency. All fits allowed for different offsets to each subject's data, but required all other parameters to be the same between subjects. r^2 values and 95% confidence intervals indicated the degree and significance of fits. All statistical analyses were performed using Matlab (MathWorks Inc., Natick, Massachusetts). To account for differences in body size, I analyzed all variables in dimensionless form. Dimensionless forms of variable take into account the size of the subject by normalizing for: the height of the subject through leg length (l), the weight of the subject through subject mass(m) and gravity (g).Times were normalized by $\sqrt{l/g}$, frequencies by $\sqrt{g/l}$, forces by mg, velocities by $\sqrt{gl}$, work by mgl, and powers or COM work rates by $\sqrt{m^2g^3l}$. For presentation purposes, I also reported variables in more familiar dimensional units by multiplying each dimensionless variable against the appropriate average normalization factor.

2.3 Results

2.3.1 Similarities between walking transition work and rocking transition work

The patterns of COM work during both forward and backward rocking transitions were similar to step-to-step transitions in walking. Figures 2-3 D and E illustrate the instantaneous COM work rate performed by the trailing and leading legs. During both directions of rocking, the trailing leg positive work rate and the leading leg negative work rate primarily occurred during double support making this phase a good approximation of the transition phase during walking. During forward rocking, the front leg was the leading leg and it did mainly negative COM work during double support, while the back leg was the trailing leg and it did mainly positive work (Figure 2-4 A). The role of the front and back legs switched during backwards rocking—the back leg became the leading leg and the front leg became the trailing leg. Like forward rocking, the leading primarily performed negative work and the trailing leg primarily performed positive work during backwards rocking (Figure 2-4 C). Most of the work during rocking was concentrated in double support with only 33% of the positive work and 35% of the negative work, on average, being performed during single support.

While the roles of the leading and trailing legs were similar in backwards and forwards rocking, they performed different amounts of work depending upon rocking direction (Figure 2-4 A). The trailing leg performed, on average, 183%

more double support positive work during forwards rocking than backwards rocking ($p= 1.5E-13$). Similarly, during backwards rocking the leading leg performed 161% more negative work than during forwards rocking ($p= 1.2E-14$) (Figure 2-4 C). Consequently, there was net positive work performed during the forward rocking transition and net negative work during backwards rocking. At the longest length, for example, the net work performed during double support was +12 ± 1.7 J for forwards rocking and -11 ± 2.3 J during backwards rocking. Subjects partially compensated for net work during transitions by performing more negative than positive work during the single support phases that followed the forward rocking transition and more positive work than negative work during the single support phases that followed the backward rocking transition (Figure 2-4 B and D).

2.3.2 Effect of step length on the centre-of-mass mechanical work and on metabolic cost

Average COM work rate depended strongly on the step length (Figure 2-5 A). The average COM work rate increased from 0.16 ± 0.03 W/kg at 60% l to 0.34 ± 0.05 W/kg at 100% l, equating to a 113% increase in work rate over a 71% increase in step length. The rocking model prediction that average COM work rate will increase proportional to the fourth power of step length described the measured relationship reasonably well ($r^2 = 0.92$). COM work appeared to exact a proportional metabolic cost (Figure 2-5 B). Similar to average COM work rate, metabolic cost increased by 116% over the lengths tested—from 1.7 ± 0.42 W/kg at 60% l to 3.7 ± 0.56 W/kg at 100% l. The relationship between metabolic cost and step length was also well fit by a fourth power relationship (r^2 = 0.89).

2.3.3 Relationship between centre-of-mass mechanical work and metabolic cost

Figure 2-6 illustrates a strong positive correlation between metabolic cost and average COM work rate (p= 4.0 E-14, r^2=0.87). The slope of the best-fit line, 0.08 ± 0.01, is a measure of the efficiency of transition work in rocking. The efficiency of walking transitions has been estimated to be 10% (Donelan et al., 2002a). Our result is less than the efficiency of walking efficiency but is within the expected range.

2.4 Discussion

The COM work rate patterns exhibited by rocking and walking are remarkably similar. During rocking transitions, the leading leg does negative work, the trailing leg does positive work, and the COM work rate and metabolic cost increased with increasing step length. I also found that the metabolic cost was correlated with COM work during transitions. These features measured in our novel rocking paradigm match the measured biomechanical and metabolic characteristics of step-to-step transitions during walking (Donelan et al., 2002a, 2002b). Further evidence to support the use of our rocking paradigm to isolate step-to-step transitions comes from comparing the COM work rate profile plots of rocking (Figure 2-7 A) and walking (Figure 2-7 B). During forward rocking, the front leg performed positive work while the front leg performed negative work. The function of the legs switched with backward rocking – the front leg now performed positive work to drive the body in a backward direction while the back leg now performed negative work to redirect the COM velocity. The similarities in the function of the legs between rocking and walking support the use of rocking as an experimental paradigm that isolates step-to-step transitions.

While rocking transitions mimics the characteristics of walking transitions there are fundamental differences between the two. First, the COM in rocking does not proceed forward as in walking. Rocking instead requires the velocity of the COM to stop and reverse directions twice for each rocking cycle. As a result, the COM velocity at the beginning of the transition is lower in rocking than walking when comparing the two at the same step length and frequency. Because of the reduced COM velocity, the step-to-step transition work is reduced in rocking when compared to walking, all else being equal. For example, our subjects performed rocking at the longest length with a length and frequency roughly equal to that used by people walking at 1.75 m/s. Comparing these two conditions, the COM velocity at the beginning of the transition in rocking was only 38% of that observed in walking and the COM work in rocking was only 36% of the walking work.

A second difference between rocking and walking is that rocking includes a forward and a backward phase while walking, of course, only includes the former. While I could separately analyze the mechanics of the two phases, accurate measures of metabolic cost require averaging data over minutes preventing us from teasing out the cost of the forward and backward phases. It is reasonable to assume that backward rocking is considerably less practiced than forwards rocking and this may have contributed to a lower observed efficiency of 8%), compared to the predicted efficiency of 10% from walking transitions (Donelan et al., 2002a). The maximum efficiency of muscle work in walking, assuming that muscles have to perform an equal amount of positive and negative work, is approximately 21% (Donelan et al., 2002a; Margaria, 1976). A second contributor to the low efficiency is likely a metabolic cost associated with supporting the leg that is off the ground during single support. This cost is due to the muscle force required to counteract the gravitational torque of the hanging leg. It is likely not constant but would instead increase linearly with step length as the hanging leg becomes closer to parallel with the ground. The absence of hip joint power suggests that supporting the hanging leg did not require mechanical work. An increase in metabolic cost with length, but without a concomitant increase in COM work would decrease our estimates of transition work efficiency.

I used the ILM and JPM to estimate external mechanical work. While the ILM captures the transition work compared to traditional combined limb methods of calculating work, it has its own limitations including the inability to distinguish the contributing source of the ILM work and the underestimation of work done outside of transition (i.e. work by the swing leg) (Donelan et al., 2002a). The JPM allows us to better understand the contribution of the individual joints but insight using this method is limited to the elements modelled by inverse dynamics. In our case, I only modelled the ankle, the knee, and the hip joint. This may explain why I was unable to fully account for the shortfall in summed joint power when compared to the ILM method (i.e. from other joints and soft tissues). By analyzing the data using both methods, I gain insight into the specificity of joint contributions using JPM but capture the total transition work using ILM.

To supplement our ILM results, I also ran inverse dynamics analysis to obtain the joint power of ankle (Figure 2-8 D), knee (Figure 2-8 C) and hip (Figure 2-8 B) and summed the powers of the individual joints to obtain a total joint power of the lower limbs (Figure 2-8 A). In this way, I was able to estimate the power contribution of the individual joints. During forward rocking, as in walking, the transition work was dominated by the trailing ankle joint as push-off redirects the COM velocity. Much of the trailing limb work during forward rocking was dominated by the ankle in contrast to the contribution of the knee and the hip joint. This is consistent with literature attributing the ankle joint power as the major contributor of forward propulsion and push-off (Lewis & Ferris, 2008; Olney *et al.*, 1991). In forward rocking, the leading leg ankle and the knee contribute to the majority of negative work performed. However, the summed contributions of the individual joints did not add up the overall leg function as measured by the ILM (Figure 2-8 A). This indicates that perhaps there are other unquantified contributions -- trunk motion or the metatarsal joint of the foot may explain the shortfall of the magnitude of JPM. Another explanation for why the summed joint power do not add up to ILM power is energy dissipation by passive sources including the sole of the shoe and soft tissue (Pain, 2001, 2002; Stefanyshyn, 2000).

