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The Effect of Movement Frequency on Interlimb Coupling During Recumbent Stepping

Pei-Chun Kao and Daniel P. Ferris

During passive lower limb movement, active use of the upper limbs increases unintentional lower limb muscle activation. We hypothesized that faster movement frequencies would amplify lower limb muscle activation during upper limb exertion but would not affect lower limb muscle activation when the upper limbs were relaxed. We studied 10 healthy participants exercising on a recumbent stepping machine that mechanically coupled the four limbs via handles and pedals. Participants exercised at four frequencies (30, 60, 90, 120 steps/min) under four conditions of active and passive movement. Self-driven lower limb motion resulted in greater muscle activation compared to externally driven lower limb motion. Muscle activation amplitude increased with frequency for all conditions except for externally driven stepping. These results indicate that fast upper limb movement facilitates neuromuscular recruitment of lower limb muscles during stepping tasks. If a similar effect occurs in neurologically impaired individuals during active stepping, self-assisted exercise might enhance neuromuscular recruitment during rehabilitation.

Key Words: gait, locomotion, rehabilitation, interlimb coordination, electromyography

Body weight-supported treadmill training with manual assistance is a promising approach in gait rehabilitation for neurologically impaired patients. Previous studies have shown that this therapy can substantially improve walking capacity of patients with spinal cord injury or stroke (Dietz & Harkema, 2004; Hesse et al., 1995; Wernig, Muller, Nanassy, & Cagol, 1995; Wernig, Nanassy, & Muller, 1998). A disadvantage of manually assisted locomotor training is that it is highly labor intensive, requiring multiple therapists to assist one patient. The expensive labor costs impede its implementation in neurological rehabilitation clinics. Furthermore, the nature of external manual assistance makes it possible for patients to receive more assistance than is necessary. To enhance activity-dependent plasticity,

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patients must be as active as possible (Wolpaw & Tennissen, 2001). An ideal gait therapy would allow patients to practice stepping actively on their own without direct therapist assistance.

Self-assisted exercise devices could allow neurologically impaired patients to facilitate active rehabilitation. An example of this type of device is a recumbent stepping machine that has mechanically coupled handles and pedals (Figure 1). Because of the coupling, patients can use their arms or legs to drive the stepping motion. For example, individuals with paraplegia could use their own arms to assist their legs to perform the stepping motion. This type of device might be beneficial to rehabilitation because it facilitates neural coupling between the upper and lower limb (Dietz, 2002; Zehr & Duysens, 2004).

A recent study (Huang & Ferris, 2004) examined healthy participants using a recumbent stepping machine and found evidence for neural interlimb coupling. Participants could not completely relax their lower limb muscles when they used

Figure 1 — Recumbent stepping machine.

their upper limbs to drive passive lower limb movement. In contrast, lower limb muscle activation was significantly lower when the participants relaxed their arms and let a second person drive passive lower limb movement. Importantly, increased lower limb muscle activation during passive lower limb movement was also apparent if participants actively used their arms to drive handles that were mechanically decoupled from the pedals (Huang & Ferris, 2004). Thus, the greater lower limb muscle activation could not be attributed to participants' desire to reduce the work performed by the upper limbs. The neural mechanism for increased lower limb muscle recruitment with active upper limb exertion is not known, but a probable explanation are spinal connections in locomotor neural networks (Dietz, 2002; Huang & Ferris, 2004; Zehr & Duysens, 2004).

The purpose of this study was to determine how movement frequency affects neural coupling between upper and lower limbs during recumbent stepping. An important aspect of the recumbent stepping machine specifically, and self-assisted devices in general, is that neurologically impaired individuals can increase the frequency of the stepping motion beyond what they are capable of with legs alone. Recent studies have found that faster stepping frequencies during manually assisted locomotor training can substantially improve gait function (Pohl, Mehrholz, Ritschel, & Ruckriem, 2002; Sullivan, Knowlton, & Dobkin, 2002). To provide insight into the effects of stepping frequency on neural interlimb coupling, we tested healthy participants as they performed stepping movements on a recumbent stepping machine over a range of stepping frequencies. We hypothesized that increasing the frequency of the stepping motion would increase lower limb muscle activation during self-driven stepping but would have no effect during externally driven stepping. We based this hypothesis on the premise that the neural interlimb effects during self-driven stepping should be dependent on the amount of upper limb motor neuron recruitment (Huang & Ferris, 2004). In contrast, any lower limb muscle activation occurring during passive movement of the lower limbs without neural facilitation from the upper limbs (i.e., externally driven stepping) should not be dependent on movement frequency. Healthy individuals can completely relax their upper limbs during passive movement at a range of speeds (Ibrahim, Berger, Trippel, & Dietz, 1993) so it would seem likely that they could also relax their lower limbs as well.

Materials and Methods

Participants

Ten healthy young adults (five females and five males; age range 19–29 years) participated in the study after giving written informed consent. The University of Michigan human subjects protection office approved the protocol and consent form.

Experimental Design

The recumbent stepping machine (TRS 4000 recumbent cross trainer, NuStep Inc., Ann Arbor, MI) (Figure 1) has handles and pedals that are all mechanically coupled in normal locomotor phase relationships (e.g., left arm and right leg move forward while right arm and left leg move backward) and can be driven by pedals or handles. A video depicting the recumbent stepper in motion is available at the company website, www.NuStep.com. The stepping machine used an eddy current disc brake to provide stepping resistance. A handle on the machine adjusted the resistance level by moving an array of magnets closer to or farther away from the disc. The movement of the magnets changed the interaction volume between the magnetic field and the conducting material, adjusting resistance to disc rotation.

