

Neuromechanical considerations for incorporating rhythmic arm movement in the rehabilitation of walking

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We have extensively used arm cycling to study the neural control of rhythmic movements such as arm swing during walking. Recently rhythmic movement of the arms has also been shown to enhance and shape muscle activity in the legs. However, restricted information is available concerning the conditions necessary to maximally alter lumbar spinal cord excitability. Knowledge on the neuromechanics of a task can assist in the determination of the type, level, and timing of neural signals, yet arm swing during walking and arm cycling have not received a detailed neuromechanical comparison. The purpose of this research was to provide a combined neural and mechanical measurement approach that could be used to assist in the determination of the necessary and sufficient conditions for arm movement to assist in lower limb rehabilitation after stroke and spinal cord injury. Subjects performed three rhythmic arm movement tasks: (1) cycling (cycle); (2) swinging while standing (swing); and (3) swinging while treadmill walking (walk). We hypothesized that any difference in neural control between tasks (i.e., pattern of muscle activity) would reflect changes in the mechanical constraints unique to each task. Three-dimensional kinematics were collected simultaneously with force measurement at the hand and electromyography from the arms and trunk. All data were appropriately segmented to allow a comparison between and across conditions and were normalized and averaged to 100% movement cycle based on shoulder excursion. Separate mathematical principal components analysis of kinematic and neural variables was performed to determine common task features and muscle synergies. The results highlight important neural and mechanical features that distinguish differences between tasks. For example, there are considerable differences in the anatomical positions of the arms during each task, which relate to the moments experienced about the elbow and shoulder. Also, there are differences between tasks in elbow flexion/extension kinematics alongside differential muscle activation profiles. As well, mechanical assistance and constraints during all tasks could affect muscle recruitment and the functional role of muscles. Overall, despite neural and mechanical differences, the results are consistent with conserved common central motor control mechanisms operational for cycle, walk, and swing but appropriately sculpted to demands unique to each task. However, changing the mechanical parameters could affect the role of afferent feedback altering neural control and the coupling to the lower limbs. © 2009 American Institute of Physics. [DOI: 10.1063/1.3147404]

Traditionally, tasks used to assist in walking rehabilitation focus on the lower limbs and do not actively engage the arms. However, recent evidence suggests that rhythmic arm movement can have beneficial effects on the neural control of the lower limbs, such as enhancing leg muscle activity during locomotor tasks. Ambiguity remains regarding the specific details of how upper limb movement influences the control of the legs. This is in part due to the limited mechanical characterization of rhythmic arm movements. We present a neural and mechanical comparison of three rhythmic arm movements (arm cycling, stationary swinging, and swinging while

walking) in an attempt to expose considerations for the use of arm movement in walking rehabilitation. Our results suggest that while there are different kinematics across tasks the general neural control is conserved. The anatomical oscillatory “anchor point” position of the arm during the tasks is distinctive which relates to different moments about the elbow and shoulder. Results from mathematical analyses are consistent with conserved common central motor control mechanisms operational for arm cycling, arm swing while walking, and arm swing alone, but appropriately sculpted to demands unique to each task. However, changes in the mechanics of the tasks

can alter the neural control and coupling to the legs and therefore it is necessary to document and explore neuromechanical interactions with different arm tasks. This comparison can assist in the determination of the necessary and sufficient conditions for arm movement to assist in lower limb rehabilitation after stroke or spinal cord injury.

I. INTRODUCTION

It is a commonplace observation that we humans swing our arms when walking or running. Despite that, only recently has the role for the arms and the linkage between the arms and legs during human locomotion received much attention. Arm cycling on an upper limb ergometer (like leg cycling but while seated upright or standing and using the arms to cycle the cranks, Fig. 1) has been used to study the neural control of rhythmic limb movement (Zehr and Duysens, 2004; Zehr *et al.*, 2004; Zehr, 2005; Zehr *et al.*, 2007a). This approach led to the characterization of linkages between rhythmic activity of the arms and associated changes in lumbar spinal cord excitability (Ferris *et al.*, 2006; Frigon *et al.*, 2004; Huang and Ferris, 2004; Hundza and Zehr, 2008; Loadman and Zehr, 2007; Zehr and Duysens, 2004). As well, it has been shown that rhythmic arm movement enhances leg muscle activation during locomotor-like tasks thus providing evidence that arm movement may be an important adjunct to locomotor rehabilitation both alone and during rhythmic leg movement (Behrman and Harkema, 2000; Ferris *et al.*, 2006). Due to the availability of arm cycling ergometers and ease of administration of rhythmic arm cycling in a rehabilitation setting, it is tempting to use this specific task as a primer for enhancing lower limb activity during walking rehabilitation. Support for the use of rhythmic arm cycling in rehabilitation comes from the premise of a common core of neural control for many rhythmic movement tasks (Zehr, 2005). Accumulating evidence suggests that a movement involving coordinated rhythmic activity of one or both limb pairs (i.e., arms and legs) shares a common neural control even when the mechanics of the tasks are markedly different (Ivanenko *et al.*, 2005; Stoloff *et al.*, 2007; Zehr and Duysens, 2004; Zehr *et al.*, 2007a). As well, discrete voluntary tasks seem to be superimposed upon more automatic locomotor activities without altering the core neural pattern (Ivanenko *et al.*, 2005). This presents the idea that there is robust and malleable control of rhythmic movements in normal healthy subjects and a myriad of rhythmic movements can be used for rehabilitation so long as they access the same neuronal patterns. Further, this concept posits that rhythmic motor tasks using the arms may access common neural circuitry related to arm swing during gait and therefore could be useful in walking rehabilitation (Ferris *et al.*, 2006; Zehr, 2005; Zehr *et al.*, 2009).

Specific afferent feedback related to movements of the lower limbs is necessary to facilitate normal gait patterns (Hogue, 1969). For example, hip extension provides an important afferent signal during locomotion in both cats and human infants (McVea *et al.*, 2005; Pang and Yang, 2000). Currently, detailed characteristics of the role of neural signals related to rhythmic arm movement to engage relevant

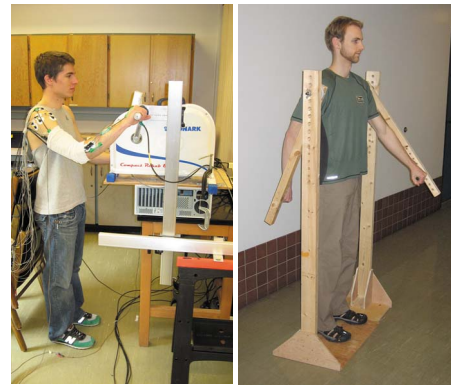


FIG. 1. (Color online) Rhythmic arm movement paradigms utilized in this experiment. The arm cycling paradigm (left panel) involves subjects performing circular movements with the arms on an upper limb ergometer. This is similar to leg cycling but while seated upright or standing and using the arms to cycle the cranks. Arm swing while standing (right panel) involves pendular movement of the arms while holding the handles of an arm swing apparatus that allows free shoulder flexion and extension yet restricts other arm movements.

leg muscle activation are still unclear. An important gap in knowledge is the determination of neuromechanical correlates related to arm movement in different tasks. That is, a neuromechanical characterization of the arm movement across rhythmic motor patterns remains to be done. As a general framework, during rhythmic movement, central motor drive activates the muscles while the mechanics of the musculoskeletal system influence the neural pattern of activity to alter and sculpt motor output (Abbas and Full, 2000; Dickinson *et al.*, 2000). Knowledge on the mechanical parameters of a task can help elucidate the type, level, and timing of neural signals and this information can be crucial for effective rehabilitation (Edgerton *et al.*, 2008). Additionally, lower limb coupling could be related to the movement state of both limb pairs, the phase of the movement, as well as specific neural signals from the upper limbs (Balter and Zehr, 2007; Ferris *et al.*, 2006; Frigon *et al.*, 2004; Hiraoka, 2001; Huang and Ferris, 2004; Kawashima *et al.*, 2008; Loadman and Zehr, 2007; Zehr *et al.*, 2004; Zehr *et al.*, 2007b). Thus a synthetic neuromechanical analysis of rhythmic arm movement could assist in quantification of the necessary and sufficient conditions for arm-assisted lower limb gait rehabilitation.