The joints responsible for energy generation in transition differed between forward rocking and backward rocking. While the trailing leg, only the ankle generated most of the power in forward rocking, both the ankle and the knee generated the majority of the power in backward rocking. Perhaps such differences can be explained by the role of the ankle. The role of the ankle may explain why net positive work was done during forward support while net negative work was done during backward double support. While the back ankle is adept at plantarflexion to produce the majority of the positive work during forward transition, the joints of the front hip, knee and ankle appears to be less adept at performing positive work (Figure 2-8 B-D) going backward. The inability of the front leg to produce positive work causes sub-optimal transitions resulting in net negative work in backward double support. As a result of having net negative work during double support, net positive work must be perform in forward double support to ensure that no net work is performed. No net work is required because the task is constrained (i.e. using metronome) such that over one gait cycle, the average COM velocity is neither speeding up nor slowing down. While rocking transitions appear to be more sub-optimal than walking transitions(Donelan et al., 2002b), where near optimal transitions were performed, the mechanics and energetics of performing rocking transitions, as set out by out hypotheses, are true for both rocking and walking.

The purpose of this study was to assess whether sagittal plane rocking isolates step-to-step transition from other contributors of the metabolic cost of walking. Sagittal plane rocking appears to isolates transitions while still preserving the mechanics and energetics of walking transitions. Since step-to-step transition is a major determinant of the metabolic cost of walking, making up about 60-70% of the total cost of walking (Kuo et al., 2005) the ability to isolate and study transitions can further our understanding into the biomechanical determinants of the metabolic cost of walking. The ability to isolate transitions allows for in-depth study into step-to-step transition. For example, this paradigm allows us to further investigate predictions from physics-based mathematical models predicting that the inability to perform work in a coordinated manner and timing incurs greater COM work resulting in greater metabolic cost (Chapter 3). By isolating transitions, researchers can also use the rocking paradigm to investigate the function of muscles during transitions to gain insight into which muscles redirect the COM velocity. Understanding the muscle function of transition and biomechanics of transition work may help us gain insight into step-to-step transitions in healthy and pathological gait.

2.5 Figures

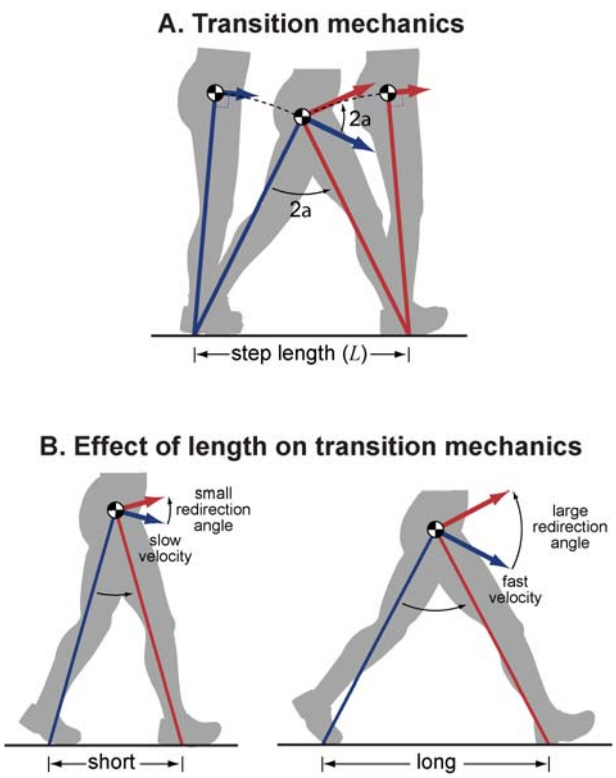
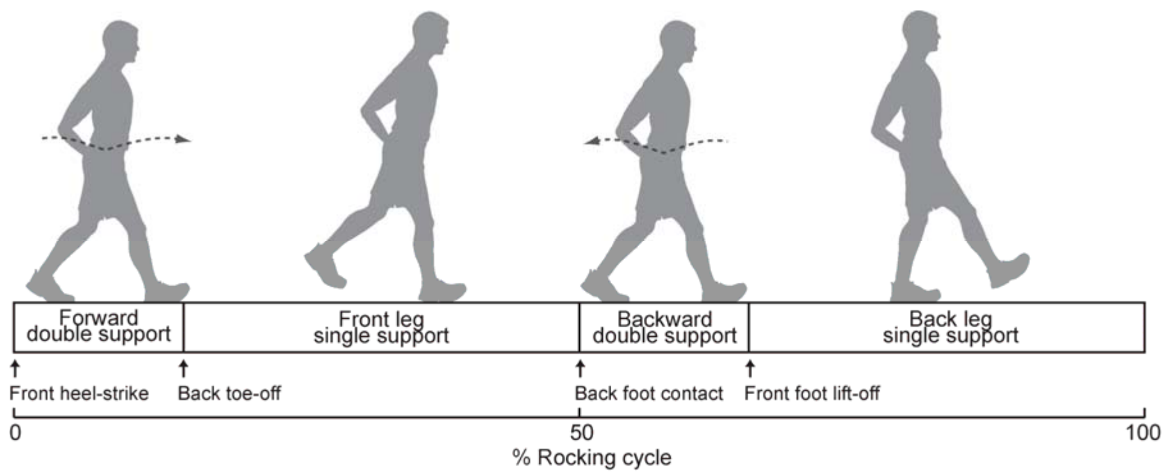


Figure 2-1. Inverted pendulum model of walking and effect of step length on mechanical work. (a) The walking cycle alternates between single support and double support phases. A consequence of the stance leg behaving like an inverted pendulum is the need to transition between the inverted pendulum of the trailing leg and the next inverted pendulum of the leading leg. The COM velocity direction must be redirected from a downward directed trajectory (blue arrows) to the upward trajectory of the next arc (red arrows). (b) Transition work depends strongly on step length during walking. Longer step lengths have larger associated com velocities and larger angle of redirection requiring more work to redirect the COM velocity.

A



B

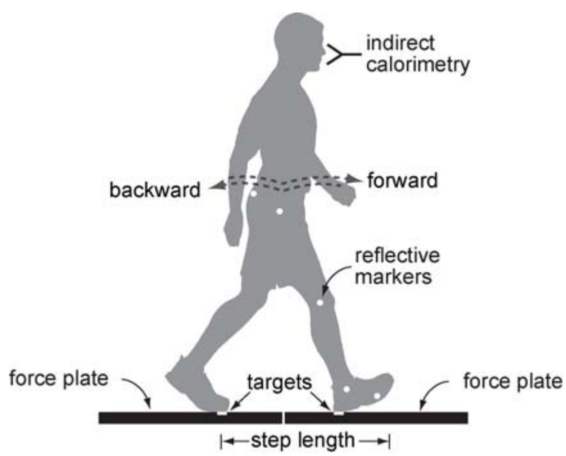


Figure 2-2. Rocking paradigm. (a) The rocking cycle is composed of four phases. Forward double support begins with front heel-strike and ends with back toe-off. During front leg single support, the COM travels forward then backwards. Backward double support commences with back toe-strike and ends with front heel-off. During back leg single support, the COM travels backward then forward before the rocking cycle begins again with front heel-strike. (b) I studied sagittal plane rocking to determine whether rocking isolates step-to-step transitions. Subjects rocked over two force plates while O_2 consumption and CO_2 production was measured with mouthpiece and metabolic analysis unit. Targets on force plates marked the desired step length while reflective markers captured joint position.