Participants used the recumbent stepping machine at four different frequencies (30, 60, 90, 120 steps/min) following the beat of a metronome. There were four conditions: (1) Arms-Legs, active stepping using both arms and legs; (2) Legs only, active stepping using legs only with participants putting their arms on their lap; (3) Self-driven, using arms to propel the stepping motion and allowing legs to move passively on the pedals; and (4) Externally-driven, arms on the lap with the stepping motion propelled by another person moving the handles. A previous study (Huang & Ferris, 2004) found similar results for the Externally driven condition whether the participant's arms were on their lap or moving passively with the handles. We collected data for the first two conditions (i.e., active stepping) to provide a comparison of muscle activation timing for the passive conditions. Mechanical stops on the recumbent stepper limited the range of motion for the pedals and handles. For all conditions, participants were instructed to move through the full range of motion until they made contact with the stops.

Participants adjusted the machine's handles and seat positions so that their knee joints were just short of full extension. Prior to data collection, participants stepped with a range of resistance levels (1-10, as set by the machine's controls) to self-select the greatest resistance they could maintain using their arms only (self-driven) at 120 steps/min for at least 20 s. The data collection period used, 20 s, allowed collection of sufficient step cycles without promoting fatigue. Participants quickly became comfortable using the machine so no long-term training was needed. Participants performed stepping trials for arms-legs, self-driven, and externally driven conditions at this resistance. For the legs only condition, participants stepped at a resistance level that was half the resistance level of the other conditions (rounded up to the nearest whole number). This was done so that legs only stepping would have muscle activation amplitudes that were more similar to the muscle activation amplitudes during arms-legs stepping.

Before each trial, participants were given time to practice. For self-driven and externally driven conditions, participants verbally confirmed that they were not intentionally using their lower limbs and were attempting to keep all lower limb muscles completely relaxed. Participants completed each condition and frequency twice in a random order. We recorded trials at 30 steps/min for 20 s and trials at other frequencies for 10 s, always after participants had been stepping for at least 10 s. Three consecutive step cycles from each trial were randomly selected for analysis.

Data Acquisition and Analysis

Goniometers. We placed six electrogoniometers (Biometrics, Ltd., Ladysmith, VA) on the participant's lower limbs to record hip, knee, and ankle joint angles bilaterally. The computer sampled electrogoniometer data at 1,000 Hz.

Electromyography. We placed surface electrodes on both lower limbs to record surface electromyography (EMG) (Konigsberg Instrument, Inc., Pasadena, CA) from six muscles on each limb: vastus lateralis, vastus medialis, medial hamstrings (semimembranosus and semitendinous), medial gastrocnemius, soleus, and tibialis anterior. Electrode sites were prepared by shaving and rubbing the skin with alcohol to ensure good contact. We placed EMG electrodes (diameter 1.1 cm, interelectrode distance 3.5 cm) longitudinally over the approximate center of each muscle belly for all muscles except the soleus. For the soleus, we placed electrodes over the distal third of the muscle belly to avoid crosstalk from the gastrocnemius. Participants then put elastic netting over both lower limbs to minimize movement artifacts. A personal computer sampled each EMG channel at 1,000 Hz via an A/D board. We verified that crosstalk from adjacent muscles and movement artifact did not substantially interfere with the signal through visual inspection of the EMG during manual muscle tests (Winter, Fuglevand, & Archer, 1994).

EMG signals were filtered with a fourth-order high-pass Butterworth filter with zero phase lag (cutoff frequency 20 Hz) and then full wave rectified the signals. To quantify muscle activation during stepping, we calculated root-mean-square (RMS) amplitude of the filtered, rectified EMG over the entire step cycle. In a previous study, we divided the EMG into flexion and extension activity based on the kinematics of the step cycle (Huang & Ferris, 2004). In that study, the results for EMG over the entire step cycle were no different for results examining EMG for flexion and extension phases separately. Based on those findings, we chose not to divide EMG into flexion and extension phases for this study because it would have artificially altered the results. EMG activity occurred for a similar percent of the stepping cycle at all frequencies. At faster movement frequencies, the onset of muscle activity occurs earlier in the cycle period relative to the kinematics because the overall cycle period is shorter. As a result, dividing EMG activity into flexion and extension phases would cause an artificial increase in one phase and a decrease in the other phase at different frequencies. For each condition and frequency, we averaged six step cycles of data (three step cycles for two trials). We normalized EMG RMS values to the maximum value recorded across all conditions for each muscle in each participant to reduce interparticipant variability (Yang & Winter, 1984).

Cross correlations were used to calculate the temporal relationship of muscle activation between Arms-Legs and Self-driven conditions (Li & Caldwell, 1999). Using low-pass filtered EMG averages (cutoff frequency 6 Hz), we calculated the time lag at which the absolute value of the correlation coefficient was maximal for each muscle. We used *t* tests to determine if the mean time lags between the two conditions were significantly different from zero.

Compression Load Cells. We mounted three compression load cells (LCWD-1000, Omegadyne, Inc., Sunbury, OH) between two aluminum plates on each foot pedal to measure foot-pedal contact force during stepping. Using three load cells for each footplate allowed for a stable configuration. Before each data collection, we calibrated each load cell with known weights. Load cell data was collected at 1,000 Hz and added values for the three load cells under each foot. The summed load cell data were then filtered with a fourth-order low-pass Butterworth filter (cutoff frequency: 6 Hz) with zero phase lag to remove movement artifact. After averaging six step cycles of data, we calculated mean pedal force for both limb flexion and limb extension phases of the step cycle. Mean pedal force values were normalized to the maximum value recorded across all conditions so that we could decrease inter-participant variability comparing pedal forces across participants.