To date, rhythmic arm cycling has not received a neuromechanical comparison to its naturally occurring equivalent: arm swing during walking. Therefore the purpose of this study was to use simultaneous biomechanical and neurophysiological measurement techniques to compare and characterize rhythmic arm cycling to arm swing during walking. Additionally we chose to include an intermediate task of arm swing alone in our comparison (Fig. 1). This task involves swinging the arms in a pendular motion 180° out of phase with one another, similar to arm swing during walking but without lower body movement. This task could be used to evaluate arm swing without the mechanical interactions during walking and is similar to currently used rehabilitative tasks (Behrman and Harkema, 2000; Ferris *et al.*, 2006). Through this comparison we highlight important neural and

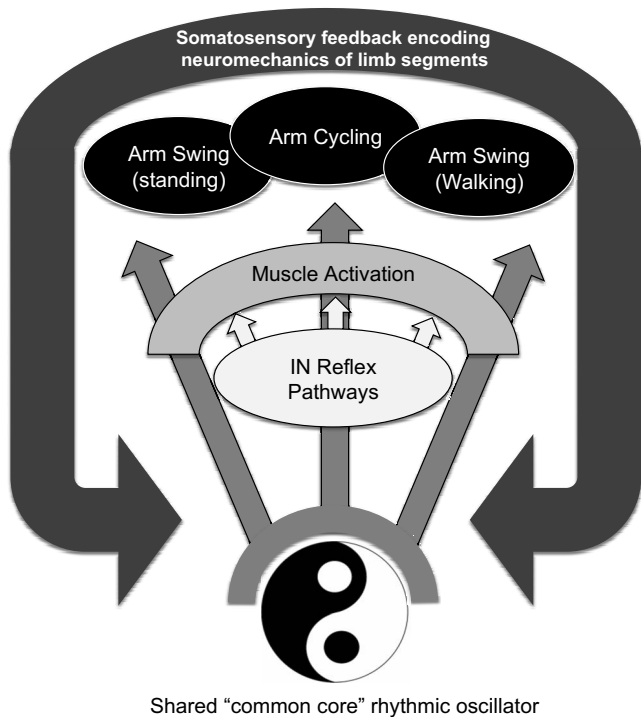


FIG. 2. General conceptual overview for the regulation of rhythmic human movement. Note that the effect of feedback encodes the neuromechanical interactions and projects to the motoneuronal pools, interneuronal pathways (IN), and the CPG yin-yang itself, and is subsumed in the output of the shared common oscillator. The effect of supraspinal input is taken as regulatory but is not shown. (Adapted from Zehr, 2005.)

mechanical task features that will be useful in future approaches using rhythmic arm movement to assist leg muscle activation during gait retraining. We hypothesized that there would be unique mechanical constraints for each task that may reflect changes in sensory feedback and function. Concurrently, there will be a common set of basic neural control parameters underlying all three tasks.

II. BACKGROUND

A. Can common core arm activities activate the legs?

The common core hypothesis presented by Zehr (2005) suggests that the control of rhythmic human movement is the responsibility of three parts of the nervous system and their interaction to produce the rhythm and pattern necessary to generate a robust compliment of movements. A basic schematic of this concept is shown in Fig. 2. The three basic components are supraspinal (not shown in the schematic but considered to have a regulatory role), spinal [in the form of central pattern generators (CPGs), yin-yang symbol, and interneurons], and afferent (somatosensory) feedback (from muscle, joint, and skin receptors). In principle this model suggests that the global timing of all tasks can be commonly set at a core level by CPGs. CPGs are a collection of neurons resident in the spinal cord that can produce oscillating activity without sensory feedback or additional supraspinal input. Sculpting of motor output occurs when information on local mechanical conditions (via afferent feedback) influences interneurons that regulate excitability of specific motoneuron

pools. That is, neuromechanical information is encoded in the somatosensory feedback which can then shape the muscle activity and alter the mechanical behavior. Therefore, understanding the control of rhythmic movement involves a synthetic analysis of both neural and mechanical task dynamics. Recently, factor analysis approaches have led to the ability to reduce mechanical and neural data to their component parts and therefore present an exciting way of ascertaining the core pattern of mechanical and neural activities (Bizzi *et al.*, 2008; Cappellini *et al.*, 2006; Daffertshofer *et al.*, 2004; Ivanenko *et al.*, 2005; McCrea and Rybak, 2008; Ting and McKay, 2007; Tresch *et al.*, 2006). This approach supports the common core hypothesis showing that many tasks share a common neural control despite slight alterations in activation profile and marked differences in mechanics (Bizzi *et al.*, 2008; Cappellini *et al.*, 2006; Ivanenko *et al.*, 2005; Stoloff *et al.*, 2007; Zehr, 2005; Zehr *et al.*, 2007a). For example we have shown that recumbent stepping, arm and leg cycling, and walking have similar neural control (Zehr *et al.*, 2007a; Stoloff *et al.*, 2007). This idea is mimicked in the schematic of rhythmic movement control (Fig. 2) which shows that arm cycling, arm swing while walking, and arm swing while standing share a common core timing. However, despite the basic core similarities, changes in afferent feedback can powerfully modulate the neural signal by exerting influence directly upon motoneuron pools, reflex interneurons, and through interneuronal networks associated with the CPG. An example of this concept is demonstrated by hip afferents which play a critical role in the neural control of walking as they can initiate, alter the tempo and halt gait (Dobkin *et al.*, 1995; Grillner and Rossignol, 1978; Knikou and Rymer, 2002; Lam and Pearson, 2001; McVea *et al.*, 2005; Pang and Yang, 2000; Steldt and Schmitz, 2004; Whelan *et al.*, 1995).

The schematic presented above is sufficient to discuss the regulation of rhythmic movement in just one limb (arm or leg) or limb pair (arms or legs). However, it is accepted that there are neural connections between the fore and hind limbs in animals and humans and the conditions of quadrupedal locomotor coupling requires further study (Dietz, 2002; Zehr and Duysens, 2004; Zehr *et al.*, 2009). Specifically, there is equivocating experimental evidence concerning the necessity of afferent signals from arm movement for proper enhancement and shaping of leg muscle activation. It is therefore pertinent to discuss the neuromechanical coupling between the arms and the legs as a precursor to appreciating quadrupedal movement (Zehr *et al.*, 2009). Recently Ijspeert (2008) has reviewed the development of mathematical models of CPGs. We would direct interest on the details of modeling to this comprehensive review. In brief there have been different approaches to mathematical modeling of CPGs depending on the area and phenomena of study. Highly detailed biophysical model, connectionist models, and systems of coupled oscillators have been used. As well, neuromechanical models have been developed to study the effect of sensory feedback on the CPG as well as to investigate concepts such as mechanical entrainment. The results of modeling have shown consistent results to human and animal experiments demonstrating that an oscillatory core is robust

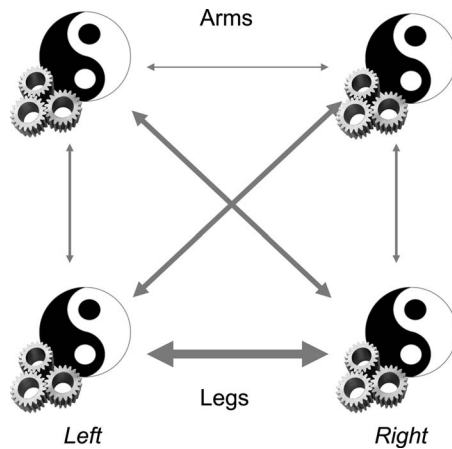


FIG. 3. Rhythmic CPG neural oscillator controllers (yin-yang) for all four limbs interacting with the mechanical environment (gears) meant to represent the regulation and coordination among and between all four limbs during movement. (Adapted from Zehr, 2005.)