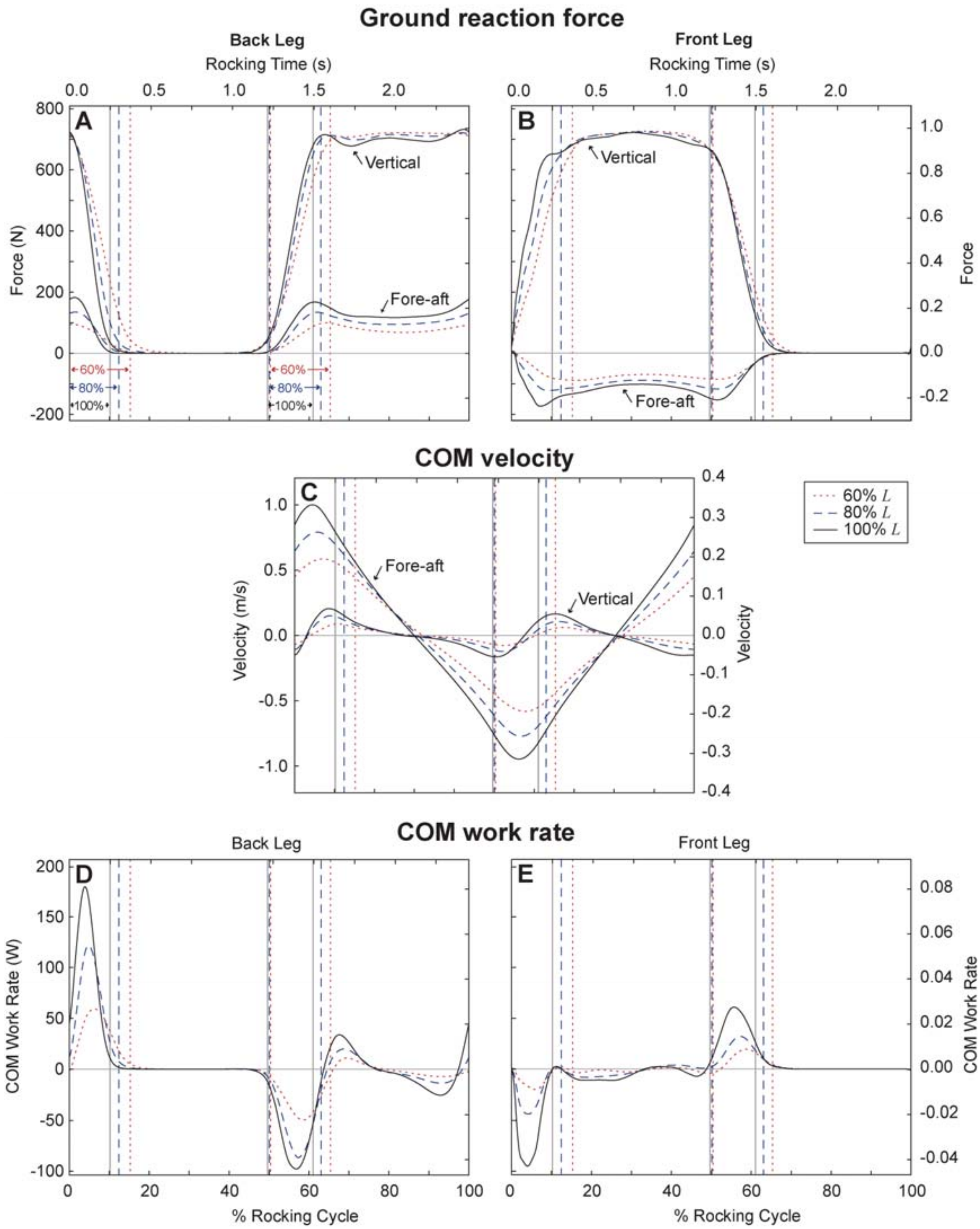


Figure 2-3. The ground reaction force, velocity and COM work rate (n=10) at the shortest, intermediate and longest rocking length. The dot product of ground reaction force (GRF) and COM velocity yielded the rate of COM work. Average GRF of the back force plate (a) and front force plate (b), average COM velocity (c) and average rate of COM work of the back leg (d) and front leg (e) for one rocking cycle (60%, 80%, 100% leg length; N = 10). Vertical lines dotted represent the double support period for different step lengths.

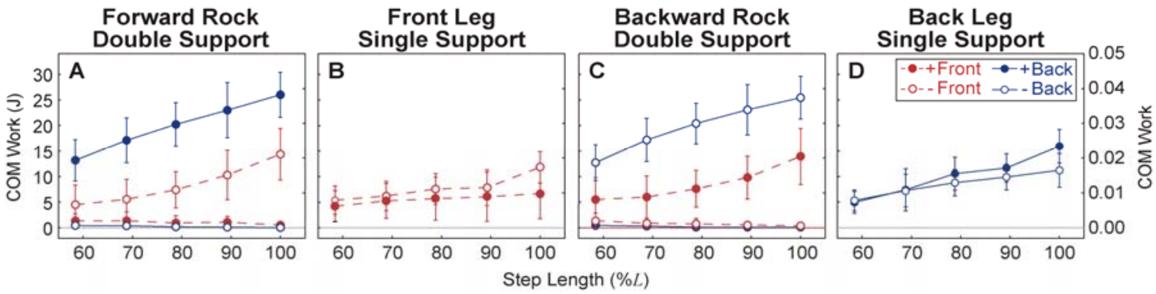


Figure 2-4. Positive and negative work (n=10) performed in chronological phases of the rocking cycle. Forward double support (a) and front leg single support (b) of the first half of the rocking cycle is followed by backward double support (c) and back leg single support (d) . For each phase and each direction of rock, I show the: the positive work by the front leg (W+ front), the absolute value of the negative work by the front leg (W- front), positive work by the back leg (W+ back) and the absolute value of the negative work by the back leg (W- back). For each rocking length, the leading leg did negative work and the trailing leg did positive work during transitions.

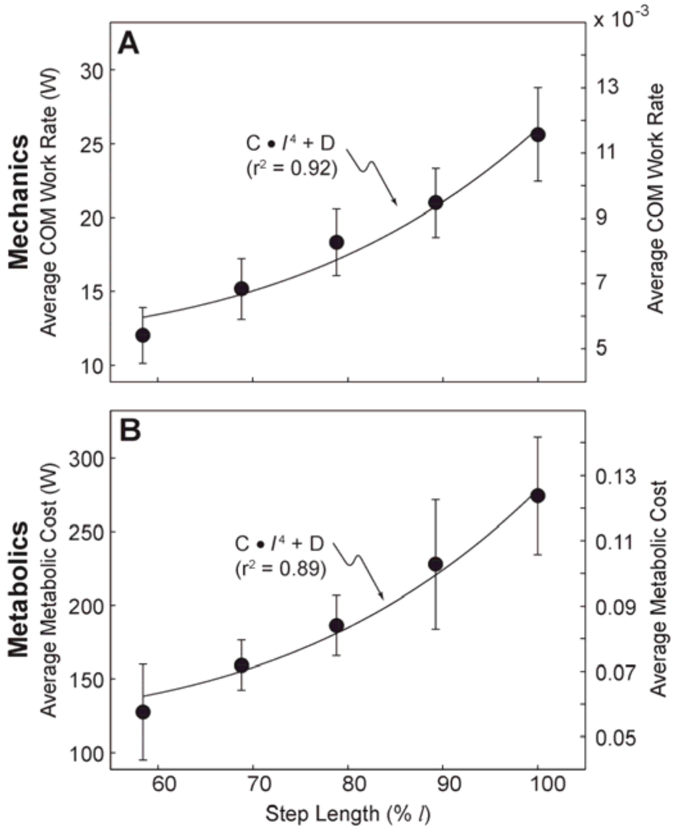


Figure 2-5. COM work and metabolic cost (n=10) over different step lengths. (a) The mean of the total positive COM work increased with increasing step length. The rate of COM work is proportional to the fourth power of step length (r^2 = 0.92). (b) The metabolic cost is proportional to the fourth power of step length (r^2 = 0.89).

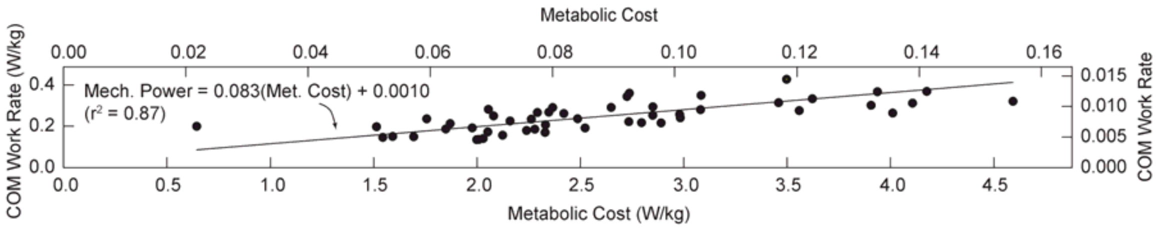


Figure 2-6. Linear regression equation between COM work rate and metabolic cost. Each point represents each subject's (n=10) averaged dependent and independent variable. The metabolic cost is correlated with COM work with the slope of the regression line or efficiency (efficiency = COM work rate/metabolic cost) of .083 or 8.3 percent (r^2 = 0.87).

