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Active Versus Passive Control of Arm Swing: Implication of the Restriction of Pelvis Rotation during Human Locomotion

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ACTIVE VERSUS PASSIVE CONTROL OF ARM SWING:
IMPLICATIONS OF THE RESTRICTION OF PELVIS ROTATION
DURING HUMAN LOCOMOTION

A Thesis

Submitted to the Graduate Faculty of the
Louisiana State University and
Agricultural and Mechanical College
in partial fulfillment of the
requirements for the degree of
Master of Science

in

The Department of Kinesiology

by
Stephen Canton
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ABSTRACT

To date, it remains unclear how passive dynamics and active neural control contribute to arm swing during human locomotion. The passive hypothesis attributes arm swing to the passive transfer of energy from the legs to the arms via biomechanical linkages, while the active hypothesis states that arm swing is actively driven by muscles via neural mechanisms. The present study aims to investigate this phenomenon further by disrupting the biomechanical linkages, thereby directly challenging the passive hypothesis. Ten healthy individuals walked on a treadmill with and without an apparatus that constrained pelvis rotation at 3 different speeds (2 mph, 3 mph, and 4 mph). Spatial (upper and lower limb movement amplitudes) and temporal (movement frequencies and phase relationships between segment trajectories) aspects of limb movement were analyzed. The pelvis rotation was reduced by an average of 60.6% while constrained. As the treadmill speed increased, the movement amplitude of the upper and lower limbs increased. While the pelvis was constrained, arm swing amplitude decreased and the muscle activity of the upper limbs and lower limbs was similar to walking in the unconstrained condition. The movement frequency patterns and phase relations between segment trajectories were also conserved irrespective of speed and pelvis constraint conditions. These results provide evidence that passive elements are a significant factor in arm swing amplitude. However, the conserved EMG patterns and movement frequencies are suggestive of an underlying neural drive that contributes to the maintenance of the temporal aspects of gait. These observations are most likely due to passive dynamics in addition to neural mechanisms that maintain the rhythmic locomotor pattern via upper and lower limb central pattern generators (CPGs).

1. INTRODUCTION

Healthy human gait is bipedal, plantigrade progression of the human body (Inman, 1966). The mechanics of human gait involve the collaboration of the skeletal, neurological, and muscular systems (Fish and Nielson, 1993). For this reason, humans' ability to transport their bodies from one location to another involves the use of many components. Of these components, reciprocal arm movement is a typical feature that has raised many questions because the role of the arms is not obvious in upright, bipedal locomotion. However, many studies have shown evidence that arm swing can be attributed to the human effort to develop the most efficient strategy during locomotion (Pontzer et al., 2009, Kuhtz-Buschbeck and Jing, 2012, Meyns et al., 2013, Goudriaan et al., 2014). The present study seeks to investigate this phenomenon further.

Arm swing optimizes stability and energy consumption while moving about an environment (Donker et al., 2002, Marigold et al., 2003, Meyns et al., 2013). When the legs swing during locomotion, they cause a mechanical transmission of energy through the body that results in torque about the body's vertical axis (Li et al., 2001, Herr and Popovic, 2008, Pontzer et al., 2009). Arm swing is said to be a modular component of this rotational motion as it provides a counter torsional effect that minimizes the body's angular momentum about the vertical axis (Elftman, 1939, Park, 2008, Meyns et al., 2013, Goudriaan et al., 2014). The minimization of body torque keeps the ground reaction forces on the stance foot low in an effort to reduce overall energy cost of the body (Li et al., 2001, Park, 2008). The metabolic cost of walking increases when arm swing is suppressed, providing further support that arm swing is beneficial to locomotion (Umberger, 2008, Kuhtz-Buschbeck and Jing, 2012).

During gait, the arms tend to swing out of phase with the legs; i.e. the left arm swings forward with the right leg and vice versa (Elftman, 1939, Donker et al., 2002, Ivanenko et al., 2005, Pontzer et al., 2009). This phenomenon incites the question: Is the source of human arm swing the result of passive interactions during gait or is it due to muscles in the arms/shoulders actively contributing to the movements of the arms? It is a difficult question to definitively answer because there are numerous components involved that are working simultaneously.

1.1 Passive Hypothesis and Support

The passive hypothesis proposes that arm swing results from the energy generated by the legs during locomotion. A purely passive model attributes arm swing solely to the byproduct of movements of all mechanical linkages between the legs and arms, gravity, and inertia – therefore suggesting that arm swing is induced by motions of lower limbs, hips, torso (spinal column), shoulders, etc. (Meyns et al., 2013). In other words, the upper body behaves like a passive mass-damped system. The legs are the active controllers that transfer energy up through the spinal column and shoulders, and these, in turn, provide spring-like dampening to the system (Pontzer et al., 2009, Meyns et al., 2013).

Collins et al. (2009) performed a study in which they had participants walk (1) with contralateral limbs swinging in phase and (2) volitional swinging of the arms in phase with the ipsilateral leg. The authors observed very little shoulder and elbow joint torques for both gait conditions, suggesting that arm swing requires very little effort, i.e. little muscular activity is needed to maintain swing (Collins et al., 2009). In another study, Pontzer et al. (2009) reported that angular acceleration of the shoulders was correlated within increased trunk torsion, and arm acceleration was strongly correlated with angular displacement of the shoulders. These positive correlations support the notion that energy up-transfer from the legs to the arms is due to passive dynamics.

The passive hypothesis also proposes that muscle activity in the arms during locomotion is related to passive elastic forces, i.e. work done by elastic tendons (Hinrichs, 1990). Specifically, the shoulder muscles act primarily to stabilize the shoulders through eccentric or co-contraction (Pontzer et al., 2009).

1.2. Active Hypothesis and Support

The active hypothesis proposes that the nervous system actively controls muscles to generate arm swing (Donker et al., 2002, Pontzer et al., 2009, La Scaleia et al., 2014, Sylos-Labini et al., 2014). Results from past literature have revealed that the interlimb neural coupling observed during locomotion could be related to proposed human evolution from quadrupedal primates (Dietz et al., 2001, Dietz, 2002, Lacquaniti et al., 2012, Meyns et al., 2013). Bipedal and quadrupedal locomotion share common neuronal control mechanisms. These commonalities lend to the discussions about whether or not

these neural control mechanisms are residual/evolutionary (Dietz et al., 2001, Lacquaniti et al., 2012).

Many studies have suggested that the functionality of upper and lower limbs are interconnected by means of autonomic specialized neural circuits that lie in the spinal cord, coined central pattern generators or CPGs (Meyns et al., 2013). The conservation of temporal and spatial coordination between limbs elicited in healthy subjects, subjects with central nervous system (CNS) pathologies (spinal cord injuries, mesocephalic infants, etc.), and quadrupedal animals (Dietz, 2003, Ivanenko et al., 2005, Lacquaniti et al., 2012) provide evidence of these interconnections. Kush-Buschbeck and Jing (2012) showed that shoulder muscle activations persisted when arm movements were absent, contradicting Pontzer and others' hypothesis that the muscle activation occurs to stabilize the shoulder joint in relation to passive arm swing. La Scaleia et al. (2014) even showed that spatiotemporal kinematic patterns of stepping can be predicted by the temporal structure of the EMG patterns in the shoulder (deltoid) muscles.