and can maintain its rhythm while being sensitive to the pattern shaping influence of mechanical changes (Ijspeert, 2008). Figure 3 shows a basic schematic of the neuronal coupling between the arms and legs. In this figure each limb is portrayed as having a regulatory CPG element (symbolized as a yin-yang symbol) interacting with the mechanical environment (shown as gears) (Zehr, 2005). Within this framework it is possible that local neuromechanical interactions within any limb can affect neural coupling to the other limbs. The strength of the connections between the limbs is denoted by the thickness of the arrows connecting them. As such there is evidence for strong coupling between the legs (Brooke *et al.*, 1997) and weak coupling between the arms (Carroll *et al.*, 2005) while the details of crossed effects and ipsilateral arm to leg connections in humans are relatively unknown (Zehr and Duysens, 2004; Zehr, 2005). While all connections are of interest, for the sake of addressing a simple understanding of rhythmic movement control and the development of rehabilitative interventions, we will focus our discussion on arm-to-leg neuromechanical interactions. An experimental model to study the effect of upper limb movement on lower limb activity is to evoke reflexes in the legs changing parameters related to arm movement. Studies inducing passive flexion/extension of the elbow (Hiraoka and Nagata, 1999), arm swinging (Hiraoka, 2001), and different static arm postures (Delwaide *et al.*, 1973; Delwaide *et al.*, 1977; Eke-Okoro, 1994) have all demonstrated altered reflexes in leg muscles. Hiraoka (2001) attributed the alteration in soleus H-reflex to stretch in spindle afferents in anterior deltoid (AD) during shoulder extension. Taken together, these studies suggest that afferent feedback from the arms can induce changes in lower limb reflex excitability. However, there have been methodological concerns and lack of control between experiments warrants further study (Frigon *et al.*, 2004). Contrary to the above findings a series of experiments studying the effect of altering parameters of rhythmic arm cycling on reflexes in stationary legs has demonstrated a potentially sparse role of afferent feedback in effecting lower limb reflexes. These studies have shown that

rhythmic arm cycling produces a nonspecific effect on lower limb reflexes that is unaffected by load, range of motion, and muscle vibration, and yet responsive to changes in frequency of movement (Frigon *et al.*, 2004; Hundza *et al.*, 2007; Hundza *et al.*, 2008; Hundza and Zehr, 2008; Loadman and Zehr, 2007). The combined results using this paradigm are suggestive of a central rather than peripheral locus responsible for the effect of arm movement on leg reflexes (Hundza and Zehr, 2008). Concurrently, studies on “reduced” locomotor preparations (that is, not walking, but rhythmic activity when both arms and legs are moving), such as arm and leg stepping or cycling, have demonstrated that the movement state of both arms and legs may be necessary for maximized interlimb coupling (Balter and Zehr, 2007; Ferris *et al.*, 2006; Huang and Ferris, 2004; Sakamoto *et al.*, 2006; Zehr *et al.*, 2007). Huang and Ferris (2004) used a coupled recumbent arm and leg stepping device to show a clear facilitation in leg muscle activation that was graded to the activation of the arms and had similar timing. Additionally, studies using combined arm and leg cycling have shown that arm movement contributes to the modulation of reflexes in the legs in a functional and phase-dependent manner (Balter and Zehr, 2007). Balter and Zehr (2007) found that arm movement made a significant contribution to the reflex expression in the leg during the power phase (comparable to heel strike in walking). Conversely Sakamoto *et al.* (2006) found no contribution from the arms to reflex expression in the legs. However this could be due to different statistical tests as well as the fact that they used a mechanically uncoupled device (Balter and Zehr, 2007). Zehr *et al.* (2007b) found that the movement state of both limb pairs may enhance afferent feedback from the arms to affect reflexes in the legs during locomotor movements. In a recent study Kawashima *et al.* (2008) showed that rhythmic arm swing during gait retraining in incomplete cervical spinal cord injured subjects had a significant shaping effect on the pattern of soleus activation during rhythmic leg movement. This shaping was shown to either increase or decrease ankle extensor (soleus) muscle activation to produce a stereotypical pattern functionally related to stance and swing during walking. Because they studied spinal cord injured subjects and the effect was noticed in passive and active conditions they contended that supraspinal commands are not the sole source of the shaping effect and that sufficient afferent feedback from the upper limb movement is necessary.

Tasks that share common neural control may be equivalent at engaging neural circuits and coupling between the arm and the legs. However, changes in the mechanics of the movement task and the movement state of the both upper and lower limbs may alter neural coupling. Exploring the mechanics of different arm movement tasks is beneficial to elucidate the necessary and sufficient conditions for the enhancement of lower limb muscle activity

III. METHODS

Nine healthy participants (21–43 yr) participated in the study after providing informed written consent. The experimental protocol and consent form was approved by the Human Research Ethics Board at the University of Victoria and

conforms to the Declaration of Helsinki. Subjects were asked to perform three rhythmic arm movement tasks: (1) arm cycling while standing using a Monark™ (Compact Rehab 871E, Sweden) inertial load arm cycle ergometer (cycle), (2) arm swing while standing using a custom made arm swing apparatus (Dave Smith, University of Victoria) (swing), and (3) level walking on a motorized treadmill (Desmo M, Woodway™) (walk). For cycle the extent of shoulder flexion/extension was defined by the movement path and the anatomical parameters of each subject. During swing the extent of shoulder excursion was matched to the range occurring during cycle and maintained through visual comparison to a manual goniometer. During walk the extent of shoulder excursion was subject selected. Additionally a swing trial with a self-selected shoulder excursion was performed to compare shoulder excursion and electromyography (EMG) activation. Movement frequency was determined by having subjects walk at a comfortable pace on a treadmill and then calculating the period for one complete right shoulder extension cycle [position corresponded to right heel contact (refer to Fig. 4)]. This frequency was then maintained for all tasks with the use of a metronome. The average corresponding frequency was 0.88 ± 0.08 Hz (mean \pm standard deviation). Prior to collection, during practice, we adjusted the handles on the swing apparatus to anatomical dimensions for each participant to allow for comfortable arm swing. Figure 4 shows the details of each task as well as the shoulder kinematic relation between tasks.

A. Kinematics

A ViconPeak motion capture system (6 Camera 460, Vicon Peak™, Centennial, CO) was used to record the movement of the whole body. Reflective markers were placed on the subject using the VICON™ Plug-in-Gait model. Anthropometric data from each subject were necessary for an inverse dynamics analysis. These data consisted of mass (kilogram), height (millimeter), leg length (millimeter), knee width (millimeter), ankle width (millimeter), shoulder offset (millimeter), elbow width (millimeter), wrist width (millimeter), and hand thickness (millimeter). Data capture was conducted at 120 Hz. Marker trajectory data were filtered using a Woltering filter (Vicon™) and Cardan angles were calculated for right ankle, knee, hip, shoulder, elbow, and wrist [Plug-In Gait, Vicon™, refer to Fig. 4(c) for sign conventions of presented kinematic variables]. Joint angular velocities and accelerations were calculated for right ankle, knee, hip, shoulder, elbow, and wrist. A two-dimensional inverse dynamics analysis was performed using the measured kinematic values to determine shoulder and elbow flexion/extension moments (Winter, 2005).

B. Kinetics

An ATI™ Industrial Automation force/torque transducer (Apex, NC, USA; Gamma model) was mounted at the right hand ergometer interface of the arm cycle ergometer and the arm swing apparatus [refer to Fig. 4(c) for force conventions]. The force transducer collected force and torque in three dimensions and data were collected at 1000 Hz.

C. Electromyography

After cleansing the skin with alcohol swabs, disposable 1cm surface electrodes (Thought Technologies Ltd.) were applied in a bipolar configuration using a 2 cm interelectrode distance over eight muscles of the right arm and back. Muscles included AD, middle deltoid (MD), posterior deltoid (PD), long head of biceps brachii (BB), long head of triceps brachii (TB), erector spinae cervical (ESC), erector spinae thoracic (EST), and erector spinae lumbar (ESL). The primary actions of these muscles produce shoulder flexion (AD), shoulder extension (PD), shoulder abduction (MD), elbow flexion (BB), elbow extension (TB), and lateral bending of the spine at a cervical (ESC), thoracic (EST), and lumbar level (ESL). Ground electrodes were placed over electrically neutral tissue either on the acromium process or clavicle. To minimize mechanical artifact the electrodes were secured with tape and medical wrap. EMG signals were preamplified and band pass filtered at 100–300 Hz (Noraxon™, Telemetry, Scottsdale, AZ) and collected at 1000 Hz.