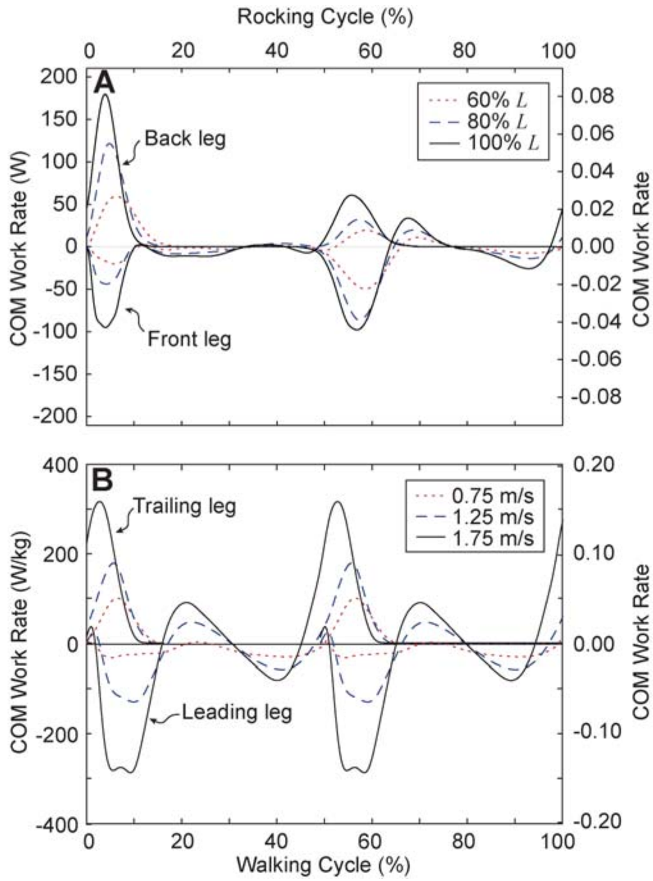


Figure 2-7. Comparison of rocking and walking COM work rate. COM work rate profile of one cycle of (a) rocking (n=10) and (b) walking for one subject at the intermediate step length. In walking, one stride is made up of two steps, while one cycle of rocking is a forward rock and backward rock. The work rate generated from the push-off and collision phases of walking and rocking have similar COM work rate profiles supporting the use of sagittal plane rocking to isolate transitions. Walking COM work rate was adapted from Donelan et al., 2002a.

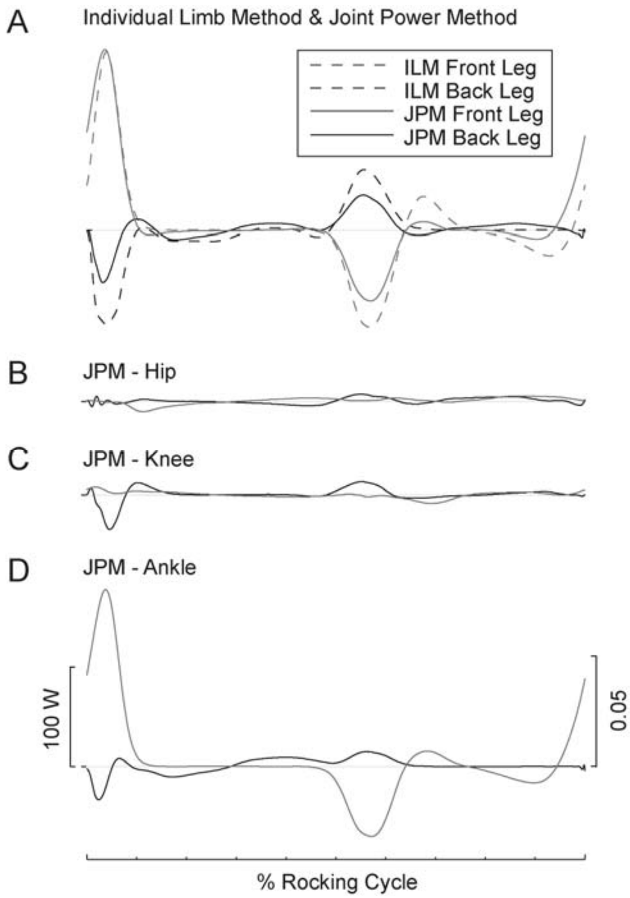


Figure 2-8. Comparison of ILM COM work rate and joint powers (n=10) from JPM analyses. (a) ILM COM work rate is compared to the sum of individual joints powers (JPM Front leg; JPM Back leg) which is the sum of the instantaneous (b) hip, (c) knee, and (d) ankle joint powers.

CHAPTER 3: SUB-OPTIMAL TRANSITIONS APPEAR TO INCREASE THE CENTRE-OF-MASS WORK RATE AND METABOLIC COST

3.1 Introduction

Pathological gait has an elevated metabolic cost compared to healthy walking. The metabolic cost of pathological gait whether from hemiparesis, amputation or ankle arthrodesis can be as much as twice that of able-bodied walking (Waters *et al.*, 1988; Waters & Mulroy, 1999). Minimizing the metabolic cost of walking is important because the economy of ambulating affects the distance a person can travel, the loads they may carry which in turn affects the quality of daily activities of individuals (Thijssen *et al.*, 2007). Consequently, an often stated functional goal of walking rehabilitation is to reduce the metabolic cost of walking (Platts *et al.*, 2006; Thijssen et al., 2007). The motivation for this study is to gain insight into biomechanical determinants, which may account for increased metabolic cost observed in gait pathologies.

In healthy walking, the body behaves much like an inverted pendulum during single stance. During this phase, the centre-of-mass (COM) moves along an arc dictated by the stance limb and the COM velocity is approximately perpendicular to the limb. As a result of behaving like an inverted pendulum, the COM velocity must be redirected, during step-to-step transitions, from the pendular arc of the trailing leg to the pendular arc of the leading leg (Figure 3-1 A). In healthy people, the combined action of the trailing and leading legs accomplish this redirection, with the leading leg performing negative work during its collision phase and the trailing limb performing positive work during its push-off phase (Donelan et al., 2002a, 2002b). This transition work exacts a proportional metabolic cost (Donelan et al., 2002a) and accounts for about 60-70% of the total metabolic cost of walking in healthy people (Kuo et al., 2005).

Physics-based mathematical models demonstrate that the timing and the magnitude of transition work performed on the COM are important determinants of the total COM mechanical work. In principle, the COM velocity could be redirected entirely with leading leg negative work with the positive work required to replace the dissipated energy being performed later in the walking cycle (Figure 3-1 B). It is advantageous, however, to use the trailing leg to perform positive work to assist the redirection because it requires less leading leg negative work to complete the redirection and thus less total COM mechanical work. Transition work is minimized when the trailing and leading legs simultaneously perform equal magnitudes of positive and negative COM work. Healthy humans perform near optimal transitions at all walking speeds (Donelan et al., 2002a).

Gait pathologies may affect the ability of one or both legs to perform transition work with the optimal timing and magnitude. Consider a patient with one affected leg, perhaps through stroke or amputation. If the affected leg is trailing during the transition, it may not be able to provide the desired push-off positive work. Physics-based models predict that the negative work by the leading leg during this transition will be as much as four times greater than if the transition was performed optimally (Kuo et al., 2005; Ruina et al., 2005). Positive work must be performed at some point to replace the dissipated energy and maintain walking speed. Were the patient to wait until the next transition, the healthy trailing leg would need to perform four times the positive work required in an optimal transition. While the COM velocity would be entirely redirected by the action of the trailing leg requiring no negative work by the leading leg during this transition, four-times the positive work in one transition and none in the other is still a two-fold increase in total COM work over a stride where both transitions were performed optimally. Were the patient to perform the needed work during single support rather than wait for the next transition, they would need to perform even more total COM work. While the capacity of an affected leg in performing positive work is often greater than zero, any reduction in its ability to perform positive work with the optimal timing and magnitude is predicted to result in an increase in COM work, albeit less than the two-fold example of above. Assuming a proportional relationship between COM work and metabolic cost, this optimal transition hypothesis may explain why pathological walking is more metabolically expensive than healthy walking.