The coordination of arm swing with other body segments has been observed not only in above ground locomotor modes, but also in other less common locomotor tasks (Dietz et al., 2001, Wannier et al., 2001). Wannier et al. (2001) observed a fixed relationship between the arm and leg movement frequencies during swimming and creeping. When flippers were added to the swimming tasks, the overall motion frequency of the arms and legs slowed, but frequency relationship remained. This fixed relationship was also supported by the EMG activity of the proximal arm and legs muscles during the different locomotor tasks. To dispute the idea that the coordination was due to mechanical interactions, the participants were also asked to swim while hanging in the air; a fixed frequency relationship between limbs still occurred. These authors argued that the presence of fixed relationships between limbs was indicative of coupled neural oscillators coordinating upper and lower limb motion. Similar findings of conserved temporal relationships have been observed across multiple populations, locomotor modes, and species (Wannier et al., 2001, Dietz, 2003, Haridas et al., 2006, MacLellan et al., 2013). This has led to the idea that there are neuromotor mechanisms that allow for beneficial coordinated use of the arms and legs during locomotion.

1.3. Statement of Problem

Past studies have investigated the effects of arm swing on locomotion by means of pendulum models/simulations, symmetric and asymmetric loading, inhibition of arm swing via bounding/held conditions, removal of arm excitation via simulation, etc. (Donker et al., 2002, Kuhtz-Buschbeck and Jing, 2012, Goudriaan et al., 2014). Other studies have evaluated the relative phasic relationships of movement between the arms and legs, pelvis and thorax, or, in rare cases, a combination of some of the aforementioned elements (Li et al., 2001, Bruijn et al., 2008, Pontzer et al., 2009, Sylos-Labini et al., 2014). If all the elements are included, studies begin to be limited in their analysis due to arduous task of managing numerous degrees of freedom.

Therefore, common discrepancies in these studies lie in the limitations of the model used or in the parameters evaluated to analyze the data – commonly being oversimplified and possibly skewing the results. The proposed study seeks to provide greater insight into whether arm swing is passive or active by directly challenging the passive hypothesis and restricting pelvis rotation during locomotion. It is hypothesized that if arm swing occurs due to passive mechanics, arm swing amplitude will increase based on speed effects and decrease while the pelvis is constrained. Due to this passive control, amplitudes of muscle activity will not differ when the pelvis is constrained. However, these activities will function to maintain the temporal aspects of this arm motion.

2. MATERIALS AND METHODS

2.1 Participants

Ten healthy adults (5 males and 5 females) participated in the study. Participants were excluded if they reported any previous musculoskeletal or neurological disorders that affect locomotion. All participants signed a written informed consent prior to participation in accordance with the Institutional Review Board at Louisiana State University. See Table 1.

Table 1. Subject Demographics. Gender, age, mass, height, and preferred walking speed (PWS) were recorded for each subject.

Subject	Gender	Age (yrs)	Mass (kg)	Height (m)	PWS (mph)
1	Male	25	97.5	1.75	3.0
2	Female	25	74.8	1.65	2.7
3	Female	21	78.0	1.63	2.5
4	Male	19	70.8	1.75	2.8
5	Male	23	77.1	1.78	3.0
6	Female	21	65.8	1.65	2.5
7	Female	21	58.1	1.73	2.5
8	Male	22	78.6	1.70	2.5
9	Male	39	90.0	1.80	2.3
10	Female	21	50.7	1.64	2.5
Mean	---	23.70	74.14	1.71	2.63
Std	---	5.70	13.87	0.06	0.24

2.2 Procedures

2.2.1 Pelvis Restriction Apparatus

1.5 X 1.5 inch steel square tubing was used to construct a 72" X 96" X 96" custom made cubic frame (see Appendix for a picture of the apparatus). Winches were placed along the vertical edges of the frame. The participants were equipped with a rock-

climbing harness (Bod Harness, Black Diamond TM), which was worn throughout the entire experiment and connected to the winches via ratcheting tie-down straps and carabineers. When tightening the straps to reduce pelvis motion, participants were told to place the edge of their heels on marked locations with feet shoulder-width apart. This method ensured that participants were standing in anatomical position with toes, pelvis, and shoulder girdle in the direction of motion. The straps were attached to the harness in four places and pulled taut in a systematic way to ensure that participants were not induced into a rotated position during the tightening process. The winches were tightened until the participants could not freely rotate hips when asked to do so.

2.2.2 Protocol

Participants walked on a treadmill at three different speeds: 2 mph, 3 mph, and 4 mph. Additionally, there were two walking conditions: (1) constrained (CON), whereby pelvis rotation was reduced when the harness was attached to frame, and (2) non-constrained (NC), without the harness attached to the frame. Preferred walking speed was determined prior to recording. Participants walked on the treadmill at variable speeds and self-reported his or her preferred speed. The participants walked constrained and unconstrained for each speed – for a total of six (6) trials. Trials were randomized within each walking condition block (NC and CON) and each block was presented randomly. With each condition lasting for approximately one minute, participants walked for 10 strides (prior to recording) to allow them to properly adapt to the walking speed and constraint. A minimum of 10 stride cycles were recorded for analysis once the participant verbally confirmed that he or she was comfortable. Following each condition, the treadmill was gradually slowed to a stop.

2.3 Data acquisition and processing

Full body 3-dimensional kinematics were recorded at 120 Hz using an 8-camera Vicon 512 system (Vicon Motion Systems Ltd, Oxford, UK). Spherical reflective markers were placed on the following landmarks and locations: spine of the C7 vertebrae, acromia, suprasternal notch, lateral humeral epicondyles, ulnar styloid processes, greater trochanters, anterior superior iliac spines (ASIS), midpoint between

posterior superior iliac spines (i.e. sacral), lateral femoral condyles, lateral malleoli, calcanei, 5th metatarsals, and the halluces. Three markers were placed on the harness approximately on the right and left iliac crests and one on the frontal mid-point between these points. The markers were designated as left harness (LHAR), right harness (RHAR), and front harness (FHAR) (Figure 1). All of the markers were placed directly on the skin, except the markers for the feet and harness (which were placed directly on the participants' shoes and the harness respectively).



Figure 1. Anterior view of pelvis and harness marker sets.

Electromyography (EMG) data were collected at 1800 Hz from 24 muscles (12 bilateral) using two, 16-channel, MA400-28 systems (Motion Lab Systems, Baton Rouge, LA). The muscles collected were the trapezius (TRAP), anterior deltoid (ADELT), posterior deltoid (PDELT), long head of triceps (TRI), latissimus dorsi (LAT), external oblique (EXOB), lumbar erector spinae (ERSP), gluteus maximus (GLUT), bicep femoris (BF), rectus femoris (RF), medial gastrocnemius (GAST), and tibialis anterior (TA). In preparation for electromyography, participants were shaved if needed and antiseptic alcoholic wipes were used to cleanse the desired locations. The electrode placement of the recorded muscles was determined by Surfaces EMG Non-Invasive Assessment of Muscles (SENIAM) guidelines or by palpation. Self-adhering Ag-AgCl bipolar surface electrodes were used for trunk muscles and self-contained Ag-AgCl electrodes (Model: MA-411, Motion Lab Systems, Baton Rouge, LA) were used for the lower limb muscles.