D. Data acquisition

Two computers were synchronized by a trigger switch that signaled the start of data collection for motion capture with Vicon workstation™ software and to collect data from the ATI™ force transducer as well as EMG. Synchronization timing was evaluated through comparison of arm cycle position using an optical encoder during the arm cycling task.

E. Data analysis

A custom-written analysis program was developed in LABVIEW™ (National Instruments™, Austin, TX, USA) to condition all data. Single subject EMG data were full wave rectified and then passed through a second order Butterworth filter with a low pass cutoff frequency of 6 Hz. EMG data were then normalized to 100% of the maximum EMG from each muscle taken across all tasks. Single subject force and torque data were filtered using a second order Butterworth filter with a low pass cut off frequency of 20 Hz prior to averaging. All data were averaged to 100% of the movement cycle from right shoulder extension maximum to subsequent right shoulder extension maximum (Fig. 4).

F. Mathematical analysis

The basic patterns in neural control and mechanics were examined using separate principal components analysis (PCA) for EMG and kinematic data recorded from cycle, swing, and walk. Details of the analysis are similar to our previous task comparisons (Stoloff *et al.*, 2007; Zehr *et al.*, 2007a). Briefly each PCA was performed using the princomp function in MATLAB (The Mathworks, Inc.). First a correlation matrix was made to examine linear dependence between muscles (8×8 matrix) and kinematics (11×11 ; separate analysis for both x-flexion/extension and y-adduction/abduction). Factor scores were then determined using a varimax rotation of the eigenvectors of each matrix, in which variables with similar activity were grouped together. The variance was then calculated using the eigenvectors and ei-

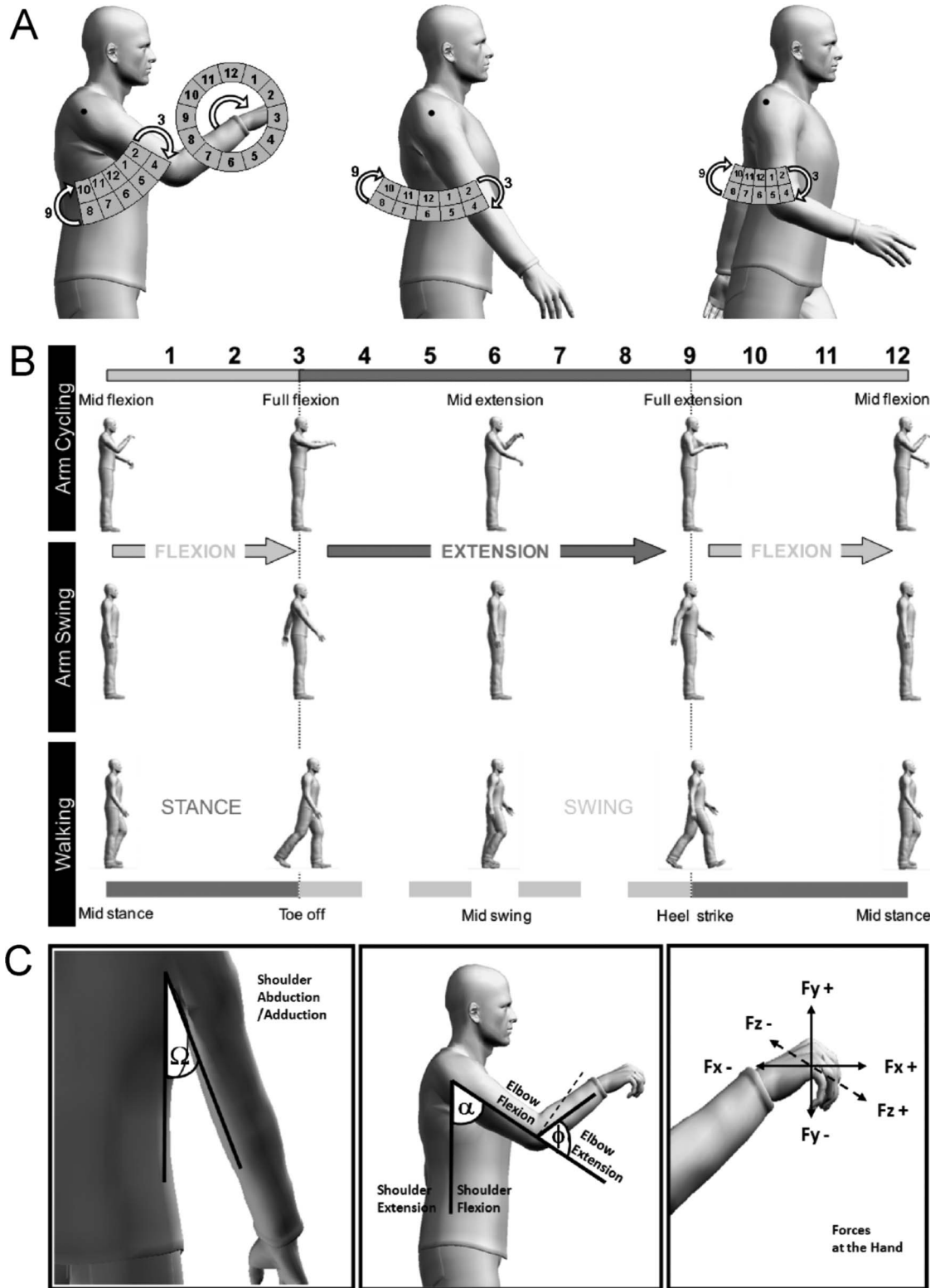


FIG. 4. (a) Schematic diagram relating the phases of movement for arm cycling, arm swing while standing, and arm swing while walking. The movement cycle was normalized to 100% of shoulder flexion/extension and displayed in reference to the clock face movement cycle from arm cycling (1–12) Note: the arm cycling and arm swing devices are not shown. (b) Sign conventions at the right arm and ergometer hand crank. Ω , α , and Φ represent shoulder abduction/adduction, shoulder flexion/extension, and elbow flexion/extension, respectively. Participants performed all three tasks while EMG, kinetic, and kinematic data were recorded.

genvalues. Next the percent of total variability for each factor was determined by dividing its variance by the sum of the variances. Because the first four factors in all three tasks for both EMG and kinematic data accounted for at least 95% of

the variability, we only considered scores from the first four factors for further analysis (Ivanenko *et al.*, 2005).

Additionally we performed a correlation analysis on factor scores to assess the similarity between cycle, walk, and