The purpose of this study was to gain insight into the biomechanical determinants of the elevated metabolic cost of pathological gait. Our general hypothesis was that the elevated metabolic cost of certain pathological gaits may be due to the increased muscle mechanical work required of sub-optimal step-to-step transitions. To test our hypothesis, I first isolated step-to-step transitions from other contributors to the metabolic cost of walking using sagittal plane rocking (Chapter 2). I previously demonstrated that this rocking paradigm retains the mechanics and energetics characteristic of walking step-to-step transitions (Donelan et al., 2002a, 2002b). I simulated gait pathology by imposing sub-optimal transitions on healthy humans by immobilizing the ankle of the front or back leg thereby reducing the ability of the muscles that cross the immobilized joints from performing positive work on the body. This biomechanical simulation of pathological gait provides insight into the relationship between sub-optimal transitions and metabolic cost. Because only healthy subjects were used, I eliminated the need to control for the effects of pathologies, such as reduced muscle mass and spasticity (Bard, 1963).

3.2 Materials and methods

3.2.1 Experimental Procedures

I measured the COM work and metabolic costs of rocking as a function of ankle immobilization (Figure 3-2 A) in 8 healthy subjects (7 males and 1 female; age = 28.0 ± 3.6 years; leg length = 0.92 ± 0.04 metres; mean ± standard deviation). Prior to testing, all subjects gave their written informed consent in accordance with the Simon Fraser University Office of Research Ethics.

I previously determined that the sagittal plane rocking paradigm isolates transitions rock while retaining the mechanics and energetics characteristic of walking transitions. Rocking consists of a series of rocking cycles with the motion restricted primarily to the sagittal plane. Each rocking cycle is comprised of a forward and backward half-cycle. Each rocking cycle has two transitions with the front leg leading during the forward half-cycle and the back leg leading during the backward half-cycle. I defined each new cycle as beginning at foot contact of the front leg.

In all trials, subjects wore commercially available ankle braces (ProGait ST Boot, Bledsoe Brace System, Grand Prairie, Texas), lockable at 10-degree increments while allowing 20 degrees of dorsiflexion and 40 degrees of plantarflexion in the free range of motion setting. While the ultimate goal of our project is to understand the underlying mechanics that may cause elevated cause in pathological walking, I am not trying to simulate any particular pathological condition. Instead, the purpose of the brace is to cause the subjects to perform sub-optimal transitions so that I may observe the associated metabolic cost.

To biomechanically impose sub-optimal transitions in the back leg, I immobilized the back leg ankle at a neutral (0°) angle while allowing full range of brace motion in the front leg ankle. Conversely, to impose sub-optimal transition in the front leg, I immobilized the front ankle at 20 degrees of plantarflexion while allowing full range in the back ankle. I compared these immobilized treatments to the control trials where both front and back ankles have full brace range of motion. To summarize, the three braced conditions are: 1) both ankles free (control), 2) front ankle locked (back ankle free) and 3) back ankle locked (front ankle free).

In the locked position, the rigid ankle braces appeared to effectively limit ankle joint motion. Figure 3-2 B illustrates a substantial reduction in the ankle angular velocity of the front ankle when the front ankle is locked compared to the control trial. When the front ankle was immobilized, the peak positive and negative angular velocities were only 9% and 18%, respectively, of the peak velocities in the control condition. Figure 3-2 C illustrates a marked reduction in the ankle angular velocity for the back ankle when the back ankle is locked compared to the control trial. With the back ankle locked, the peak positive velocity was 28% while the peak negative velocity was 21% of the peak velocities in the control condition.

To account for difference in body size, subjects rocked at 80% of their leg length while keeping rocking frequency constant. The length was enforced by asking subjects to contact two pieces of tape, placed the correct distance apart, at the front foot heel and back foot toe (Figure 3-2 A). I enforced rocking frequency by asking subjects to match their foot contacts to the beat of a metronome during both the forward and backward half-cycles, forcing the half-cycles to have equal duration. The metronome was set to a dimensionless frequency of 0.50 equating to a frequency of 0.25 for the full rocking cycle. For an average subject, this equated to 50 rocking cycles per minute. To cue the subjects to change rocking directions, the metronome sounded at 50% and at 100% of each rocking cycle. I explored other rocking frequencies in pilot experiments and settled on this frequency because it produced large amplitude rocks without requiring the subjects to be stationary during support phases. On a day prior to the day of testing, I familiarized the subjects with rocking in braces while keeping rocking frequency and step length fixed. I randomized the order of immobilization and subjects switched between having their left or right leg as the front leg after each trial. I carried out post-hoc data analyses to assess whether subjects rocked at the desired step lengths and frequencies.

For each subject and at each bracing condition, I measured the ground reaction forces, joint kinematics and the metabolic cost of rocking. The individual limb ground reaction forces were measured using two force plates—one under the front leg and another under the back leg (Bertec Corp., Columbus, Ohio). Signals were collected from both force platforms simultaneously at 960 Hz per channel.

I used a 4^{th} order, recursive, zero-phase-shift low-pass Butterworth filter with a cut-off frequency of 25 Hz to condition the ground reaction force signals. For kinematic data, I recorded the motion of 12 markers at 120 Hz with an 8-camera motion capture system (Vicon Motion Systems, Oxford Metrics Limited, Oxford, England). Markers were placed bilaterally on the fifth metatarsal of the foot, the lateral malleoli, the lateral epicondyles of the knee, the greater trochanters, and on the sacrum. Additionally, a secondary marker was placed on each thigh and shank segment to facilitate marker identification. I used a 4^{th} order, recursive, zero-phase-shift Butterworth filter with a cut-off frequency of 6 Hz to condition the kinematic data. To determine metabolic cost, I measured and calculated average oxygen consumption ($\dot{V}_{O2}$) and carbon dioxide production ($\dot{V}_{CO2}$) using a metabolic analysis system (Vmax Encore, SensorMedics Corp., Yorba Linda, California). The metabolic cost was estimated by using the standard equation (Brockway, 1987):

Metabolic Cost = 16.58 ($\dot{V}_{O2}$) + 4.51 ($\dot{V}_{CO2}$).

To ensure that a steady state in rocking had been reached, I averaged only the last three minutes of measured respiratory gases and the last one minute of measured ground reaction forces for each six-minute trial. Prior to rocking trials, I determined each subject's resting metabolic cost during standing.

3.2.2 Data Analyses

I used the measured ground reaction forces and the individual limbs method (ILM) to calculate the COM work and COM work rate (Donelan et al., 2002b). Briefly, the rate of work performed on the COM by each leg is defined as the dot product of the individual limb ground reaction forces and the COM velocity. To determine COM velocity, I first calculated the acceleration of the COM from the vector sum of the individual limb ground reaction forces and gravity and then integrated the COM acceleration with respect to time. Because rocking requires zero displacement over a complete cycle time, the integration constants were determined by forcing the average vertical, fore-aft and mediolateral COM velocity to equal zero for each rocking cycle to control for integration drift. I determined COM work from the cumulative time-integral of the instantaneous COM work rate with positive and negative COM work determined by restricting the integration to intervals over which the instantaneous COM work rate was positive or negative, respectively. Similarly, I calculated COM work for a full rocking cycle by integrating over the complete cycle and for phases of the rocking cycle by integrating only over those phases (e.g. double support). The reported measures of COM work are averages across subjects and I determined each subject's values by first averaging across rocking cycles during the last minute of each six-minute trial. I defined average COM work rate as the mean positive COM work divided by the mean rocking cycle period.

I used inverse dynamics or the joint power method (JPM) to determine the ankle, knee and hip joint power during rocking. Joint power was calculated by taking the product of the angular velocity of that joint and the net joint moment. For each joint, the angular velocity was determined by differentiating the joint angle with respect to time. Markers on the lateral epicondyle of the knee, the lateral malleoli and the fifth metatarsal form the ankle joint angle. Markers on the greater trochanter, the lateral epicondyle of the knee and the lateral malleoli, formed the knee joint angle. Markers on the sacrum, the greater trochanter and the lateral malleoli formed the hip joint angle. To determine the segments anthropometrics, I used standard tables (de Leva, 1996). All analyses were performed using custom-written code in Matlab (MathWorks Inc., Natick, Massachusetts). As the primary plane of motion is the sagittal plane, I only calculated two-dimensional joint power in that plane. I used the instantaneous sum of the individual joint powers to calculate the instantaneous leg power.