Statistics

We used a repeated measures factorial ANOVA to assess the effects of frequency and condition (4 frequencies × 4 conditions) on the dependent variables (EMG RMS amplitude and pedal force data). Because we had to use six repeated measures ANOVAs to test EMG data from the six muscles, we set the significance level of the ANOVAs at $p < .0083$ using Bonferroni correction (Vincent, 1995). We used Tukey honestly significant difference (THSD) tests for post hoc comparisons ($p < .05$) if ANOVA results indicated a significant effect.

Results

Participants produced the required step frequencies very well under the different conditions. During the arms-legs stepping, the actual frequencies (mean/*SD*) performed by participants at the frequencies of 30, 60, 90, and 120 steps per minute were 30.0 (0.3), 59.7 (0.8), 89.7 (1.3), and 118.8 (2.5) steps/min, respectively. During the legs-only condition, the frequencies performed were 29.8 (0.7), 59.8 (0.9), 90.8 (1.8), and 119.4 (1.2) steps/min, respectively. During the self-driven condition, the frequencies were 30.0 (0.3), 60.4 (0.9), 91.5 (2.7), and 117.6 (3.6) steps/min, respectively. During the externally driven condition, the frequencies were 30.0 (0.2), 59.9 (0.4), 90.2 (1.4), and 118.5 (2.7) steps/min, respectively.

Electrogoniometer data indicated that joint angles during recumbent stepping for all participants were similar. This was expected given that mechanical stops limited the range of motion for handles and pedals and participants moved through the full range for all conditions. The average range of motion for hip joint was $44^\circ \pm 4^\circ$ to $73^\circ \pm 5^\circ$ of hip flexion (for all three joints, 0° was standing posture). The average range of motion for knee joint was $27^\circ \pm 2^\circ$ to $81^\circ \pm 3^\circ$ of knee flexion. The average range of motion for ankle joint was $8^\circ \pm 2^\circ$ of ankle plantar flexion to $13^\circ \pm 3^\circ$ of ankle dorsiflexion (mean ± *SE*).

Self-driven lower limb motion resulted in greater muscle activation compared to externally driven lower limb motion (Figures 2 and 3). For all six muscles, Self-driven EMG amplitude was significantly higher than externally driven EMG amplitude (THSD, $p < .05$). The smallest differences between the two passive conditions occurred in the vastus lateralis muscle (Figure 3). At 120 steps/min, vastus lateralis EMG amplitudes for self-driven and externally driven conditions were 20 ± 12% and 8 ± 6%, respectively (mean ± *SD*). The largest differences between the two passive conditions occurred in the tibialis anterior muscle (Figure 3). At 120 steps/min, the values were 66 ± 36% and 15 ± 6%, respectively. In all muscles, the difference between the conditions increased at higher step frequencies (Figure 3).

Muscle activation increased with frequency in all six muscles for all conditions except externally driven stepping (THSD, $p < .05$). For self-driven stepping, the largest increase in EMG amplitude over the four frequencies was in tibialis

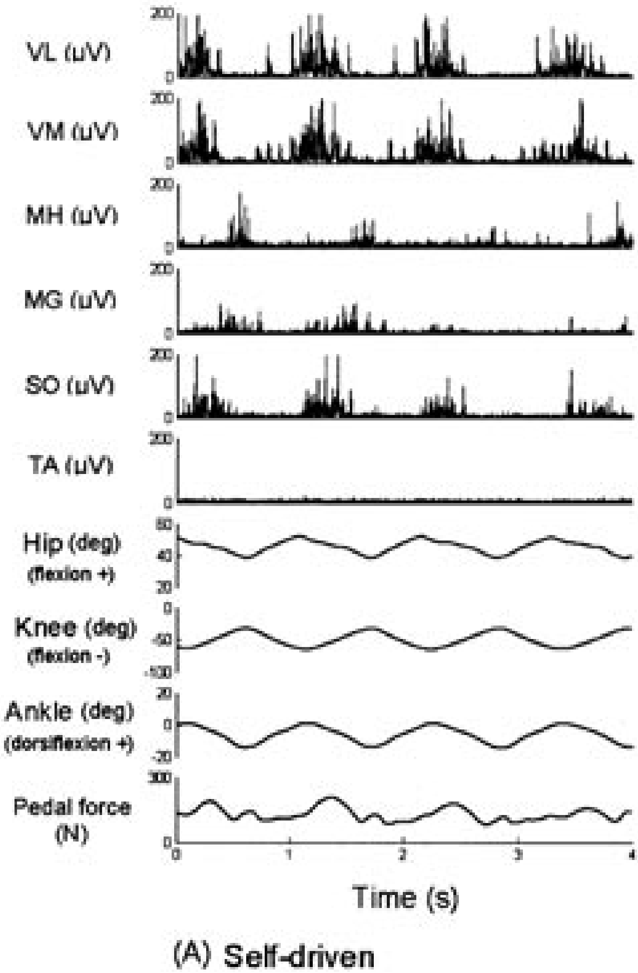

Figure 2 — Rectified EMG, joint angle, and pedal force data from one participant at 120 steps per minute during (A) Self-driven and (B) Externally-driven conditions. Recordings are from the vastus lateralis (VL), vastus medialis (VM), medial hamstrings (MH), medial gastrocnemius (MG), soleus (SO), and tibialis anterior (TA).

anterior (Figure 3). Muscle activation in tibialis anterior increased from 10 ± 5% at 30 steps/min to 66 ± 36% at 120 steps/min. The smallest increase in EMG amplitude over the four frequencies for self-driven stepping was in vastus lateralis (Figure 3). Muscle activation in vastus lateralis increased from 5 ± 4% at 30 steps/min to 20 ± 12% at 120 steps/min.