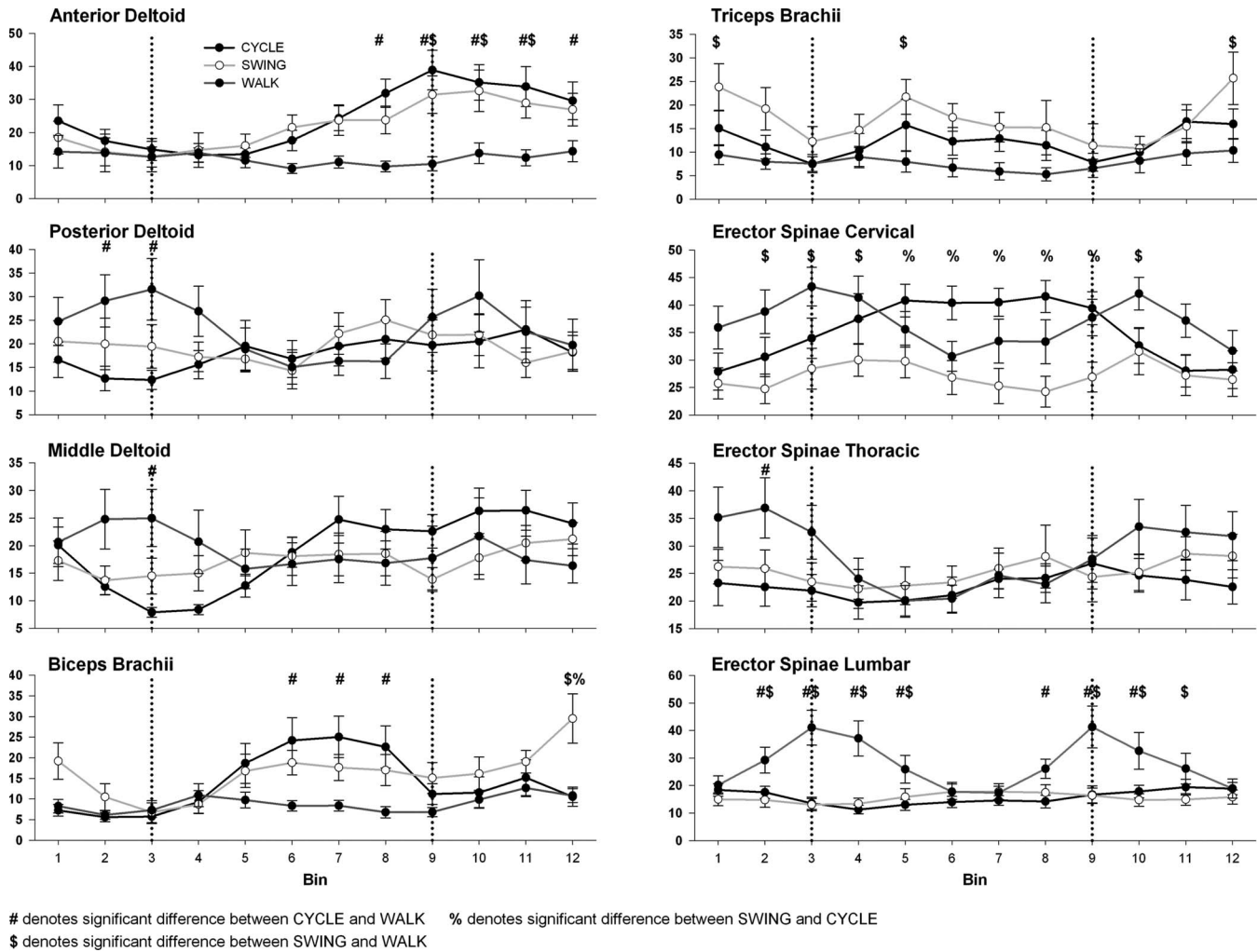


FIG. 5. Comparative EMG traces for muscles of the right arm and trunk averaged across all subjects for cycle, swing, and walk tasks. Significant differences between cycle and walk, swing and walk, and swing and cycle are denoted by #, \$, and %, respectively. Vertical dashed lines denote the 3 o'clock (shoulder flexion) and 9 o'clock (shoulder extension) positions during the movement cycle, respectively.

swing using the corrcoeff function in MATLAB™. Briefly we calculated the correlation coefficient of each first factor score between each pair of tasks. We chose a criterion of 0.4 to indicate a good correlation between factors based on a previous research using similar methods (Ivanenko *et al.*, 2005; Zehr *et al.*, 2007a).

G. Additional statistics

To determine the effects of task and phase (three tasks and twelve phases) on average EMG amplitude across phases we used a (3 task × 12 phase) repeated measure analysis of variance (RM ANOVA). Additionally we used tests of planned comparison to determine differences between tasks at the same phase and differences between phases within the same task. Also we performed RM ANOVAs on factor loadings and Fisher z-transformed correlation coefficients of the first factors from Kinematic and EMG PCA with planned comparisons. Significance level was set to $P < 0.05$

IV. RESULTS

A. Electromyography

Overall the results of muscle activation showed greater differences between walk and other two tasks. Figure 5 shows the activation profiles for all muscles with the statistical results for task dependent difference. The x-axis is presented as equidistant bins represented as “hourly” increments on the clock face movement cycle from arm cycling. The dashed line represents maximum shoulder flexion (3 o'clock) and extension (9 o'clock). For simplicity, results will be discussed in the functional frame of reference from Fig. 4 with respect to full and midrange flexion and extension. Significant task dependent differences in muscle activation between tasks are marked with symbols (#,\$,%) denoting differences between cycle and walk, swing and walk, and swing and cycle, respectively. There are greater differences between cycle and walk (19 differences) and swing and walk (16 differences) than between swing and cycle (six differences). Cycle and swing are different from walk in AD muscle activity from full extension into midflexion and ESL during full

flexion and full extension. The only cases where swing and cycle are different are in ESC from mid to full extension and BB during midflexion. Cycle and walk are different for both PD and MD at full flexion where these muscles are usually relatively silent during cycle and active during walk. The phase-dependent modulation of muscles during cycle, swing, and walk is in general agreement with past research (Balles-teros *et al.*, 1965; Hinrichs, 1990; Zehr and Chua, 2000; Zehr and Kido, 2001). In summary AD is active from midextension to midflexion for both cycle and swing but relatively silent during walk. PD is active during full flexion in walk while this muscle is relatively low with no major pattern for cycle and swing. MD is active concurrently with AD during cycle in midextension to midflexion, silent during swing, and active concurrently with PD during walk in full flexion. BB reaches a maximum peak during midextension in cycle and midflexion concurrent with TB in swing. For back muscles ESC is active during cycle from midextension to full extension and during walk for both full flexion (concurrent with EST and ESL) and full extension (with ESL). Additionally for the self-selected swing trial there was a low level of AD activity during extension and a low level of PD activity during flexion without a noticeable pattern in other muscles recorded (not shown).

B. Kinematics

Data relating to shoulder flexion/extension (α), abduction/adduction (Ω), and elbow flexion/extension (Φ) are displayed as cardan angles in Fig. 6(a) (conventions for the presentation of these results are shown in Fig. 4). The range of shoulder flexion/extension was smallest in walk and largest in cycle. The absolute shoulder position for walk and swing was such that it traveled into flexion and extension passing equally through the sagittal midline while during cycle the arm is always in a flexion quadrant in front of the subjects' frontal plane. Cycle moved from abduction to adduction during the flexion phase and from adduction to abduction during the extension phase. This pattern was inverted during walk. Swing reached the greatest peak abduction during full flexion and a smaller peak during full extension while being adducted during the transition stages of midflexion and midextension. Elbow flexion/extension was greatest during cycle, smallest in swing, and almost reversed to cycle during walk. The pattern during cycle is elbow extension during mid to full flexion and elbow flexion during mid to full extension. This is different to walk where elbow flexion occurs during full flexion to midextension and elbow extension occurs during midextension to full extension.

C. Kinetics

The forces at the hand are shown in Fig. 6 for cycle and swing. The literary convention for the discussion of forces will be in relation to Fig. 4(c) and will be upward (F_y+), downward (F_y-), forward (F_x+), backward (F_x-), medial (F_z-), and lateral (F_z+). During cycle there is a peak downward and medial force at full flexion that persists into midextension. From midextension to midflexion the downward force decreases with an increasing backward and lateral force

at full extension. During swing there is a peak downward and forward force at midextension. Additionally there is a peak medial force at midflexion that is decreased at full extension. Shoulder and elbow moments from the two-dimensional inverse dynamics analysis are shown in Fig. 6(c) with positive values representing flexor and negative values extensor moments, respectively. For both elbow and shoulder moments during walk there is a peak extensor moment occurring during full flexion and a peak flexor moment during full extension. This is a similar pattern during swing with a noticeable increase in amplitude and a more prolonged extensor moment during full flexion. For cycle, for both shoulder and elbow, there is a constant extensor moment throughout the cycle that is smallest during flexion and largest during extension.

D. Mathematical analysis

1. Kinematics

For all tasks four factors explained more than 95% of the variance for x and y kinematics [Figs. 7(a) and 7(b)]. For cycle and swing x -kinematics the first factor accounted for 58% and 62% of the variance, respectively, while the first factor in walk accounted for only 45% of the variance [Fig. 7(a)]. Further walk required four factors to explain more than 92% of the variance where cycle and swing only required three factors. For cycle and swing y -kinematics the first factor accounted for 63% and 73% of the variance, respectively, while the first factor in walk accounted for only 45% of the variance [Fig. 7(a)]. Further walk required four factors to explain more than 95% of the variance where cycle and swing only required three factors. Correlation coefficients for the comparison of the first factor show similarity for all tasks for kinematics (both x and y directions) (Table I). Further, analysis of factor loading reveals that for kinematics there was no difference between shoulder flexion/extension or shoulder abduction/adduction but differences between elbow flexion/extension [Fig. 7(d)].