I performed our statistical comparisons using paired t-tests and repeated measures analysis of variance, with a level of significance of $p < 0.05$. All statistical analyses were performed using Matlab (MathWorks Inc., Natick, Massachusetts). To account for differences in body size, I analyzed all variables in dimensionless form. Times were normalized by $\sqrt{l/g}$, frequencies by $\sqrt{g/l}$, forces by mg, velocities by $\sqrt{gl}$, work by mgl, and powers or COM work rates by $\sqrt{m^2g^3l}$, where m is body mass, g is gravitational acceleration, and l is leg length. For presentation purposes, I also reported variables in more familiar dimensional units by multiplying each dimensionless variable against the appropriate average normalization factor.

3.3 Results

3.3.1 Role of ankle immobilization on the centre-of-mass mechanical work

Limiting an ankle joint reduced that leg's COM work rate and increased the work rate of the contralateral leg. Consider first the condition where the back ankle is immobilized. The COM work rate patterns shown in Figure 3-3 illustrate that this immobilization tended to restrict the ability of the trailing back leg to perform positive push-off work during the forward rocking transition resulting in a larger collision phase by the non-immobilized leading front leg. This pattern is somewhat supported by Figure 3-4 which quantifies the COM work during the various phases of the walking cycle. Immobilizing the back leg reduced its positive push-off work during double support by 32% (p= 7.0E-5) however the negative collision work by the leading leg increased slightly but was not significant (Figure 3-4 A). The reduction in positive work did not appear to be replaced during the subsequent single support—there was a non significant increase in positive and negative work during this phase, when compared to the control condition, but in fairly equal amounts suggesting that the net single support work is near zero (Figure 3-4B). Positive work did increase non-significantly during the backward transition when the non-immobilized front leg was trailing and could perform additional push-off work (Figure 3-3). As a consequence of the larger push-off, the immobilized back leg required 32% less negative work to complete the redirection of the COM velocity when compared to the control trial (p= 1.7 E-5) (Figure 3-4 C).

Immobilizing the front leg had similar functional effects to immobilizing the back leg (Figure 3-3). To understand these effects, it is instructive to begin with the backward rocking transition when the immobilized leg is trailing. During backward rocking, immobilizing the front ankle limited the ability of the front leg to do positive work while the collision work of the leading leg increased by a notable but not statistically significant 24% (Figure 3-4 C). The reduction in positive work did not appear to be replaced during the subsequent back leg single support—there was a non-significant decrease in positive and negative work during this phase, when compared to the control condition, but in fairly equal amounts, again, suggesting that the net single support work is near zero (Figure 3-4 D). Instead, the reduction of positive work appears to be compensated during forward double support by the non-immobilized back leg where 25% more work was performed (p= 0.05) (Figure 3-4 A) while collision work by immobilized front leg was slightly reduced (n.s.) (Figure 3-3). The COM work during the subsequent immobilized leg single support phase was similar to the control condition (Figure 3-4 B). While I did not obtain statistically significant differences for every COM work comparison between the control and immobilized trials, the general patterns indicate that immobilizations successfully induced sub-optimal transitions by limiting the positive work produced by the immobilized leg while increasing the negative work produced by the contralateral leg.

3.3.2 Effect of sub-optimal transitions on the centre-of-mass mechanical work and on metabolic cost

Sub-optimal transitions tended to increase in the total average COM work rate. Figure 3-5 illustrates the COM work rate and the metabolic cost of our immobilization conditions compared to the control condition. Immobilizing the front ankle increased the COM work rate from 16.3 ± 2.7 W/kg of the control trial to 18.1 ± 3.7 W/kg, equating to a 10% increase (p= 6.0E-3). A similar increase in the metabolic cost was observed – locking the front leg yielded a 15% increase (p= 8.0 E-3) in the metabolic cost from 178.0 ± 25.5 W/kg of the control trial to 208.2 ± 36.3 W/kg. The average efficiency of the immobilized front ankle condition is 8.8 ± 1.3%, similar to the previously estimated efficiency during length modulated walking (Figure 2-6). No statistical differences were found in the COM work rate or metabolic cost between the control condition and the back leg immobilized condition.

3.4 Discussion

Locking the front ankle successfully immobilized the front leg and caused sub-optimal transitions. During backward rocking, immobilizing the front ankle limited the ability of the front leg to do positive work while increasing the collision work of the leading leg. This reduction of positive work appears to be compensated during forward double support by the non-immobilized back leg while tending to reduce collision work by immobilized front leg. As a result of performing sub-optimal transitions, the total COM work rate and the resulting metabolic cost was elevated - supporting the optimal transition hypothesis.

In the similar way that the sub-optimal from locking the front leg resulted in elevated COM work and metabolic cost, I also expected locking back condition to yield the same results. It was surprising that I could not find any significant differences in the COM work and metabolic cost when the back leg was locked because, in general, the ankle contributes substantially to the forward propulsion compared to other leg joints (*Observational gait analysis*, 2001). To understand our results, I further investigated the actual frequency and step length the subject rocked. I conducted post-hoc analyses and found that when compared to controls, there was no differences in the rocking frequency for both conditions ($p = 0.99$) and the rocking length when the front ankle was locked was not different from the control condition ($p = 0.07$). The step length when the back leg was locked was significantly shorter ($p = 8.4E-3$) than the control trial by 4% percent resulting in an 8% decrease in the COM work. While this amount may seem small, the non-linear relationship between step length and the COM work may partially explain why the COM work rate for our back ankle locked condition was not significant. Using the non-linear regression equations derived from Chapter 2 (Figure 2-5 A), a reduction of 4% percent in the step length represents a 6% decrease in the COM work. The reduction in rocking length may partially explain the decrease in COM work.

Another possible explanation, which may help explain our findings for the back ankle locked condition, is the presence of a rocker bottom on our braces

(Figure 3-2 A inset). Vanderpool et al. (2008) found that ankle fixation did not necessarily increase the metabolic cost of walking and attributed this to the rocker shape bottom of the boot. The braces I used had a rocker boot sole designed for forward propulsion. Thus by locking the back ankle, a subject may take advantage of the shape of the rocker bottom during the forward phase of rocking thereby masking the back leg locked treatment by limiting the COM work and resulting metabolic cost. Since the brace was not designed for backward rocking, no advantage could be gained by the subject while rocking that direction. This allowed us to see differences when the front ankle was locked. Locking the back ankle did not increase the COM work rate nor the metabolic cost. While this is unfortunate, this finding does not nullify our hypothesis that suboptimal transitions are associated with increased COM work rate and metabolic cost. This finding suggest that future studies eliminate a rocker bottom from braces when the trying to impose suboptimal transitions by locking the ankle.

While the individual limb method(ILM) measures all the work done on the COM without attributing it to joints, inverse dynamics analysis provides insight into the work of joints – but only the joints modelled in the analysis. These two methods of analysis are complementary as I gain the overall big picture of total COM work using the individual limbs method but can still attribute the work to specific joints using inverse dynamics analysis. As such, I also performed inverse dynamics analysis to obtain the joint power of ankle (Figure 3-6 E), knee (Figure 3-6 D) and hip (Figure 3-6 C) and summed the powers of the individual joints to obtain a total joint power of the lower limbs (Figure 3-6 B). This analysis has two purposes. First, it allows us to further gain insight into the effect of the immobilization. Second, this analysis allows us to estimate the relative joint power contribution of the aforementioned joints in comparison to the ILM results (Figure 3-6 A).

The success of immobilization was assessed when comparing the summed joint powers (Figure 3-6 B) to the ILM COM work rates (Figure 3-6 A). Hypothetically, if the both figures were identical, the sum of COM work rate of the three joints accounted for the entire COM work rate in ILM. While the patterns of ILM COM work rate and joint powers are similar, there are some marked differences. The most profound difference is the peak positive work when the back leg is locked during the first double support. The summed joint peak power was only 53% of the ILM trailing leg peak COM work rate. There are two explanations why there may be a difference. First, there may be movement within the brace. The reliance of the joint power method on marker position supports the idea that while the brace itself is not moving, the movement inside the brace can still do work which is measure by the individual limb method. The second explanation is that other joints or muscles, such as the trunk, are doing substantial work while not being accounted for by our inverse dynamics analysis of the hip, knee and ankle joint.

The second purpose for performing inverse dynamics analysis is to gain insight into the distribution of work over various joints. The other joints from the immobilized leg do not compensate for the immobilized joints. In particular, the back leg knee or hip does not perform more positive work when the back ankle was fixed during forward double support (Figures 3-6 C, D and E). Similarly, the knee and the hip of the front leg do not do more work with the front leg is locked.