The bipolar electrodes were placed with an inter-electrode distance of two centimeters. All of the electrodes were secured over the muscle belly in line with the muscle fibers using adhesive tape. Self-adhesive elastic sports bandages were also used to provide additional security of the lower extremity electrodes.

2.4 Data analysis

2.4.1 Kinematics

Kinematic data were filtered offline using a zero-lag, second order low-pass Butterworth filter with a cut off frequency of 7 Hz. A stride cycle was defined as the time between two consecutive heel strikes of the right foot. Heel strike and toe-off were determined from the kinematic data by a velocity threshold program that was set at 0.05 m/sec. The right calcaneus and hallux markers were used to identify heel strike and toe-off times respectively. Each stride was time-normalized to 200 data points. A twelve (12) segment 3-dimensional linked-segment model was constructed consisting of the upper arms, lower arms, thighs, shanks, feet, pelvis, and trunk. Using the kinematic model, the limb trajectories, shoulder girdle rotation, and pelvis rotation were estimated. The anterior-posterior trajectories of the ulnar process and lateral malleolus markers were used to determine the upper and lower limb excursions respectively. In order to account for whole-body sagittal movements on the treadmill, the ulnar process marker time series was subtracted from the respective instantaneous acromial marker positions and the lateral malleolus marker time series was subtracted from the instantaneous greater trochanter positions. Finally, the upper and lower limb excursions were determined as the difference between the minimum and maximum peaks for the ulnar process and lateral malleolus markers respectively in the anterior-posterior direction. The values were calculated per stride and averaged over 10 total strides.

The shoulder girdle rotation about the longitudinal axis was calculated from the Z-Y-X Euler angle sequence with respect to the anatomical coordinate system. Due to frequent obstruction of the sacral marker, the pelvis and harness rotations were calculated using a two-dimensional analysis of the RASIS and LASIS markers (LHAR and RHAR of the harness) about the longitudinal axis. The use of harness markers was intended for the assessment of rotation of the pelvis within the harness. However, the relative rotation

was minimal. Rotational amplitude was determined similar to the trajectory of the limbs, as the difference between the minimum and maximum angles per stride cycle.

2.4.2 Electromyography

The EMG data were filtered offline by first using a 30Hz zero lag, second order Butterworth filter to attenuate any low frequency noise. Next, a second order 60 Hz bandstop Butterworth filter was used attenuate common electrical noise artifacts. The signal was then rectified and finally low-pass filtered at 10Hz to smooth the data. To quantify the EMG signals, the mean level of activity of the filtered EMG signals was calculated per stride for each participant. The muscle activity for each muscle collected was time-normalized to 200 points for a stride cycle, two consecutive heel strikes of the right leg. The activity was averaged for 10 consecutive strides.

2.4.3 Temporal Kinematics and EMG

The temporal kinematics and EMG were determined for following pairs of trajectories: (1) the right arm and right leg (ipsilateral segments), (2) the left arm and right leg (contralateral segments), and (3) the pelvis and shoulder girdle. The segment trajectories were normalized to one stride cycle (two consecutive heel strikes of the right foot). Using a Fast Fourier Transform (FFT), the phase angle of the fundamental harmonic was calculated for the time-normalized trajectories of the right arm, left arm, right leg, pelvis, and shoulder girdle for each stride and averaged over 10 stride cycles. The difference in the phase angle between trajectories pairs was used to determine the temporal relationship between the pairs of interest (ex. phase angle of the fundamental harmonic of the right arm trajectory and phase angle of the fundamental harmonic of the right leg trajectory). For ipsilateral segments, the fundamental harmonic phase angle of the right arm trajectory was subtracted from that of the right leg. For the contralateral segments, the fundamental harmonic phase angle of the left arm trajectory was subtracted from that of the right leg. For the pelvis and shoulder girdle, the fundamental harmonic phase angle of the shoulder girdle trajectory was subtracted from that of the pelvis. These differences were calculated in order to provide insight to the potential changes in temporal aspects of gait while walking in the pelvis constraint condition.

The temporal difference was also determined for the right PDELT activation and the excursion of the right arm. Due to the several frequencies present in the PDELT, the Fourier Transform was not used. Instead, the comparison of the PDELT and right arm trajectory was calculated using the time point of the maximum peak of the right arm trajectory subtracted from the peak value of the time-normalized averaged EMG profile for used for the PDELT activation.

An FFT was also applied to the anterior-posterior trajectory data of each limb to determine the movement frequency. The movement frequency was defined by the peak power in the FFT transform.

2.5 Statistical Analysis

A two-way pelvis constraint (NC versus CON) by walking speed (2, 3, and 4 mph) repeated measures ANOVA was used to determine statistical differences between experimental constraint conditions for the following variables: (1) arm swing excursion, (2) leg swing excursion, (3) pelvis rotation, (4) shoulder girdle rotation, (5) the mean muscle activity of all the muscles collected, (6) the phase angle difference between the ipsilateral upper and lower limb excursions, (7) phase angle difference between contralateral upper and lower limbs excursions, (8) phase angle difference between shoulder girdle and pelvis excursions, (9) the difference the time of peak activation of the PDELT and the time point of the peak arm excursion, (10) the frequency associated with the peak power of the right arm FFT, and (11) the frequency associated with the peak power of the left arm FFT. The significance level was $p < 0.05$ (two tailed).

Post-hoc Tukey HSD tests were conducted to investigate planned comparisons between the NC and CON conditions for the given speeds. Since data were similar on both sides of the body, only right side values were reported. Table 2 shows the muscles that were included in the study for each subject. Some muscle groups were excluded due to excess noise causing extreme outlier data. Also, pelvis and arm data were excluded for one subject (Subject 8) because the markers were obstructed.

Table 2. Muscles recorded for each subject. An ‘X’ denotes muscle groups of each participant used for the study. Any muscles excluded were outlier data due to obstruction or excessive noise observed during post-processing.

	SUB1	SUB2	SUB3	SUB4	SUB5	SUB6	SUB7	SUB8	SUB9	SUB10
TRAP	X	X	X	X	X	X	X	X	X	X
TRI	X	X	X		X	X	X	X	X	X
ADELTA	X	X	X	X	X	X	X	X	X	X
PDELTA		X	X	X	X	X	X	X	X	X
LAT	X	X	X	X	X	X	X	X	X	X
EXOB	X	X	X	X	X	X	X		X	X
ERSP	X	X	X	X	X	X	X	X	X	X
GLUT	X	X	X		X				X	X
BF	X	X	X	X	X	X	X	X	X	X
RF	X	X	X	X	X	X	X	X	X	X
GAST		X	X	X	X	X	X		X	X
TA	X	X	X	X	X	X	X		X	X

3. RESULTS

3.1 Kinematics

A representative set of trajectories are presented in Figure 2 for arm swing, leg swing, shoulder girdle rotation, and pelvis rotation. The CON condition significantly reduced pelvis rotation as compared to the NC condition (Figure 3A). When the pelvis was constrained, pelvis excursion was reduced by 55.2%, 52.5%, and 72.4% for 2 mph, 3 mph, and 4 mph respectively. Overall, the pelvis constraint reduced the pelvis excursion by an average of 60.6%. As walking speed was increased, pelvis rotation also increased and this was shown to be more prominent in the NC condition versus the CON condition, ($F_{(2,15.73)} = 14.40$, $p < 0.001$). Post-hoc tests showed that the interaction was driven by significant differences between constraint conditions at 2 mph ($p = 0.039$), 3 mph ($p = 0.013$), and 4 mph ($p < 0.001$).