2. EMG

Four factors explained more than 98% of the variance for all tasks [Fig. 7(c)]. For swing the first factor accounted for more than 80% of the variance while the first factor for cycle and walk only accounted for 72% and 69% of the variance, respectively. Additionally while two factors explained more than 92% of the variance for cycle and swing an addition of a third factor was necessary for walk to account for more than 94% of the variance. Correlation between the first factor shows differences between walk and cycle and walk and swing with no difference between cycle and swing (Table I). Analysis of the factor loading shows that there are no noticeable differences between individual factor loadings for different muscles [Fig. 7(e)].

V. DISCUSSION

In the present study we compared three rhythmic arm movement tasks in order to highlight neuromechanical features that require consideration with respect to using arm movement to assist walking rehabilitation. The results dem-

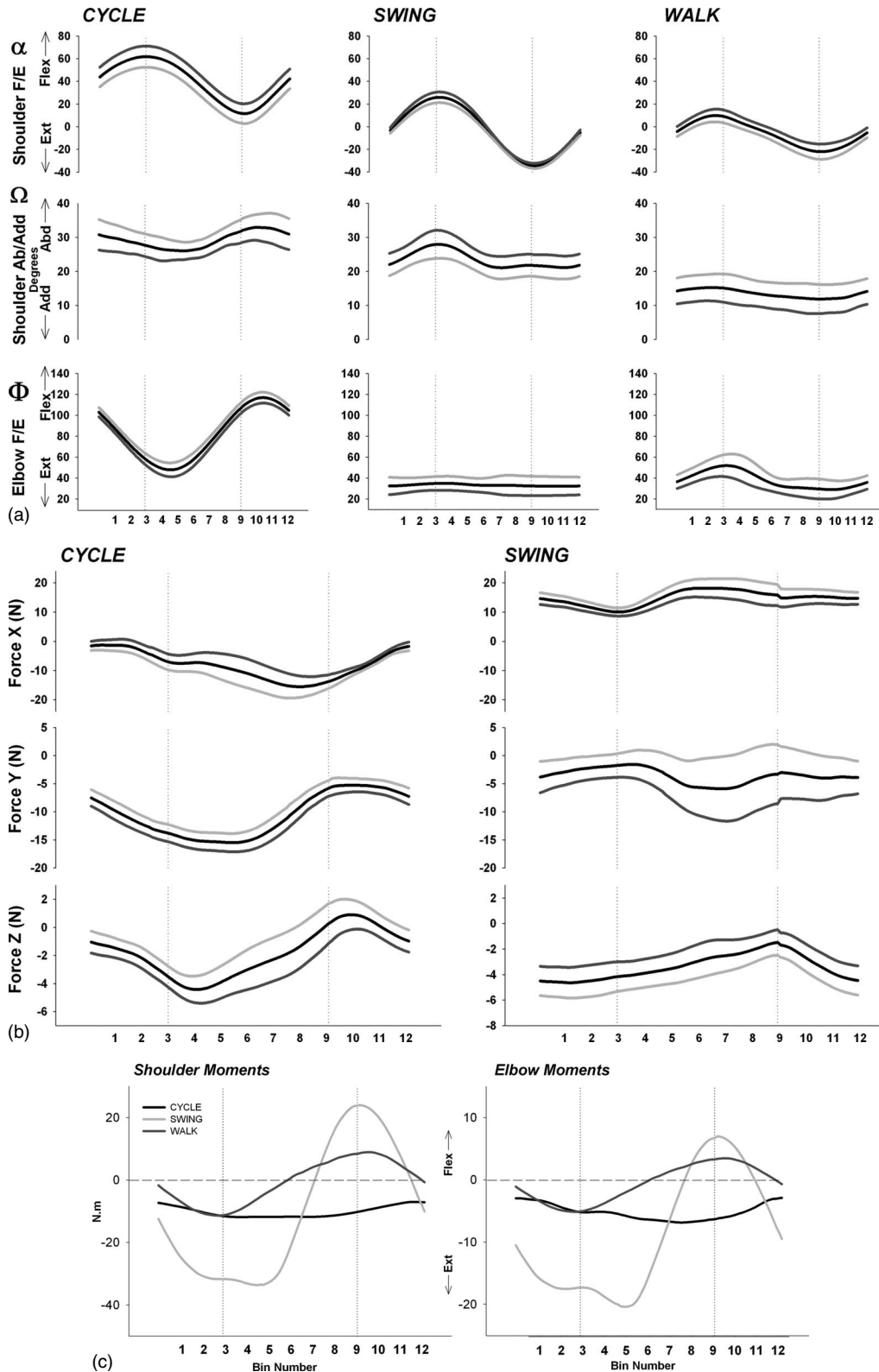


FIG. 6. (a) Kinematic changes at the right shoulder and elbow during arm cycling, swinging, and walking for shoulder flexion/extension (α), shoulder abduction/adduction (Ω), and elbow flexion/extension (Φ). (b) Kinetics measured at the right hand during arm cycling and swinging. (c) Moments calculated about the shoulder and elbow during arm cycling, swinging, and walking. Vertical dashed lines denote the 3 o'clock and 9 o'clock positions during the movement cycle, respectively.

onstrate both neural and mechanical differences between tasks. Specifically while the range of shoulder excursion is similar between tasks elbow flexion/extension kinematics are different. Additionally the anatomical “anchor” position of

the arms is different between the cycle task and the other two tasks that result in an altered moment noticed about the elbow and shoulder. As well, there are notable differences between the muscle activation patterns and amplitude that may