As a comparison to pathological walking, where the metabolic cost can be twice that of their healthy counterparts in the extreme case, our study only modestly increase the sub-optimal metabolic cost by 15% compared to the control condition. I attribute this modest increase to movement within the brace and an intermediate rocking length (i.e. 80% leg length). To increase the mechanical work resulting in increased metabolic cost, future studies should increase the work required by rocking at a longer length and ensuring proper brace fit, thus limiting movement within the brace. Nevertheless, I was able to simulate sub-optimal transitions by locking the front ankle which resulted in elevate levels of total COM work rate and metabolic cost. This supports the sub-optimal transition hypothesis, which attributes the elevated metabolic cost of walking to the inability to do equal and opposite amounts of push-off and collision work during transitions. While I have found a relationship between sub-optimal transitions, our study did not test this in subjects with pathological walking. By testing in healthy subjects, I controlled for the level and the severity of pathology – which would be hard to control in subjects with gait pathology. Even without testing in a pathological population, our statement relating sub-optimal transitions to gait pathologies is not without evidence. Houdijk et al. (2009) found that there was a proportional relationship between elevated negative transition work and metabolic cost in unilateral transtibial amputees. The same relationship was found also found in gait after total ankle arthroplasty (Doets *et al., In Press*).

These studies strengthen statement that regardless of the etiology of pathological walking, the inability to perform optimal transition work causes increased metabolic demands.

The results of our study suggest that one way to improve the economy of pathological walking is to focus rehabilitation therapy strategies and assistive devices on achieving optimal transitions. Rehabilitation of pathological walking can minimize the metabolic cost of walking on improving the timing and magnitude of transition work. For assistive devices such as orthoses and prostheses, our study suggests that augmenting or improving the patient's work output so that they perform optimal transitions would be beneficial. In particular, researchers are developing the Controlled Energy Storage and Release (CESR) prosthesis foot which stores energy and release the energy at the start of step-to-step transition (Collins, 2005). Wearing the CESR prosthesis yielded a lower metabolic cost of walking compared to a conventional prosthesis. While researchers and engineers have believed the importance of sub-optimal transitions in pathological gait, our experiments are among the first to systematically test the effect of sub-optimal transitions on elevated COM work and metabolic cost.

3.5 Figures

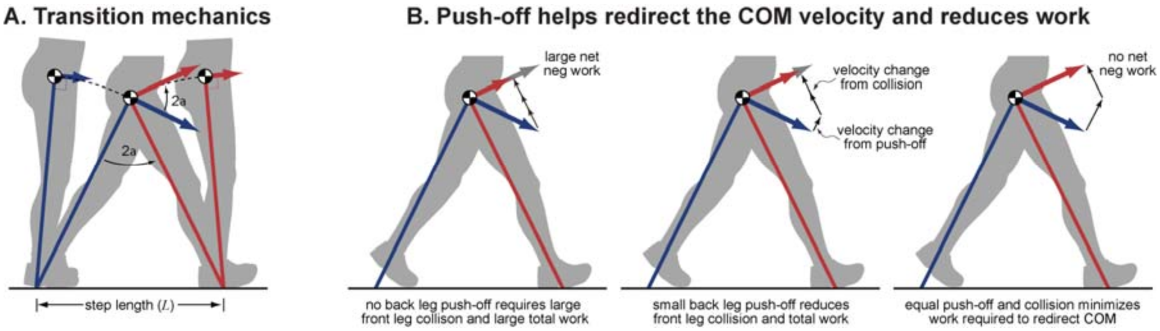


Figure 3-1. Inverted pendulum model of walking and effect of sub-optimal transitions on mechanical work. (a) The walking cycle alternates between single support and double support phases. A consequence of the stance leg behaving like an inverted pendulum is the need to transition between the inverted pendulum of the trailing leg and the next inverted pendulum of the leading leg. The COM velocity direction must be redirected from a downward directed trajectory (blue arrows) to the upward trajectory of the next arc (red arrows). (b) Any reduction in its ability to perform positive work with the optimal timing and magnitude is predicted to result in an increase in mechanical work and resulting metabolic cost.

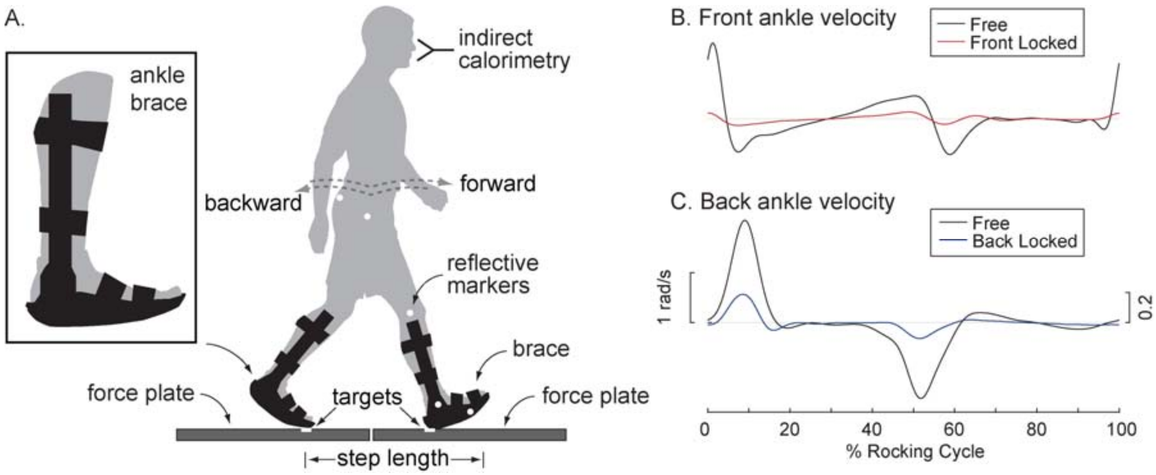


Figure 3-2. Braced rocking paradigm. (a) I studied sagittal plane rocking to determine whether rocking isolates step-to-step transitions. Subjects wore immobilization braces (inset) and rocked over two force plates while O_2 consumption and CO_2 production is measured with mouthpiece and metabolic analysis unit. (b) Immobilizing the front ankle limited the front ankle velocity (n=8). (c) Immobilizing the back ankle limited the back ankle velocity (n=8).

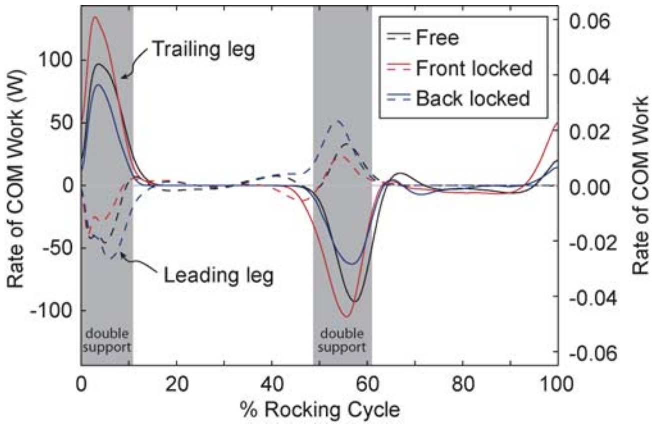


Figure 3-3. Limiting an ankle joint appears to cause sub-optimal transitions (n=8). During forward double support (left shaded region), immobilizing the back leg reduced its positive push-off work (solid blue) and tended to increase the negative collision work (dashed blue) by the leading leg. During backward double support (right shaded region), immobilizing the front leg reduced its positive push-off work (dashed red) during double support and tended to increase the negative collision work (solid red) by the leading leg. Double support region are approximate.