Shoulder girdle rotation decreased with walking speed, ($F_{(2,18)} = 6.17$, $p = 0.009$;) (Figure 3B). Post hoc tests revealed that this effect was only significant between the speeds of 2 mph and 4 mph ($p = 0.009$). It also decreased in the CON condition as compared to the NC condition, ($F_{(1,9)} = 19.97$, $p = < 0.001$). However, the interaction was not significant ($p > 0.05$). These results imply that the shoulder girdle rotation differs significantly with greater disparity in speed, and it also differs between the two constraint conditions. The decrease in shoulder girdle/thorax rotation with increases in speed is a commonly observed phenomenon (Bruijn et al., 2008).

Arm excursion increased with walking speed. The increase was more pronounced in the NC when compared to the CON condition as shown by an interaction effect ($F_{(2,18)} = 13.74$, $p < 0.001$). The magnitudes of the excursions between constraint conditions were also greater with increased walking speed (2mph: not significant; 3mph: $p < 0.001$; 4mph: $p < 0.001$). The results imply that the differences of arm excursion are increasingly significant at greater walking speeds, i.e. 3 mph and 4mph (Figure 3C).

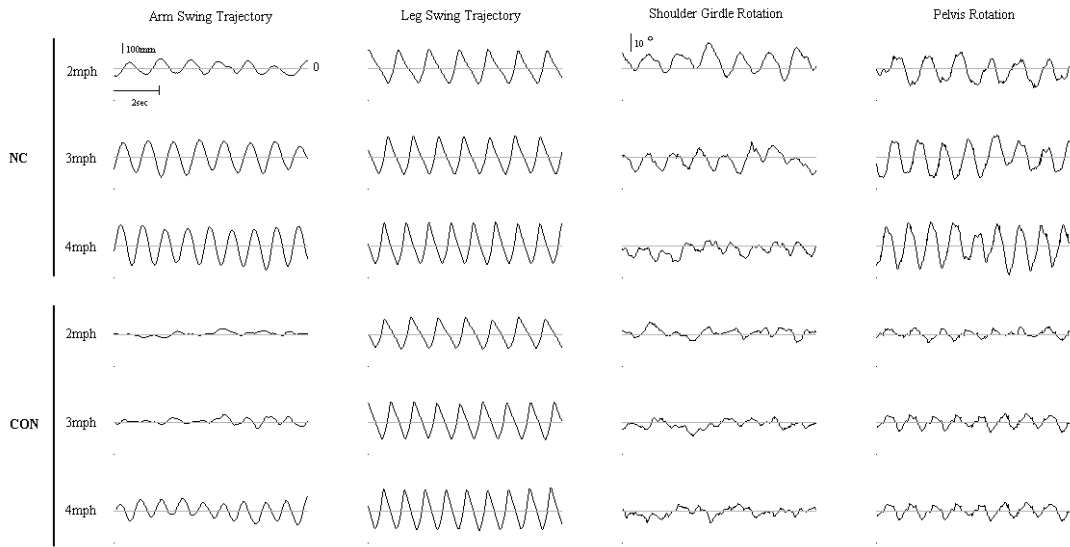


Figure 2. Kinematic trajectories. NC and CON denote non-constrained pelvis and constrained pelvis conditions respectively. These trajectories are of a representative subject.

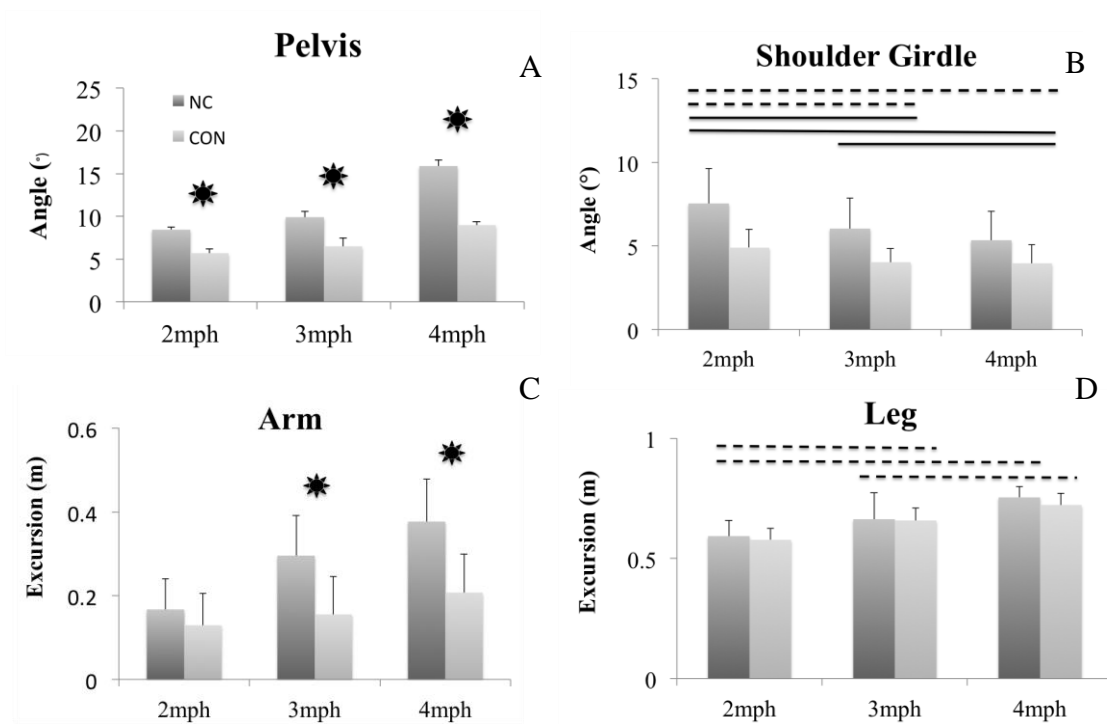


Figure 3. Mean and standard deviations of the excursions of the pelvis (A), shoulder girdle (B), arm swing (C), and leg swing (D) for all subjects. * denotes the significance of the interaction of speed and constraint, a solid line (—) denotes the significant of the main effect of constraint and a dotted line (---) denotes the significance of a main effect of speed.

Lower limb excursion increased with walking speed ($F_{(2,18)} = 50.1391$, $p < 0.001$). Significant differences were observed between all pairs of speeds (2mph-3mph: $p < 0.001$; 3mph-4mph: $p < 0.001$; 2mph-4mph: $p < 0.001$). The leg swing excursion increased with walking speed, but the effects of pelvis constraint were not significant ($p > 0.05$) (Figure 3D).