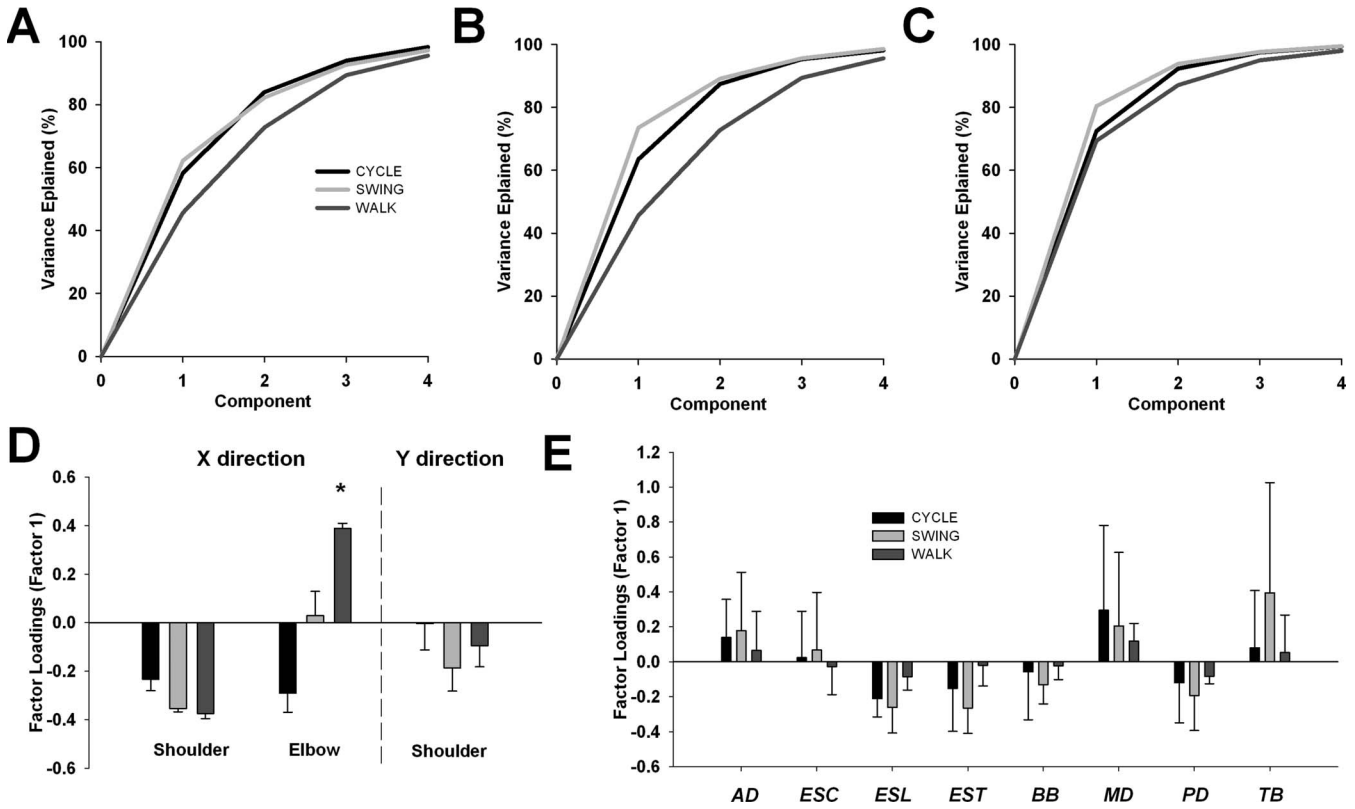


FIG. 7. (a) The averaged ($n=9$) cumulative percentage of the variance explained by each of the first four factors is shown for the three tasks for x -direction, (b) y -direction, (c) and EMG. (d) The average factor loadings for the first factor in the x -direction (flexion/extension) for shoulder and elbow and the average factor loadings for the first factor in the y -direction (adduction/abduction) for the shoulder for all tasks. (e) The average factor loadings for all muscles from the EMG PCA analysis. Asterisks denote significant differences between factor loading and the cycle.

reflect differences in mechanical assistance and constraints between tasks. Differences in task dependent neural control could reflect altered neural coupling to the lower limb. Below we present a neuromechanical description of the three tasks to ameliorate the study of details of arm to leg coupling that can assist in rehabilitation.

The principal component analyses clarify that the most prevalent mechanical difference is found in the pattern of elbow flexion/extension. The difference noticed between cycle and walk in the pattern of elbow flexion/extension demonstrates that the orientation of the limbs during walk is equivalent to a cycle movement in reverse as the pattern of elbow flexion/extension during cycle is the elbow in its most extended position during full flexion and the elbow is in its most flexed position full extension. This is different to walk

which has a peak elbow flexion at full flexion. This is maintained until midextension and then the elbow is most extended at full extension making a reverse cycling motion at the hand. This alteration in kinematics may require different muscle activation and subsequently neural control. However, forward and backward cycling have been shown to share common neural control with a CPG running in reverse through the analysis of reflex responses at functionally equivalent portions of the movement cycle (Zehr and Hundza, 2005). This concept is in agreement with similar results for forward and backward walkings and supports the idea of a common core of neural control despite mechanical differences (Grasso *et al.*, 1998; Zehr, 2005). Consequently despite common neural control for forward and backward movements functionally equivalent phases of the movement cycle may be different. This may account for the fact that the first factor in the EMG PCA analysis from cycle and swing was negatively correlated with walk while there were no differences between factor loadings for muscles. This would suggest an inverted neural signal with common drive to muscles across tasks. Thus, a functional context may allow proper comparison between tasks. Therefore, proper utilization of rhythmic arm movements to enhance locomotor activity requires knowledge on the basic premise of neural control set forth in the common core hypothesis (Zehr, 2005) backed by simple and elegant experimental design alongside detailed neuromechanical and functional comparisons.

TABLE I. The correlation coefficients (\pm standard error) of the first factor between tasks for separate x -kinematic, y -kinematic, and EMG PCA analysis. Asterisks denote significant difference between the cycle vs swing correlation.

| | Cycle versus swing | Cycle versus walk | Swing versus walk |
|-----------------|--------------------|---------------------|----------------------|
| X -kinematics | 0.712 ± 0.071 | 0.500 ± 0.148 | 0.688 ± 0.076 |
| Y -kinematics | 0.778 ± 0.036 | 0.5961 ± 0.13 | 0.645 ± 0.087 |
| EMG | 0.216 ± 0.162 | $-0.079 \pm 0.13^*$ | $-0.227 \pm 0.154^*$ |

Below we discuss other relevant neuromechanical considerations that may assist in the development and testing of rehabilitative tasks.

A. Neuromechanical considerations for the use of arm movement to facilitate leg muscle activity during locomotor activities

1. Arm swing while walking (walk)

During the arm swing of walking, our calculated moments agree with previous work of Hinrichs and Elftman concerning shoulder and elbow kinetics (Elftman, 1939; Hinrichs, 1990). Starting from midflexion there is a net extensor moment when the shoulder is approaching a maximum flexion (contralateral heel strike) which would tend to slow forward swing and initiate the backward swing. This moment is consistent with heightened PD and MD EMG during forward swing (Ballesteros *et al.*, 1965; Hinrichs, 1990; Hogue, 1969; Jackson, 1983). These muscles could assist in braking the forward momentum of the arm and initiate backward swing as well as producing abduction or resisting adduction. Additionally, it has been suggested that movement-related neural signals from this arm position may provide a shaping effect on lower limb muscle activity (Kawashima *et al.*, 2008). Afferent feedback from muscle, cutaneous, and joint receptors may be the source of this neural signal. The next major shoulder moment occurs just prior to and during shoulder extension (ipsilateral heel strike) where there is a net flexor moment which would tend to slow the backward swing and initiate forward swing (Hinrichs, 1990). Anatomically the arm has a limited range of extension in this position. Therefore at this position passive tension could produce heightened afferent signals from muscle spindle and tendon receptors in shoulder flexor muscles as well as cutaneous receptors over the anterior surface of the upper arm (Hiraoka, 2001; Kawashima *et al.*, 2008). In agreement with previous studies there is very little activity in AD at this position and it is possible that latissimus dorsi and teres major assist with active transition during extension because they are active at this phase of swing (Hinrichs, 1990). PD is active just following full extension which could offset the force at ipsilateral heel strike and concurrently could reflect an important neuromechanical strategy. For example, as the arm could be considered a mechanical pendulum with a resonant frequency, passive tension in the musculoskeletal system along with muscle activation (PD) could entrain the arm to move at a different frequency (Webb *et al.*, 1994). Therefore activity in PD could be responsible for resetting the relative frequency at this point. This potential delay in movement produced by PD would also slow the forward acceleration of the upper arm in position while allowing the lower limb to move forward under its own inertia. This would change the moment of inertia and virtual pendulum length, thus altering the resonant frequency of the whole system (Preuschoft and Witte, 1991). This concept is in agreement with the idea that a CPG coupled to a mechanical system can alter the resonant

frequency of the whole system (Ijspeert, 2008; Taga, 1998; Verdaasdonk *et al.*, 2006). Therefore full extension may be an important landmark for studying neural control during walking as it may have a crucial role in frequency entrainment of the system.

Walk is the only task in this study where whole body momentum may mechanically affect the movement of the limb and thereby change the functional role of the muscles in the movement. This could be related to the function of arm swing during walking, which has been shown to effect whole body moment of inertia to assist balance regulation as well as to offset transverse forces, angular momentum, and free vertical moments (Elftman, 1939; Hinrichs, 1990; Li *et al.*, 2001; Misiaszek and Krauss, 2005; Umberger, 2008). Overall the mechanical benefits of arm swing while walking are thought to arrive through a clear active component interacting with the passive motion of the limb, both of which are considered necessary to control the limb to produce smooth locomotion (Jackson, 1983).

