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RESEARCH ARTICLE

Increases in muscle activity produced by vibration of the thigh muscles during locomotion in chronic human spinal cord injury

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Abstract The purpose of this study was to determine whether the muscle vibration applied to the quadriceps has potential for augmenting muscle activity during gait in spinal cord injured (SCI) individuals. The effects of muscle vibration on muscle activity during robotic-assisted walking were measured in 11 subjects with spinal cord injury (SCI) that could tolerate weight-supported walking, along with five neurologically intact individuals. Electromyographic (EMG) recordings were made from the tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF), vastus lateralis (VL), and medial hamstrings (MH) during gait. Vibration was applied to the anterior mid-thigh using a custom vibrator oscillating at 80 Hz. Five vibratory conditions were tested per session including vibration applied during: (1) swing phase, (2) stance phase, (3) stance-swing transitions, (4) swing-stance transitions, and (5) throughout the entire gait cycle. During all vibration conditions, a significant increase in EMG activity was observed across both SCI and control groups in the RF, VL, and MH of the ipsilateral leg. In the SCI subjects, the VL demonstrated a shift toward more appropriate muscle timing when vibration was applied during stance phase and transition to stance of the gait cycle. These observations suggest that the sensory feedback from quadriceps vibration caused increased muscle excitation that resulted in phase-dependent changes in the timing of muscle activation during gait.

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Introduction

Direct muscle or tendon vibration generally has excitatory effects on homonymous muscles (Prochazka 1996), although the effects in humans are variable. Although tendon vibration activates other afferents (e.g., muscle group II afferents (McGrath and Matthews 1973)) it produces particularly strong activation of muscle spindles, as evidenced by microneurographic recordings (Burke et al. 1976; Roll and Vedel 1982) and the resulting Ia activity causes reflex activation of the homonymous muscle (Eklund and Hagbarth 1966; Hagbarth and Eklund 1966). Vibration also produces prolonged effects, including increased and prolonged motor unit activation following removal of the afferent stimulus, which may be due to intrinsic motoneuron properties that sustain firing (Kiehn and Eken 1997) and suppression of other spinal pathways (Rymer and Hasan 1981). Further, the vibration can change the perception of limb location, specifically causing illusory movements (Goodwin et al. 1972), which can induce antagonist muscle activation (Calvin-Figuiere et al. 1999, 2000).

In individuals with neurological injury, the use of vibratory stimuli has been employed clinically to augment muscle activity (Trombly 1995). In individuals with chronic

spinal cord injury (SCI), the effects of vibration are variable (Hagbarth and Eklund 1966) with a reported increase in an early phasic component, but a decrease in the longer tonic component of the vibratory response (Dimitrijevic et al. 1977). More recent data suggest that the volitional activity during a single-joint task may be increased when paired with vibration of homonymous muscle groups (Ribot-Ciscar et al. 2003), although there is little data demonstrating vibratory effects during performance of a functional, volitional task in human SCI.

In neurologically intact humans, the effects of muscle vibration have been investigated during functional movement tasks, including walking. Recent data suggest that the vibratory stimuli applied to selective lower extremity muscles, particularly the hamstrings, can cause inadvertent forward stepping when subjects attempt to step in place, and increases gait speed during forward walking on a treadmill (Ivanenko et al. 2000). Detailed analysis of electromyographic (EMG) patterns during overground walking with concomitant hamstring or quadriceps vibration significantly increases activity in the rectus femoris and vastus lateralis at phase-appropriate times (Verschueren et al. 2003). In resting subjects in a side-lying position, some evidence indicates that the vibration may also activate locomotor generators. Specifically, this concept is supported by the observation that when the leg is suspended, continuous vibration of selected lower extremity muscles produces rhythmic, locomotor-like movements similar to overground stepping (Gurfinkel et al. 1998) in some neurologically intact subjects. The combined data suggest that the muscle vibration might be useful for increasing phase-appropriate muscle activity during gait in people with impaired locomotor control, such as those with SCI. As a result, the muscle vibration presents promise as a non-invasive technique for enhancing locomotor activity for people with SCI.

In the current study, we tested the effect of vibration of the quadriceps on locomotor muscle activity during treadmill walking with assistance provided by a robotic device. The quadriceps was chosen as a vibration site because (1) quadriceps muscle activity is commonly ill-timed during gait in SCI (Fung and Barbeau 1989; Dobkin et al. 1995), (2) vibration might increase quadriceps muscle activity at heel strike, which might enable greater weight bearing during gait, and (3) quadriceps vibration effects can be compared to prior studies in noninjured subjects during walking (Ivanenko et al. 2000; Verschueren et al. 2003). In neurologically-intact subjects and individuals with chronic SCI, a custom, programmable vibrator placed over the quadriceps provided continuous or phasic vibratory stimuli during assisted treadmill walking at constant speeds and with 40% unloading of body weight. During controlled kinematic-stepping behaviors, the effects of vibration on muscle activity of the ipsilateral leg were quantified through the gait cycle by examining the magnitude and phasing of EMG signals in the tested extremity. We hypothesized that the muscle vibration would increase EMG activity of the homonymous muscles in the ipsilateral leg, with phasic vibratory stimuli potentially increasing the muscle activity during selected portions of the gait cycle. Evidence of enhanced muscle activity during assisted treadmill stepping combined with vibratory stimuli may improve locomotor performance with repeated training.