3.2 Temporal Patterns of Segment Coordination

The phase difference of the pelvis and girdle trajectories (girdle-pelvis, Figure 4A) differed significantly between the NC and CON condition ($F_{(2,9.16)} = 5.18$, $p = 0.048$). Ipsilateral upper and lower limb segments (right arm-right leg, Figure 4B) exhibited a main effect of speed ($F_{(2,18.81)} = 4.72$, $p = 0.022$), but post hoc tests showed that the effect was only significant between 2mph and 4mph ($p = 0.017$). The phase difference between contralateral trajectories (left arm-right leg, Figure 4C) was not statistically significant (p

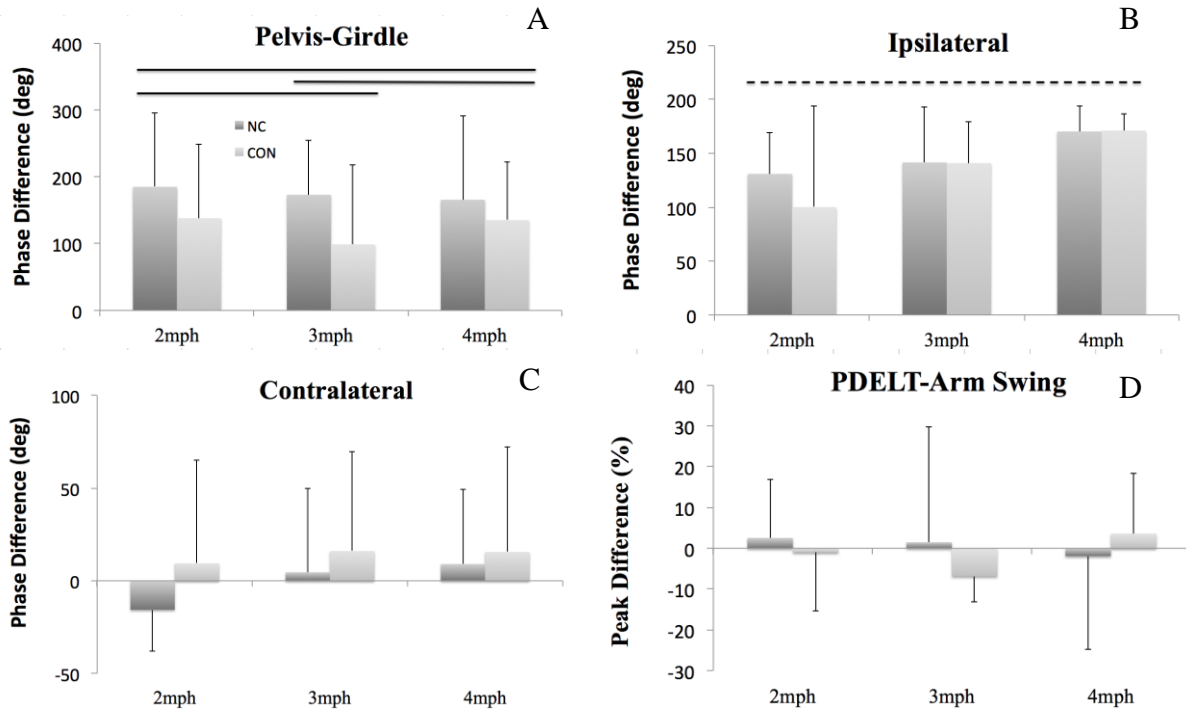


Figure 4. Phase differences of the pelvis-harness trajectories (A), ipsilateral trajectories (B), contralateral trajectories (C), and peak difference between PDELT activation and arm swing excursion (D). The mean and standard deviations are of all subjects. A solid line (—) denotes the significance of the main effect of constraint, and a dotted line (---) denotes the significance of a main effect of speed.

> 0.05). These results are evidence that, while the temporal coordination of pelvis-girdle rotation was affected by constraint, temporal relationships between contralateral and ipsilateral segments were conserved.

Walking speed affected the movement frequencies of arm swing trajectory ($F_{(2,18)} = 607.18$, $p < 0.001$) and leg swing trajectory ($F_{(2,18)} = 493.68$, $p < 0.001$). From 2 to 3 mph, the arm swing movement frequency increased from 0.81Hz to 0.99Hz, and to 1.11Hz at 4 mph (Figure 5A). The leg swing frequencies exhibited tendencies similar to the arm. The leg swing frequencies were 0.82Hz, 0.98Hz, and 1.11 at 2 mph, 3 mph, and 4 mph respectively (Figure 5B). For both the arm and leg, post hoc tests revealed effects of speed between all pairs of speed conditions ($p < 0.001$). The pelvis constraint also affected the movement frequencies of arm swing trajectory ($F_{(1,8.927)} = 17.42$, $p = 0.002$) and leg swing trajectory ($F_{(1,9)} = 11.33$, $p = 0.009$). An interaction of speed and pelvis constraint did not exist for the arm swing trajectory and leg swing trajectory ($p > 0.05$). These results show that movement frequencies of the ipsilateral and contralateral limbs increased in the pelvis constraint condition and with increased speed. However, the absence of an interaction shows that the effect of the pelvis constraint is only additive and therefore the overall temporal pattern remains consistent for each pelvis constraint condition.

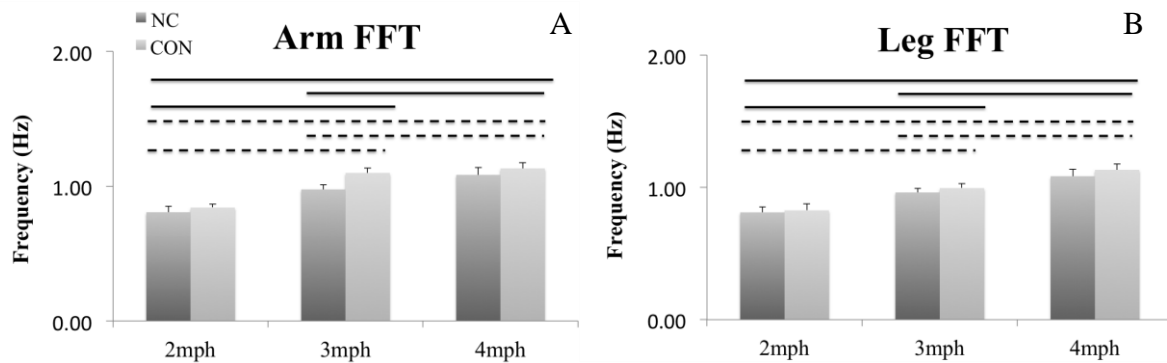


Figure 5. Mean and standard deviations of the power spectrum maxima of the Fast Fourier Transform (FFT) of the arm trajectory (A) and the leg trajectory (B) for all subjects. A solid line (—) denotes the significant of the main effect of constraint, and a dotted line (---) denotes the significance of a main effect of speed.

3.3 Muscle Activity

A main effect of walking speed upon the mean EMG activity for the following muscle groups: TRAP ($F_{(2,18)} = 18.9785$, $p < 0.001$), ADELTA ($F_{(2,18)} = 6.2207$, $p = 0.009$), LAT ($F_{(2,18)} = 14.3348$, $p < 0.001$), BF ($F_{(2,18)} = 13.8135$, $p < 0.001$), RF ($F_{(2,18)} = 78.0605$, $p < 0.001$), GAST ($F_{(2,18)} = 10.5151$), and TA ($F_{(2,18)} = 46.6898$, $p < 0.001$). In each of these muscles, as speed increased, the mean EMG activity increased. See Figure 6 and Table 3.