EMG amplitude during the externally driven condition did not show a significant difference with frequency (THSD, $p > .05$). There was the possibility that

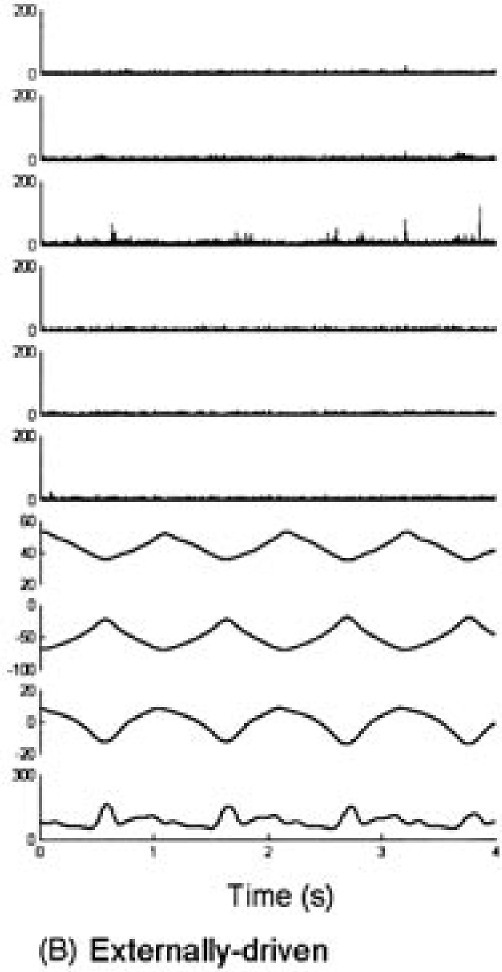

Figure 2 — *(continued)*

we did not detect a change in muscle activation at different frequencies resulting from a type II error. Even if there was a real change that we did not detect, however, it was a small effect. For externally driven stepping, the mean muscle activation amplitude for the six muscles increased from 7 ± 2% at 30 step/min to 13 ± 6% at 120 steps/min, less than a twofold change. In contrast, the respective values for self-driven stepping were 9 ± 2% at 30 steps/min and 36 ± 18% at 120 steps/min, a fourfold change. The muscle that had the greatest change in activation from 30 to 120 steps/min during externally driven stepping was the medial hamstrings. It went from 8 ± 4% at 30 steps/min to 32 ± 13% at 120 steps/min. None of the other muscles even doubled their activation from 30 to 120 steps/min during externally driven stepping.

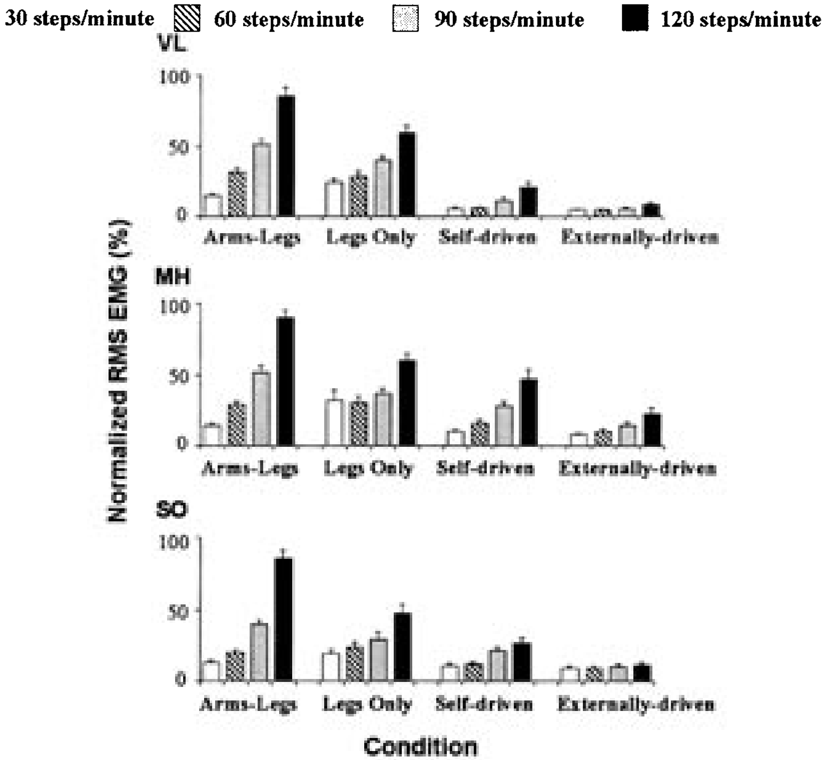

Figure 3 — Normalized root-mean-square EMG for all participants. Stepping actively with both arms and legs (Arms-Legs) had the highest EMG amplitudes of the four conditions while the passive Externally-driven condition had the lowest EMG amplitudes of the four conditions. Repeated measures ANOVA indicated significant differences by condition and frequency ($p < .0001$) for all six muscles. Post hoc tests on condition indicated that Self-driven EMG amplitudes were significantly higher than Externally-driven EMG for all muscles (THSD, $p < .05$). In addition, Arms-Legs, Legs Only, and Self-driven conditions all showed a significant increase in EMG amplitude with frequency (THSD, $p < .05$), but Externally-driven did not (THSD, $p > .05$). Increasing movement frequency did not increase EMG amplitude when the lower limbs were passive and someone else controlled the movement. For vastus lateralis (VL) and vastus medialis (VM) muscles, all frequencies (30, 60, 90, and 120 steps per minute) were significantly different from each other (THSD, $p < .05$). For medial hamstrings (MH), medial gastrocnemius (MG), soleus (SO), and tibialis anterior (TA), all frequencies were significantly different except 30 and 60 steps per minute.