2. Arm cycling (cycle)

During cycle there is a consistent extensor moment at the shoulder throughout the movement that is highest during extension and smallest during flexion. During early flexion there is a decrease in the extensor moment as the shoulder is flexing and adducting while the elbow is extending and following the forward and upward paths of the cycle. This is concurrent with a decreased downward force on the handle and activity in AD and MD. These muscles would act together to assist in flexion and resist adduction of the shoulder. In late flexion there is an increase in the extensor moment which reaches a plateau as the shoulder is approaching peak flexion. Here, there are diminishing AD, PD, and MD activities as the arm is following the downward and forward movement paths of the crank arm. Peak flexion is the position at which the center of mass of the arm is producing the greatest moment about the shoulder. Hinrichs (1990) used this arm positioning to illustrate the relative magnitude of the moment about the shoulder. The magnitude of the current values is in agreement with these calculations suggesting that the position of the arm is responsible for much of the moment at this portion of the cycle. In early extension the moment is maintained as the elbow is flexing and producing an extensor moment about the shoulder. Thus even though the moment arm from the center of mass of the whole arm segment is diminishing and decreasing its contribution to the shoulder moment, the change in orientation of the elbow is resulting in an increasing extensor moment about the shoulder that is highest during peak extension. As well there are increased backward and downward forces on the handle that are congruent with the movement path. Also, there are increasing AD, MD, BB, and ESC activities which could serve to slow the backward movement of the arm. During late extension the extensor moment is maintained as there is a peak extensor moment at the elbow as the shoulder extends and abducts. As well there is a peak backward force with diminished downward force on the handle consistent with the upward and backward movement paths. Here, there is an increase in AD, MD, BB, and ESC activities. The combined

activity of these muscles could serve to stabilize the limb (ESC), slow the backward swing of the upper arm while assisting abduction and initiating forward swing (AD, MD), all while lifting the forearm, hand, and cycle ergometer through the movement path (BB). One would assume that because the moment arm from the center of gravity of the whole limb to the shoulder joint is smallest at this point this would diminish the extensor moment about the shoulder. However during late extension is also where the direction of the movement path of the ergometer is upward and toward the body, thus producing a large extensor moment about the shoulder because the arm must be lifted congruent with this path and against gravity. Thus there is a balance between the orientation of the limb and the movement path direction with respect to gravity. Overall, a consistent flexor moment would be necessary to resist the gravitational moment due to the anatomical position of the limb. As mentioned above, during arm swing while walking, transition points such as peak flexion and extension may present areas where afferent feedback may be heightened. However the constant extensor moment may present a source of afferent feedback that reduces or increases the effectiveness of other afferent signals. The cycle task could effectively be altered by making an apparatus that allows cycling beside the body in the same shoulder excursion range as walk. AD and MD activities are highest during late flexion and early extension during cycling to properly halt the backward progression and initiate forward progression. This is qualitatively different to walk which has greater activation of PD and MD during peak flexion. Therefore in terms of phases where the greatest propulsion to the movement occurs these seem to be opposite between the two tasks in reference to the shoulder muscles which could relate to the regulation of forward and backward cycling, control variable, or anchor position differences between tasks (Zehr *et al.*, 2007a). The control variable idea suggests that the kinematic control of the task is constructed to produce muscle pattern that relates to a specific mechanical parameter (Grasso *et al.*, 1998; Zehr *et al.*, 2007a). During cycle and swing arm swing is a manipulation of limb orientation to produce force at the hand interface to propel the hand through the prescribed movement path, and therefore the control variable would be the trajectory of the hand. This contrasts with walk where the manipulated variable is the moment of inertia of the whole arm (Preuschhof and Witte, 1991). Thus control variable differences may account for the differences in muscle activation. This is similar to a comparison of forward and backward walkings where the kinematic parameters were conserved despite substantial differences in energy cost (Grasso *et al.*, 1998). Differences in the anatomical anchor position could require differential activation of muscles that cross the shoulder joint. This is because changing kinematics alters the moment producing capacity of muscles about a joint due to force-length relationships and differences of line of action (Klimstra and Zehr, 2006). Thus tasks may require different optimal muscle activation profiles (neural drive) based on anchor position. For example, during cycle, AD is at a shortened length and PD lengthened compared to walk and swing because the arm is in front of the body midline. This would change the relative contribution

that each muscle could have to the movement. However, we did not measure all muscles that could contribute to the movement and it is important to measure from all possible contributing muscles before a true discussion of control variable and anchor position can be made.

Another interesting feature of cycle is that the movement of the apparatus required no z -force because of its planar orientation. However there were small but measurable changes in the z -force at the flexion/extension transitions consistent with adduction and abduction of the shoulder. Thus there may be inefficient movement due to the anatomical constraints. The mechanical assistance from the rest of the body is purely to stabilize the upper body and limbs during the movement. This form of assistance is different from that observed during walk. However, the arm cycle ergometer used in this experiment had an inertial load that could provide mechanical assistance to the movement and could change the neural feedback and strategy. This could be tested by altering the inertial parameters of the arm cycle or inducing mechanical assistance through a servomotor attached to the crank (Kautz *et al.*, 2006; Ting *et al.*, 1998; Ting *et al.*, 1999; Ting *et al.*, 2000). The interactions of anatomical position, constraints, inertial load, and movement path present possible interactions that may affect neural control.

3. Arm swing alone (swing)

During arm swing there are two noticeable moments occurring near peak flexion and extension similar to walk. From late flexion to early extension there is an extensor moment when the arm is transitioning from flexion to extension while abducting. This occurs alongside only relative low level PD and MD activities, suggesting that the transition could be a function of the inertia of the arm or other musculature not measured in this study such as latissimus dorsi and teres major (Ballesteros *et al.*, 1965; Hinrichs, 1990). Although this movement does not exhibit the same muscle activation as walk there may be similar peripheral feedback due to the anatomical position and the range of motion. There is a flexion moment during late extension and early flexion to slow the backward progression of the limb and initiate forward progression. This occurs with a peak negative z -force, suggesting an interaction between the desired movement path and the prescribed movement path induced by the apparatus similar to cycle. During late extension early flexion there is an increasing AD activity that would be consistent with activity to break backward progression and initiate forward progression. It is important to note that due to apparatus, throughout the movement there is a very little elbow extension (less than 5°). Therefore the upper arm, forearm, and hand would behave as a rigid segment. The added mass of the forearm and hand as well as the apparatus could add a considerable moment about both the elbow and shoulder. This is noticed in the increased shoulder and elbow moments compared to walk. Additionally during extension to flexion there is a noticeable increase in BB and TB to stabilize the joint against the torque. The activity of BB during this stage could also serve to accelerate the shoulder into flexion as it crosses both the elbow and the shoulder joint.

The increased moment may produce a strong afferent signal during this phase of the movement. As swing is similar to walk in the anatomical position of the limb and the basic motion it is surprising to see that the muscle activity is different. However there are differences noticed in the anatomical range as well as shoulder abduction/adduction and elbow flexion/extension due to the experimental setup. These could effectively alter the muscle activation profile. For example, Ballesteros *et al.* (1965) showed that during arm swing alone at different excursions muscle activation changed from only PD activity at less than 30° to include activity from AD and MD at excursions greater than 30°. This was also noticed when comparing the self-selected excursion trials in the current study. Thus it could be the difference in the shoulder excursion range between walk and swing that produces differential activation of muscles.

VI. FUNCTIONAL IMPLICATIONS AND CONCLUSION

The current study presents a neuromechanical characterization of three rhythmic arm movement tasks. This attempt to highlight rhythmic arm movement could be used to assist in the determination of the necessary and sufficient conditions for arm induced lower limb activation in locomotor rehabilitation in both stroke and spinal cord injury. Through factor analyses these tasks were shown to have different mechanics with conserved central neural control. Interestingly, the three arm tasks can be considered to share the same basic rhythmic pattern except that the anchor point about which the arm oscillates in the three tasks is shifted [shown clearly in Fig. 6(a)]. This allows for conservation of neural control parameters while still allowing for incorporating altered mechanics that will affect afferent feedback and supraspinal contributions. Additionally the movement state of the arms and the legs, the phase relationship between the limb pairs, mechanical coupling, and augmented afferent feedback could enhance or reduce the effect of arm movement on lower limb activation (Ferris *et al.*, 2006; Zehr *et al.*, 2009). Incorporation of parameters related to arm movement in the restoration of leg activation after neurotrauma is the next step. It remains to test the effect of different sources of afferent feedback as well as altering existing apparatus to assist in the development and testing of a functional task to assist neuronal coupling between the arms and legs in a manner consistent with locomotor rehabilitation.

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