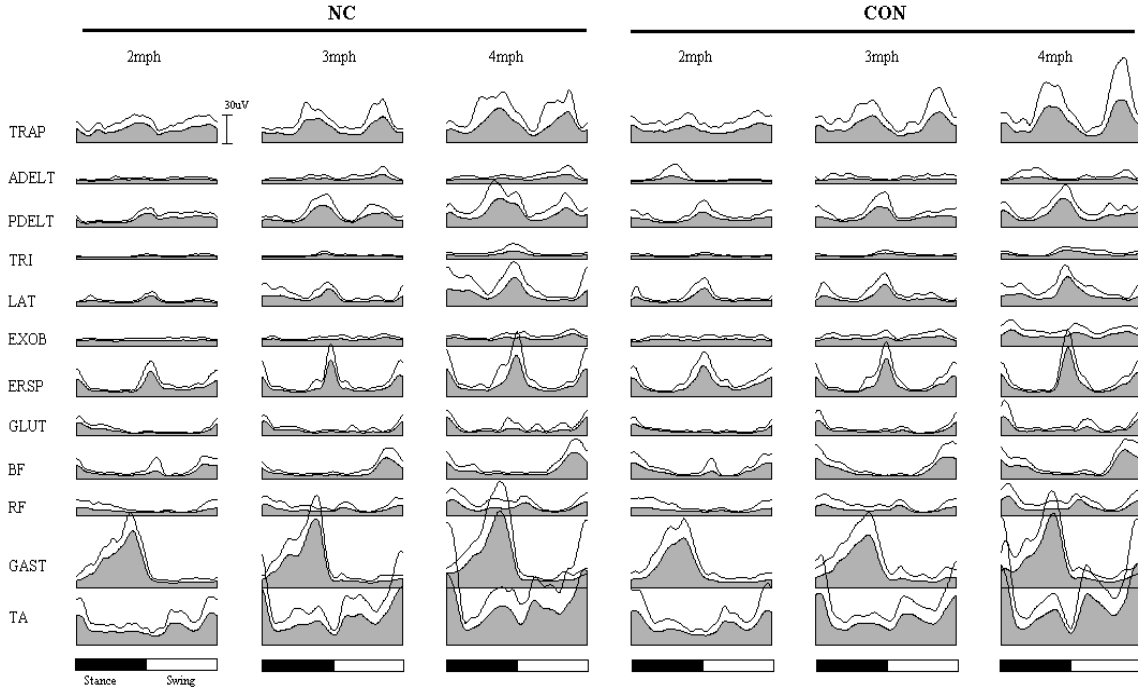


Figure 6. Time-normalized averaged EMG profiles of upper and lower extremities for all subjects. The solid line represents the standard deviation and the solid bar represents the stance (black) and swing (white) phases of a stride cycle. NC and CON denote non-constrained pelvis and constrained pelvis conditions respectively.

BF activity increased in the CON condition when compared to the NC, as shown by a main effect of constraint condition ($F_{(1,9)} = 6.4437$, $p = .032$). The increase in the mean muscle activity of the BF seems to be due to an increase in muscle activity during heel strike in the CON condition. An interaction between speed and constraint existed for ERSP activity, ($F_{(2,18)} = 6.5352$, $p = 0.007$). Further analysis revealed that the interaction effect only existed for the 2mph condition ($p = 0.015$). No significant differences were found for the TRI, EXOB, and GLUT muscles ($p > 0.05$).

The difference between the time point of the maximum peak of the PDELT muscle activation and time point of the maximum peak of the arm swing excursion was not significant ($p > 0.05$) (Figure 4D). This provides support to the temporal kinematic findings as it further suggests that not only are the phasic relationship of the body segment excursions maintained, but also the timing of the maxima between arm excursions and the PDELT, a muscle widely accepted to play a role in arm swing (Donker et al., 2002, Ivanenko et al., 2005, Pontzer et al., 2009). The EMG results, in general, suggest that muscle activity increases with speed but this pattern is conserved between the constraint conditions.

Table 3. Mean muscle activity (μV) of time normalized average EMG. The significant difference of a speed main effect is denoted by ^A, a constraint main effect is denoted by ^B, the interaction of the two is denoted by ^C

	2mph				3mph				4mph			
	NC		CON		NC		CON		NC		CON	
	Mean	Std	Mean	Std	Mean	Std	Mean	Std	Mean	Std	Mean	Std
TRAP	18.76	8.21	19.50	1.14	20.58	6.86	24.28	13.69	26.57	8.80	31.29	19.56
ADELT	5.86	1.98	5.60	1.69	7.00	3.24	6.38	3.92	7.73	3.82	7.46	4.02
PDELT	11.53	3.18	11.34	1.06	14.76	5.01	14.50	5.03	20.76	7.19	18.72	3.88
TRI	4.51	0.87	5.00	0.52	5.03	0.62	5.66	1.38	6.89	3.22	7.71	2.69
LAT	7.28	1.59	10.06	1.29	11.27	5.49	12.35	5.39	17.86	10.93	16.86	8.09
EXOB	7.76	1.87	8.71	0.69	9.08	2.50	10.51	4.69	10.71	2.51	15.50	6.87
ERSP	13.00	4.11	16.18	0.97	14.12	4.31	15.79	3.63	18.58	5.10	17.98	3.18
GLUT	7.13	2.17	6.93	0.30	7.71	2.24	7.79	2.34	9.26	3.85	10.34	3.16
BF	9.52	3.49	10.56	0.68	10.23	2.54	12.97	5.16	13.63	3.36	15.56	4.73
RF	6.12	4.59	6.92	0.47	7.87	4.57	8.59	6.02	10.98	6.01	12.15	6.97
GAST	24.99	5.18	24.17	0.59	27.52	6.07	27.59	12.13	32.64	8.42	33.36	15.18
TA	25.06	8.04	25.90	1.96	35.31	11.91	36.03	15.58	51.43	16.63	51.58	18.13

4. DISCUSSION

In previous literature, there is evidence that there are both passive and active elements to arm swing during human locomotion. Passive elements primarily exist due to the up-transfer of energy from the lower body (Pontzer et al., 2009, Kuhtz-Buschbeck and Jing, 2012, La Scaleia et al., 2014). On the other hand, active components have been shown to increase arm swing amplitude to aid in reduced energy expenditure (reducing motion about the vertical) and to create an out-of-phase walking pattern with the legs (Elftman, 1939, Li et al., 2001, Donker et al., 2002, Pontzer et al., 2009, Bruijn et al., 2010, Sylos-Labini et al., 2014).

The present study aimed to determine the effects of constraining the pelvis on arm swing during human locomotion. In accordance with previous studies, the amplitude of arm swing and the EMG activity of the arm muscles increased with increasing treadmill velocity (Figure 2, Figure 3C, and Table 3) (Murray et al., 1967, Donker et al., 2002, Kuhtz-Buschbeck and Jing, 2012). These results provided a well-studied baseline to compare the effects of the pelvis constraint. Also, the results showed that pelvis rotation was significantly decreased in the constrained (CON) condition, allowing for the primary research question of this experiment to be justifiably evaluated (Figures 2 and 3A).