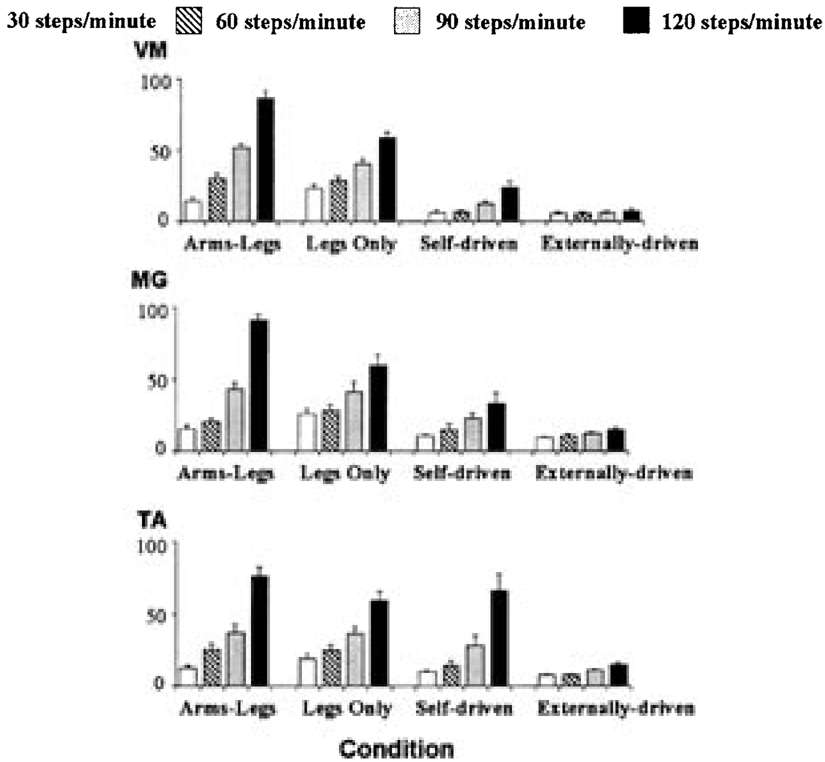

Figure 3 — *(continued)*

Muscle activation patterns during self-driven passive lower limb movement were similar to patterns during active stepping (Figure 4). The time lags at peak correlation between self-driven and arms-legs conditions were not significantly different from zero ($p > .05$; Table 1). Thus, there was no significant difference in muscle activation timing between self-driven and active stepping.

Extension phase pedal forces were much higher during the active stepping conditions than during the passive stepping conditions (Figure 5). Flexion pedal forces were not very different for any of the stepping conditions. Comparing just the passive conditions, self-driven stepping had slightly smaller flexion phase pedal forces than externally driven stepping (THSD, $p < .05$). In the extension phase, there was no significant difference in pedal force between self-driven and externally driven (THSD, $p > .05$). Figure 6 shows averaged pedal force over the complete step cycle at all four frequencies. The self-driven condition had similar force patterns as the externally driven condition during 30 and 60 steps/min. At 90 and 120 steps/min, the self-driven condition showed noticeable increases in force during the extension phase.

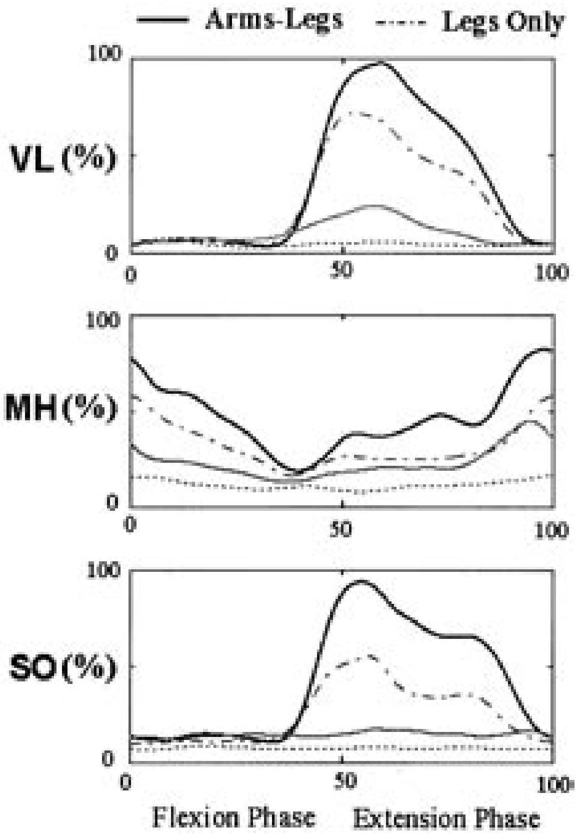

Figure 4 — Averaged low-pass filtered EMG profiles at 120 steps per minute for all participants. Self-driven muscle activation patterns were very similar for active stepping with arms and legs (Arms-Legs) and active stepping with only the lower limbs (Legs Only). When there were clear increases in muscle activation for the passive Self-driven condition, they occurred with similar timing as in the two active conditions. The passive Externally-driven condition showed virtually no increases in muscle activation throughout the stepping cycle. Data were normalized to the peak linear envelope amplitude for the four conditions.