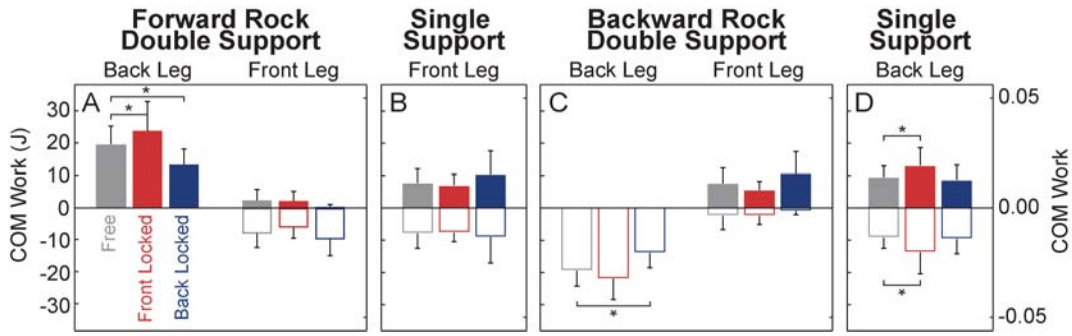


Figure 3-4. Positive and negative work (n=8) performed in chronological phases of the rocking cycle. Forward double support (a) and front leg single support (b) of the first half of the rocking cycle is followed by backward double support (c) and back leg single support (d). The general patterns show that immobilizations successfully induced sub-optimal transitions by limiting the positive push-off work produced by the immobilized leg while increasing the negative collision work produced by the contralateral leg.

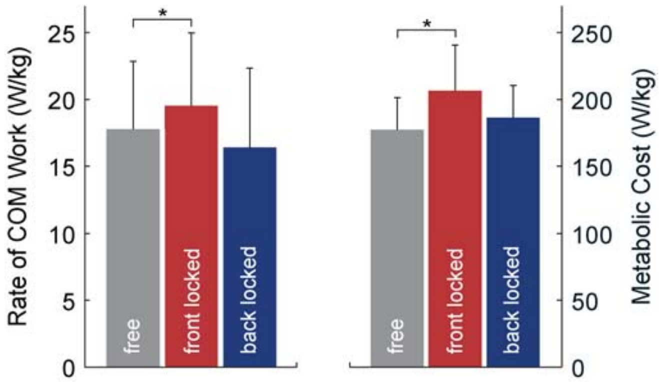


Figure 3-5. Sub-optimal transitions from locking the front ankle increased the total average COM work rate and the resulting metabolic cost (n=8).

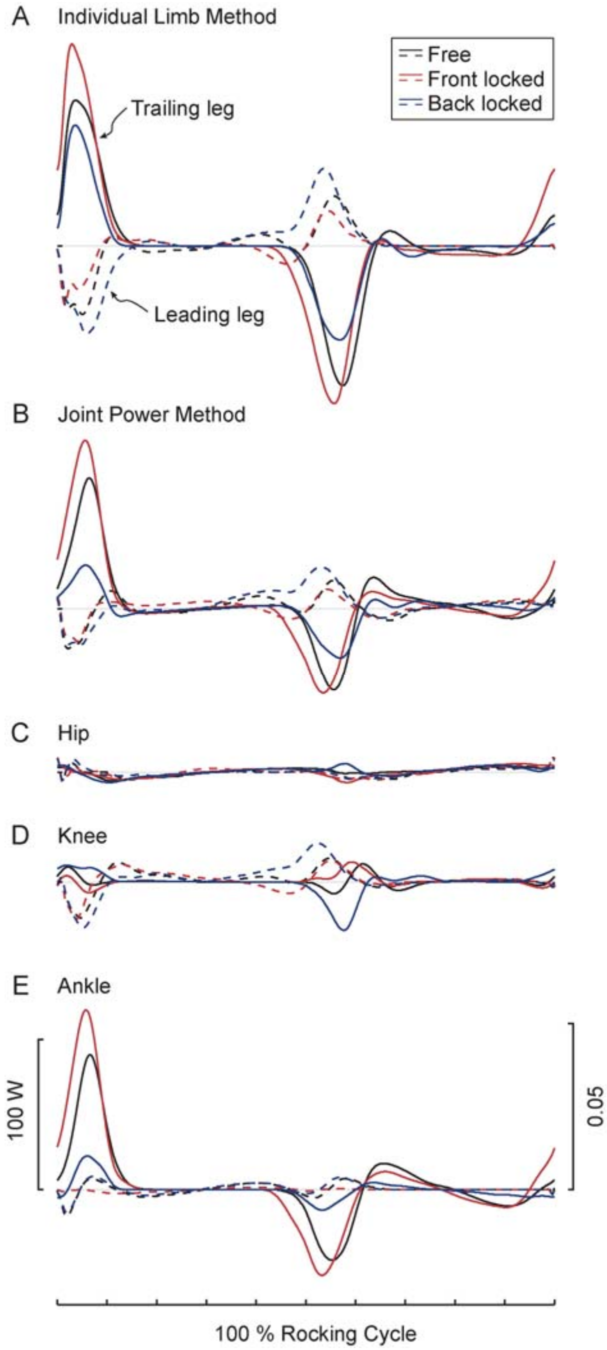


Figure 3-6. Comparison of ILM COM work rate and joint powers from JPM analyses (n=8). ILM COM work rate (a) is compared to the sum of individual joints powers (b). The latter is the sum of the instantaneous (c) hip, (d) knee, and (e) ankle joint powers.

CHAPTER 4: GENERAL CONCLUSIONS

4.1 Overview

The purpose of my thesis was to better understand the contribution of step-to-step transition work to the metabolic cost of walking and to determine the effect of sub-optimal step-to-step transitions on the magnitude of mechanical work and the resulting metabolic cost. The results of understanding the mechanics and energetics of step-to-step transitions have the potential to improve the rehabilitation of gait pathology by establishing a relationship between the mechanics of how we walk to the resulting metabolic cost of walking.

In Chapter 2, I isolated step-to-step transitions from leg swing by using sagittal plane rocking. The rationale for isolating transition is because while transition work makes up about 60-70% of the total energy of walking, other contributors such as leg swing may confound our measures of work. I found that sagittal plane rocking retained the energetic and mechanical characteristics of walking transitions. During rocking, the leading leg does negative work and the trailing leg does positive work, while the COM work rate and metabolic cost increased with increasing step length. I also found that the metabolic cost correlated with COM work during rocking transitions.

Since step-to-step transition is a major determinant of the metabolic cost of walking and since an often-stated functional goal of pathological walking rehabilitation is the minimization of metabolic cost of walking, I used our rocking paradigm to test whether sub-optimal transitions elevate metabolic cost.

In Chapter 3, I tested whether sub-optimal transitions are associated with increased metabolic cost. The work undertaken in Chapter 3 attempted to study transitions in more detail while applying the optimal transition hypothesis to pathological gait. In this chapter, I demonstrated that sub-optimal transition, from immobilizing the front ankle, caused a modest but significant increase in the total mechanical work and metabolic cost.

4.2 Limitations

Several limitations to the study require discussion. In Chapter 2, I isolated transitions by showing that the mechanical and metabolic energetic characteristics between walking transitions and rocking transitions were similar but the tasks themselves are different. The need to switch the COM velocity in rocking limited the magnitude of peak COM velocity and the magnitude of COM work rate. The efficiency of rocking (8%) is only slightly lower than walking (10%); however, this difference may be attributed to the need to transition backward during rocking.

In Chapter 3, I found that sub-optimal transitions are associated with increased COM work and metabolic cost in the front ankle locked condition. However, I was unable to find a statistically significant increase in the COM work rate and the metabolic cost when the back ankle was locked. A significantly shorter step length in our back ankle locked condition partially explained why there was no significant increase in the mechanical work when the back leg was locked. The presence of a rocker bottom may also help explain why the mechanical work of the back leg locked condition was not elevated. Future works should be mindful that a rocker bottom brace could confound any attempts at causing sub-optimal transition by ankle immobilization. Even though locking the back ankle did not increase the COM work rate or metabolic cost, this finding does not nullify our hypothesis that suboptimal transitions are associated with increased COM work rate and metabolic cost because the rocker bottom and a shorter rocking length prevented us from adequately imposing sub-optimal transitions.

4.3 Significance

Physics-based mathematical model predict that sub-optimal transitions causes elevated mechanical work. The cost of doing this work requires additional metabolic cost. Our study is the first to systematically test whether suboptimal transitions are related to increased metabolic cost in human empirical experiments and may provides researchers with another explanation for the elevated metabolic cost of pathological walking. The results of Chapter 3 suggest that one way to improve the economy of pathological walking, is to focus on achieving optimal transitions. Rehabilitation therapy strategies and assistive devices, such as prostheses and orthoses may help patients by matching the magnitude of work performed by each leg during step-to-step transition. The significance of this study may influence the rehabilitation field because it provides researchers with evidence that testable physics based-mathematical models can partially explain why pathological gait is so expensive.

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