The passive arm swing hypothesis proposes that upper body movement is driven by the up-transfer of energy from the legs to the pelvis and the shoulder girdle via biomechanical linkages (Pontzer et al., 2009, Kuhtz-Buschbeck and Jing, 2012). The excursions of the legs, arms, and pelvis increased with treadmill speed (Figure 2 and Figures 3A, 3C, 3D). When the pelvis was constrained, the excursions of the pelvis rotation, shoulder girdle rotation, and arm swing all decreased when compared to the non-constrained condition. The leg swing excursion, on the other hand, remained similar between constraint conditions (Figure 3D). The phenomenon of decreased of shoulder girdle rotation with increased treadmill speed may be in an effort to reduce the torsion on the spinal cord during high velocity locomotor modes. Nonetheless, the observed upper body kinematics were affected above the pelvis constraint, while the legs were not. The reduction of the excursion of upper body segments appears to be associated with the reduction of the excursion of the pelvis. This provides evidence that there is a disruption

in the biomechanical linkages through the body and, therefore, the changes in arm swing amplitude are a result passive mechanics.

Contrarily, the active arm swing hypothesis proposes that upper limb swing is driven by muscles in an effort to maintain cadence consistency and stability of the walking pattern (Elftman, 1939, Donker et al., 2002, Ortega et al., 2008, Pontzer et al., 2009, Pijnappels et al., 2010, Kuhtz-Buschbeck and Jing, 2012, Lacquaniti et al., 2012, La Scaleia et al., 2014, Sylos-Labini et al., 2014). The present study showed that mean EMG amplitudes in the recorded muscles were conserved between constraint conditions at every speed. This begged the question: As a result of the pelvis constraint, why was there an observed significant decrease in arm swing and shoulder girdle excursion, but conserved mean EMG amplitude of the arms and the legs? To reiterate, arm swing amplitude increased as walking speed increased. This occurred in both pelvis constraint (NC and CON) conditions, which implies that the patterns of arm swing amplitude are maintained regardless of the pelvis constraint condition. Meaning, the restriction of pelvis rotation did not disrupt the pattern of increased of arm swing amplitude with speed, coinciding with the pattern observed in the non-constrained pelvis condition. The EMG activity was conserved between pelvis constraint conditions (with the exception of BF and ERSP muscle activity), and, moreover, the phase differences between contralateral/ipsilateral limbs were also conserved. These results occurred despite the significant decrease of the phase difference between the pelvis and shoulder girdle in the pelvis constraint (CON) condition. This is an important result, given the passive hypothesis would predict a change in the phase correlation between limb segments associated with the change in the phase correlation of the shoulder and girdle (Pontzer et al., 2009). In summary, passive mechanisms appear to be a large factor in natural arm swing amplitude. However, the conservation of the upper/lower limb movement frequencies and EMG activation patterns between the pelvis constraint conditions may suggest an underlying neural drive to the upper limbs. The results are indicative of that upper limb movement is partly due to active neural mechanisms, i.e. active muscle control used to mediate temporal aspects of arm swing.

Furthermore, the conserved EMG patterns – increased activation associated with increased speed irrespective of constraint condition – may support notion that the speed

of locomotion is controlled through supraspinal input acting upon proposed CPGs. Prior research on decerebrate cats has shown evidence that quadrupedal stepping can be evoked by direct electrical stimulation of the mesencephalic locomotor region (MLR) of the brain (Garcia-Rill et al., 1983, Noga et al., 1988). Participants in the present study may be utilizing mechanisms similar to the MLR of cats to modulate upper and lower limb EMG activity based on the speed of the treadmill. Here, supraspinal inputs are analogous to the accelerator (modulator) for the engine of a moving car (the CPGs of the body), and this modulation does not seem to be affected by the pelvis constraint condition. It should be noted though that few studies have evaluated spatial EMG activity in the decerebrate cats. In a study by Debarae et al 2001, it was found that coordinated wrist and foot movements led to distributed activity in the cingulate motor cortex (CMC), supplementary motor area (SMA), premotor cortex (PMC), primary sensorimotor cortex (M1/S1), and the cerebellum, which were greater than the sum of activations during isolated limb movements. These results support the idea that the central nervous system innervates upper limb muscles in rhythmic way during locomotion.

There is a significant amount of literature that suggests coordination between the arms and the legs is very important and possibly deeply embedded in the human nervous system (Dietz, 2003, Haridas et al., 2006, MacLellan et al., 2013, Meyns et al., 2013). The results of this study are consistent with previous literature in that the temporal relationships between ipsilateral and contralateral segments are conserved (Zehr et al., 2001, Zehr and Duysens, 2004, MacLellan et al., 2013). Multiple studies have shown that the movement frequencies of limb trajectories (the correlation between contralateral and ipsilateral segments) are also conserved in atypical locomotor conditions, such as split belt walking or locomotion in response to a perturbation (Bruijn et al., 2010, Pijnappels et al., 2010, MacLellan et al., 2013). An interesting study evaluated the bi-directionality of interlimb coordination in which researchers suspended participants in an exoskeleton horizontal to the ground. With this setup, researchers told the participants to “walk” on a treadmill with their hands on an overhead treadmill to see if it would evoke leg movements similar to normal locomotion. They observed normal locomotion-like movements in 58% of their participants and also reported rhythmic activity of the proximal leg muscles. These results suggest that interlimb coupling is bi-directional, and

reinforces ideas that arm and leg temporal patterns are driven by functional neuronal innervation from the CNS (Meyns et al., 2013, Sylos-Labini et al., 2014). The FFTs of the upper and lower limbs provide evidence that the stride frequency and arm swing frequency patterns change for speed and pelvis constraint. However, the lack of an interaction suggests that the effects of pelvis constraint are additive – the movement frequencies are “adjusted” for the upper and lower limbs in a systematic way. This additive phenomenon can be attributed to mechanical interactions similar to ones observed in the aforementioned swimming study by Wannier et al. (2001). Donker et al. (2002) provided additional evidence of this observation in study that asked subject to walk on a treadmill in four different loading conditions. The limbs were loaded by adding a small mass to the wrists and ankles providing the following four conditions: (1) loading of the right arm, (2) loading of the both arms, (3) loading of the right leg, and (4) no loading on any limbs. The resulting movement frequencies were unaffected by the added mass for all conditions. Donker et al. (2002) presume that the observed adaptations were required to preserve a fixed temporal relationship between upper and lower limbs. To expound further, they argue that the result is due to the body’s effort to keep the limbs at the same frequency – via motor output – to maintain the stability of the walking pattern. It should also be noted that, similar to the current study and Wannier et al. (2001), there was an observed additive effect of the mechanical perturbation (added mass) to the movement frequency.

The current study results also revealed no significant change in the difference between the time of peak activation of the PDELT and the peak excursion of arm swing between constraint conditions and speed. This is an interesting finding because it suggests that the peak activation of arm muscle activity (within a stride cycle) occurs in a consistent temporal manner to maintain rhythmic arm swing. Harridas et al (2003) reported that stimulation of the superficial peroneal (foot) led to inhibition of the ipsilateral posterior deltoid during stance. On the other hand, the same stimulation facilitated activity in the posterior deltoid of the contralateral limb during contralateral stance. The consistency and proximity of the time points of the maximum activation of the PDELT and peak excursion of arm swing provide further evidence of the

phenomenon reported by Harridas and others (Haridas and Zehr, 2003, Zehr and Duysens, 2004).