Discussion

Passive self-driven lower limb motion resulted in greater muscle activation than externally driven lower limb motion. This confirms previous findings (Huang & Ferris, 2004) and supports the existence of neural pathways coupling upper and lower limb muscle activation during human locomotion (Dietz, 2002; Zehr & Duysens, 2004). One of the novel findings from the present study was that movement frequency enhanced the unintended muscle activation in the passive lower limbs.

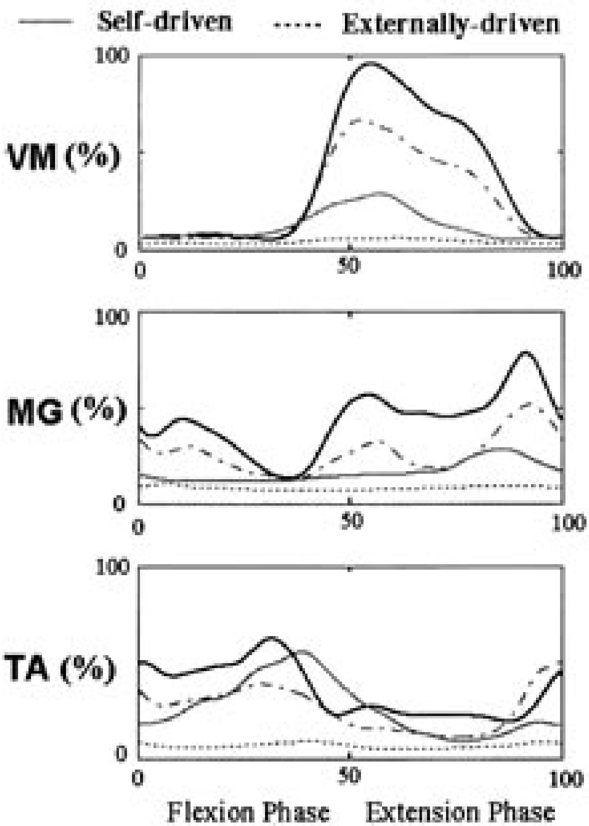

Figure 4 — *(continued)*

This likely occurred because of greater recruitment of upper limb motor neurons. A previous study (Huang & Ferris, 2004) found that increasing upper limb resistance at the same frequency resulted in greater lower limb neuromuscular recruitment. A second novel finding of the present study was that lower limb muscle activation did not increase with higher movement frequencies during externally driven passive lower limb movement. This indicates that even fast passive movement by itself is not sufficient to produce substantial recruitment of lower limb motor neurons. A possible exception is the medial hamstrings, as it showed a fourfold increase in muscle activation across movement speeds during externally driven stepping. This observation is in agreement with other studies that suggest stretch reflex activation contributes substantially to hamstrings EMG during human walking (Duysens, van Wezel, van de Crommert, Faist, & Kooloos, 1998; Van de Crommert, Faist, Berger, & Duysens, 1996).

Table 1—Cross-Correlation Coefficients and Time Lags for Six Muscles Between Arms-Legs and Self-Driven Conditions at Different Frequencies

	VL		VM		MH		MG		SO		TA	
Frequency	***r***	**Lag (% cycle)**	***r***	**Lag (% cycle)**	***r***	**Lag (% cycle)**	***r***	**Lag (% cycle)**	***r***	**Lag (% cycle)**	***r***	**Lag (% cycle)**
30 bpm	0.87	1	0.89	0	0.92	-5	0.97	0	0.97	0	0.94	-3
	(0.06)	(2)	(0.07)	(0)	(0.07)	(14)	(0.05)	(0)	(0.03)	(0)	(0.09)	(10)
60 bpm	0.82	1	0.83	1	0.92	0	0.90	0	0.91	1	0.90	-1
	(0.11)	(4)	(0.10)	(2)	(0.04)	(0)	(0.07)	(0)	(0.05)	(4)	(0.11)	(2)
90 bpm	0.88	7	0.90	2	0.93	0	0.90	0	0.87	1	0.91	-1
	(0.11)	(15)	(0.08)	(5)	(0.04)	(0)	(0.03)	(1)	(0.08)	(12)	(0.06)	(6)
120 bpm	0.94	4	0.94	5	0.90	0	0.88	0	0.84	1	0.93	-4
	(0.05)	(5)	(0.04)	(5)	(0.05)	(0)	(0.08)	(3)	(0.08)	(8)	(0.06)	(6)

Note. r, mean (*SD*) peak coefficient of cross correlation; Lag, mean (*SD*) time lag (% of step cycle) at peak correlation coefficient;
Negative value of time lag means that the muscle activations in Self-driven condition started earlier than those in Arms-Legs condition.