While there is evidence of underlying neural mechanisms, it is difficult to pinpoint the specific mechanism(s) contributing to the maintenance of the temporal pattern of gait. Many studies have proposed the contentious role of CPGs. To reiterate, CPGs are proposed mechanisms by which motor neurons of the arm and leg muscles are innervated in a rhythmic manner during walking and running. The motor output to the muscles may be derived endogenously (i.e. without sensory or central input) from a spinal neuronal network, as suggested from research on locomotion of quadrupedal animals (Dietz, 2003). While passive elements exist, our results suggest that upper limb segments are modulated through active neural mechanisms. It has been argued that, during locomotion, the neuromotor system induces muscle activity in reaction to afferent stimuli – such as changes in body position (proprio-spinal connections) – in order to maintain temporal patterns of upper limb segments and muscles (Donker et al., 2002, Sylos-Labini et al., 2014). In sum, these mechanisms are believed to reduce the energy cost of walking and increase overall gait stability (Donker et al., 2002, Bruijn et al., 2010).

There were limitations in this study. Some muscle groups were removed from the analysis in particular subjects due to excess noise; real-time feedback of muscle activities during the experiment may have prevented this. A treadmill was used for practical reasons as it allowed for locomotion to occur when the pelvis was constrained. Finally, the harness caused frequent obstruction of the sacral marker on the pelvis. Future studies may include a harness that precludes the obstruction of the pelvis markers for a more complete, accurate analysis of pelvis rotation.

5. CONCLUSION

Walking with the pelvis constrained decreased the excursion of the upper limbs and shoulder girdle. It was hypothesized that if arm swing is mostly passive, arm swing amplitude and muscle activity would increase based on speed effects, but decrease while the pelvis was constrained. The current study allowed for the conservation of neural control parameters while still allowing for altered mechanics that may affect feedback and supraspinal contributions. The results suggest passive elements are a significant factor in arm swing amplitude. However, in support of the active arm swing hypotheses, the conserved muscle activation and movement frequency patterns are suggestive of an underlying neural drive that contributes to the maintenance of the temporal aspects of gait irrespective of speed or constraint. With this, the muscle activation described supports the notion of a coupling between cervical and lumbosacral spinal motorneuron output (Ivanenko et al., 2008). The movement state of the arms and the legs and the phase relationship between the limb pairs have been implicated to assist individuals with locomotor deficiencies due to trauma such as spinal cord injury, stroke, or even Parkinson's disease (Zehr et al., 2009). It should be noted that the contributions of active or passive arm swing could be affected by the extensive task-dependency observed during rhythmic arm movement (Zehr et al., 2001, Zehr and Duysens, 2004). Nevertheless, the observations still support the suggestion that rhythmic arm movements are controlled by CPGs similar to the legs, and this phenomenon has clinical relevance to gait rehabilitation and optimization.

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APPENDIX A: ADDITIONAL VIEWS OF EXPERIMENTAL SETUP



Figure 7. Front view of experimental setup

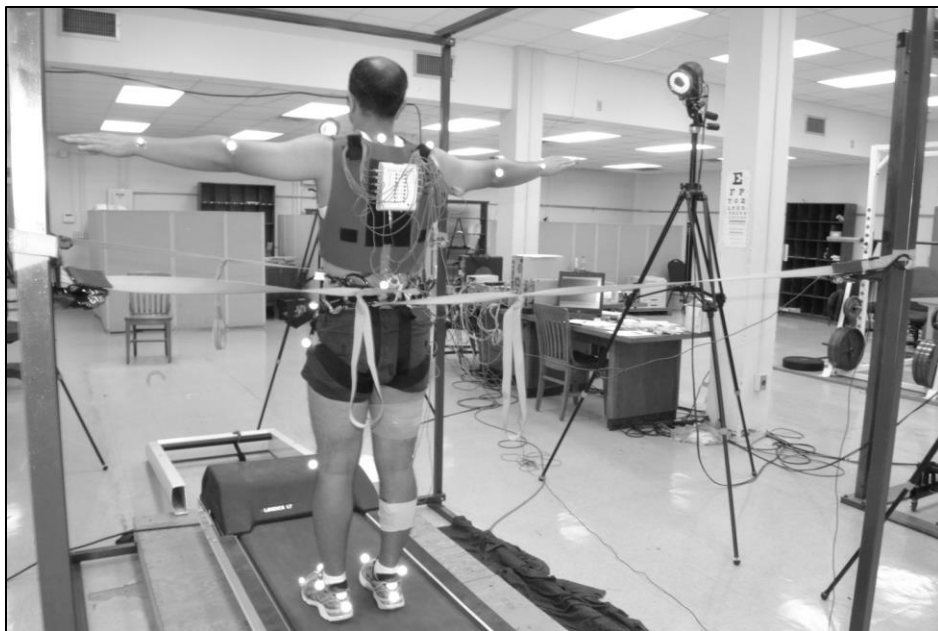


Figure 8. Rear view of experimental setup

APPENDIX B: INSTITUTIONAL REVIEW BOARD APPROVAL FORM

ACTION ON EXEMPTION APPROVAL REQUEST



TO: Michael MacLellan
Kinesiology

FROM: Dennis Landin
Chair, Institutional Review Board

DATE: February 12, 2015

RE: IRB# E9189

TITLE: Active versus passive control of arm swing: implications of the restriction of pelvis rotation during human locomotion

Institutional Review Board
Dr. Dennis Landin, Chair
130 David Boyd Hall
Baton Rouge, LA 70803
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F: 225.578.5983
irb@lsu.edu | lsu.edu/irb

New Protocol/Modification/Continuation: New Protocol

Review Date: 2/11/2015

Approved X **Disapproved** _____

Approval Date: 2/11/2015 **Approval Expiration Date:** 2/10/2018

Exemption Category/Paragraph: 2a,b

Signed Consent Waived?: No

Re-review frequency: (three years unless otherwise stated)

LSU Proposal Number (if applicable): _____

Protocol Matches Scope of Work in Grant proposal: (if applicable) _____

By: Dennis Landin, Chairman 

**PRINCIPAL INVESTIGATOR: PLEASE READ THE FOLLOWING –
Continuing approval is CONDITIONAL on:**

1. Adherence to the approved protocol, familiarity with, and adherence to the ethical standards of the Belmont Report, and LSU's Assurance of Compliance with DHHS regulations for the protection of human subjects*
2. Prior approval of a change in protocol, including revision of the consent documents or an increase in the number of subjects over that approved.
3. Obtaining renewed approval (or submittal of a termination report), prior to the approval expiration date, upon request by the IRB office (irrespective of when the project actually begins); notification of project termination.
4. Retention of documentation of informed consent and study records for at least 3 years after the study ends.
5. Continuing attention to the physical and psychological well-being and informed consent of the individual participants, including notification of new information that might affect consent.
6. A prompt report to the IRB of any adverse event affecting a participant potentially arising from the study.
7. Notification of the IRB of a serious compliance failure.

8. SPECIAL NOTE:

**All investigators and support staff have access to copies of the Belmont Report, LSU's Assurance with DHHS, DHHS (45 CFR 46) and FDA regulations governing use of human subjects, and other relevant documents in print in this office or on our World Wide Web site at <http://www.lsu.edu/irb>*

VITA

Stephen Canton is a native from Pittsburgh, Pennsylvania. He graduated from the University of Pittsburgh with a degree in bioengineering and a minor in mechanical engineering design. Stephen began his studies at Louisiana State University in 2013. He is a candidate to receive his master's degree in December 2015 and plans to work as a full-time clinical/biomedical engineer upon graduation. His technical interests are biomechanics and assistive robotics as it applies to physical medicine and rehabilitation.