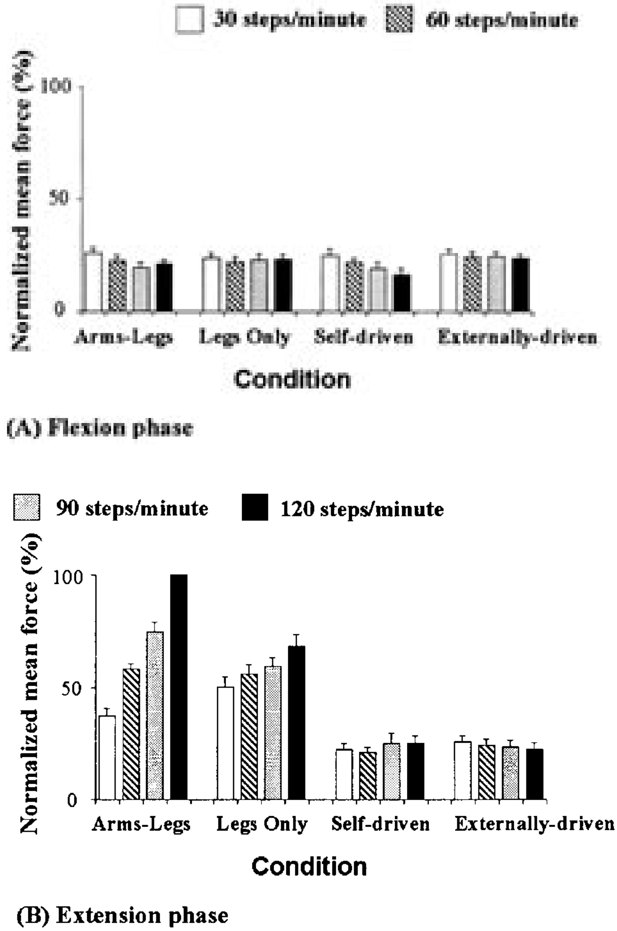

Figure 5 — Normalized mean pedal force in flexion and extension phases for all participants. (A) During the flexion phase, mean pedal forces were fairly consistent but there were small significant differences by condition and frequency (ANOVA, $p < .0001$). Post hoc tests on frequency revealed that 30 steps per minute was significantly different from the other frequencies (60, 90, and 120 steps per minute) (THSD, $p < .05$). Comparing the two passive conditions, Self-driven and Externally-driven conditions were significantly different though differences were small (THSD, $p < .05$). (B) During the extension phase, pedal forces were much higher for the two active conditions (Arms-Legs and Legs only) than for the two passive conditions (Self-driven, Externally-driven). Repeated measures ANOVA revealed significant differences by condition and frequency ($p < .0001$). All frequencies (30, 60, 90, and 120 steps per minute) were significantly different from each other (THSD, $p < .05$). Self-driven and Externally-driven conditions were not significantly different from each other (THSD, $p > .05$).

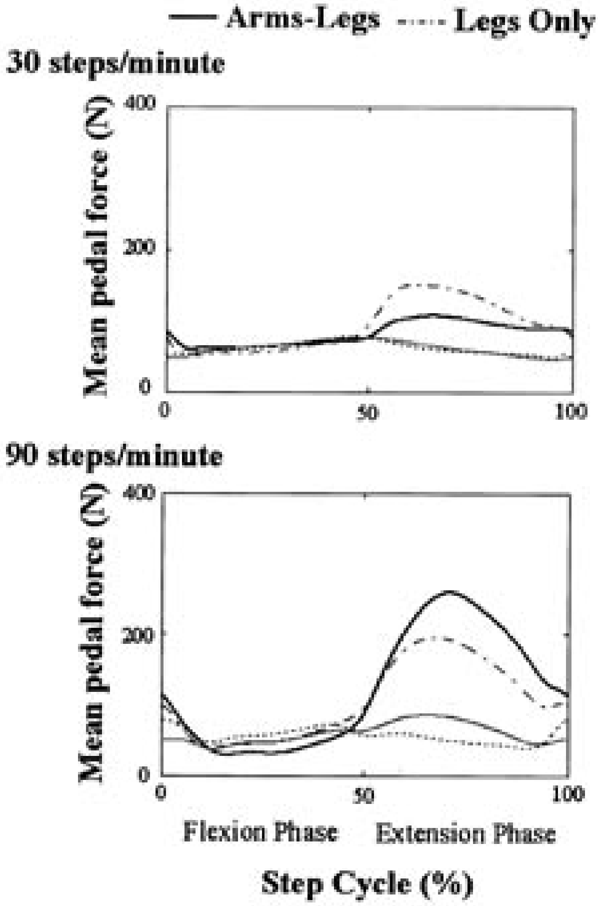

Figure 6 — Averaged pedal force at 30, 60, 90, and 120 steps per minute for all participants. Active stepping with arms and legs (Arms-Legs) and with legs (Legs Only) had similar pedal force patterns. Passive Self-driven stepping had much lower pedal forces but there was a noticeable increase during the extension phase at 90 and 120 steps per minute.

It is not likely that stretch reflexes were primarily responsible for most of the muscle activation during self-driven stepping. Muscle activation for uniarticular muscles occurred during the concentric portion of the step cycle and had similar temporal relationship with the muscle activation of the arms-legs condition. If the muscle activity was evoked by stretch reflexes, it would have occurred during or slightly after the uniarticular muscles were stretched in the eccentric portion of the step cycle.

Several possible neural mechanisms might contribute to the neural coupling between the upper and lower limbs. Among the most likely mechanisms are spinal

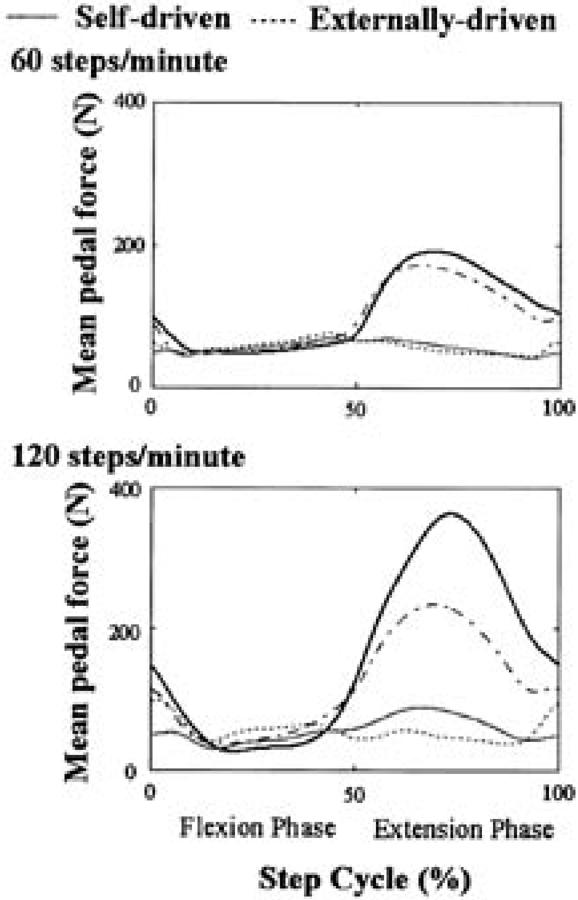

Figure 6 — *(continued)*

connections in locomotor neural networks and supraspinal connections increasing motor neuron excitability. Based on the importance of neural networks in the spinal cord for controlling human locomotion (Dietz, 2003; Dietz, Colombo, Jensen, & Baumgartner, 1995; Ferris, Gordon, Beres-Jones, & Harkema, 2004; Grillner, 1975; Harkema et al., 1997), the most likely candidate is spinal interlimb connections linking upper limb and lower limb motor neurons (Dietz, 2002; Dietz, Fouad, & Bastiaanse, 2001; Frigon, Collins, & Zehr, 2004; Haridas & Zehr, 2003; Wannier, Bastiaanse, Colombo, & Dietz, 2001; Zehr & Duysens, 2004). The increased neuromuscular activation at faster frequencies could be a direct result of upper limb motor neurons transmitting excitatory signals through spinal pathways to lower limb motor neurons (Miller, Reitsma, & van der Meche, 1973; Miller, Van Der Burg, & Van Der Meche, 1975). We are currently conducting a number of other studies using reflex measurements (cf. Frigon et al., 2004), and neurologically injured populations (cf. Ferris et al., 2004), to provide greater insight into the neural mechanism behind this phenomenon. Regardless of mechanism, the results from the present study indicate that fast upper limb movement results in greater lower limb neuromuscular activation.

Maximal stride frequency is a major limiting factor of walking speed in individuals with neurological injury (Pepin, Ladouceur, & Barbeau, 2003; Suzuki et al., 1999). Once they reach their limit in stride length, they are unable to increase stride frequency further as neurologically intact individuals would. This finding led Pepin and colleagues to conclude that rehabilitation after spinal cord injury should emphasize the capacity to produce rapid alternate rhythmical stepping movements with the lower limbs (Pepin et al., 2003). A number of other studies have also demonstrated that stepping at higher speeds produces greater muscle activation, more normal EMG patterns, and better walking capacity after training in neurologically impaired participants (Beres-Jones & Harkema, 2004; Ferris et al., 2004; Hesse, Werner, Paul, Bardeleben, & Chaler, 2001; Patel, Dobkin, Edgerton, & Harkema, 1998; Pohl et al., 2002; Sullivan et al., 2002).

Higher movement frequencies could benefit neurologically impaired individuals via increased proprioceptive feedback. Sensorimotor stimulation is a critical component of activity dependent plasticity during rehabilitation (Muir & Steeves, 1997; Wolpaw & Tennissen, 2001). Faster stepping speeds result in greater afferent feedback to the central nervous system compared to slower stepping speeds because of greater muscle displacements, velocities, and forces (Hof, Geelen, & Van den Berg, 1983; Komi, Fukashiro, & Jarvinen, 1992). As a result, self-assisted exercise at high movement frequencies might accelerate motor recovery by providing greater sensorimotor cues to locomotor neural networks. While physical therapists have traditionally believed that faster speeds might increase unwanted muscle activation or spasticity, there appears to be little evidence for this notion (Brown & Kautz, 1999; McIlroy, Collins, & Brooke, 1992).

A drawback to self-assisted recumbent stepping for gait rehabilitation is that the stepping motion does not faithfully reproduce normal walking mechanics. There are differences in joint kinematic patterns, cutaneous feedback during limb flexion, and postural orientation and stability between recumbent stepping and walking (Huang & Ferris, 2004). Similar arguments, however, could be made regarding cycling and elliptical training to some degree, but there is evidence that both can be beneficial for neurological rehabilitation (Bose et al., 2004; Brinkmann & Hoskins, 1979; Potempa et al., 1995; Werner, Von Frankenberg, Treig, Konrad, & Hesse, 2002). The most significant advantage of self-assisted recumbent stepping/cycling/elliptical training is that they could easily be implemented in the clinic or a patient's home because of low equipment and labor requirements.

Above all, the results of this study indicate that any type of gait rehabilitation device might benefit by taking advantage of neural coupling between the upper and lower limbs. The addition of handles to commercially available mechanized gait trainers (Colombo, Wirz, & Dietz, 2001; Hesse, Uhlenbrock, Werner, & Bardeleben, 2000) could mechanically couple arm and leg movements. This would be a simple way to test if the neural interlimb facilitation found in this study carries over to more task specific locomotor training devices.

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