Materials and methods

Subjects

The clinical features of the 11 individuals with SCI who participated in this study are described in Table 1. SCI subjects were recruited through the inpatient and outpatient physical therapy clinics of the Rehabilitation Institute of Chicago (RIC) (mean age 32.3 years, range 14–62). Primary inclusion criteria consisted of history of SCI above the T10 neurological level, no history of recent (<6 months) lower extremity fracture, and tolerance of upright standing and supported walking for approximately 45 min without symptomatic orthostatic hypotension. Ten of the participants were clinically incomplete (American Spinal Injury Association (ASIA) classification B or C) with one participant clinically complete (ASIA Impairment Scale (AIS) classification A). Average duration of post-SCI was 55 months. Each motor incomplete participant (subjects SCI-1–SCI-9) had a minimum of 8 weeks experience using the Lokomat for treadmill walking. Subjects SCI-10 and SCI-11 had no prior Lokomat experience. Five healthy, neurologically-intact (NI) participants (2 male, mean age 27.8 years) also participated. Informed consent was obtained and all procedures were conducted in accordance with the Helsinki Declaration of 1975 and approved by the Institutional Review Boards of Northwestern University and Marquette University.

Test apparatus

The experiments were performed on a Lokomat driven gait orthosis (Hocoma, Inc, Zurich, Switzerland), which has been described in detail previously (Colombo et al. 2000, 2001). Briefly, the subjects were fitted into the robotic walking device through adjustable straps at the trunk, pelvis, and lower limbs with alignment of hip and knee joints, while an independent pulley system provided weight support through a harness over a motorized treadmill. Each motor incomplete SCI subject (SCI-1–SCI-9) and all control subjects were supported with approximately 40% of their body weight with treadmill speed set at 2.0 kmph.

Table 1 Study participants

SCI [d]	Injury level	ASIA[a] score	LEMS[b]	Age (years)	Sex	Ashworth[c] score	SCATS[d]	Time post injury (months)	Medications
1	C4-5	D	20/17	38	M	0/N/A	0,1,2/N/A	10	
2	C5-6	C	3/21	24	M	0/0	0,1,2/0,1,3	5	
3	C5	D	25/25	36	M	2/2	0,1,1/0,1,1	56	
4	T2-4	D	22/21	26	M	0/1	1,0,0/1,0,0	33	Neurontin (20 mg tid); Lasix (20 mg tid)
5	C2-4	D	25/23	66	M	0/1	1,3,0/2,2,1	164	
6	C8	C	3/0	25	M	0/0	2,1,3/2,13	15	
7	C6	C	11/12	14	F	0/0	2,1,1/2,1,3	20	
8	C4-5	C	12/9	31	M	0/1	0,1,3/1,1,3	60	4-AP (40 mg bid)
9	C5-6	D	25/8	49	M	0/1	0,1,2/0,1,3	179	Cardura (1 mg)
10	C5	A	0/0	17	M	N/A	N/A	5	
11	T4	B	0/0	30	F	N/A	N/A	61	

[a] ASIA score is the American Spinal Injury Association impairment scale

[b] LEMS is the lower extremity motor score of the ASIA neurological classification test (Maximum score is 50)

[c] Ashworth Score (Ashworth 1964). Ashworth Scores are listed for left/right knee extensors

[d] SCATS is a measure of spastic reflex excitability (Benz et al. 2005). SCATS is listed for left/right extensor, flexor, and clonus reflexes

Subjects were not allowed to use their arms for weight support. The 40% body weight support condition necessitated muscle activation during gait, and all NI and incomplete SCI subjects were instructed to walk with the Lokomat while stepping. Motor complete SCI subjects (SCI-10–SCI-11) were provided 100% body-weight support (i.e., airstepping). These conditions remained consistent throughout the duration of the experiments.

A custom vibrator was constructed with a Faulhaber Brushless DC Servomotor (GMBH & CO. KG, Schönaich, Germany) with dual shafts and linear hall-effect sensors controlled with the Faulhaber MCBL 2805 Motion Controller using the EEPROM and analog controller. The system is a PI controller that was run in velocity mode with a step input in command voltage. Biaxial eccentrics (6 g with a center of mass 15 mm from the shaft center) were attached in-phase to generate, approximately, 10 N peak force at a rotational speed of 80 Hz. Motor and eccentrics were encased in a 32-mm diameter, 89 mm length polyvinyl chloride sleeve. Vibrator force amplitude (10 N) and timing were checked by affixing the device to a load cell (JR3, Woodland, CA), and testing 0.85 s on–off cycling of the vibration. Vibration achieved the targeted force (10 N) within approximately 25 ms and varied by less than 0.25 N during the 0.85 s of vibration. During walking, the vibrator was placed over the anterior thigh, approximately over the rectus femoris (RF) muscle belly and secured with moderate pressure using a strap placed around the thigh.

The control signal to the vibrator was adjusted using a phasing circuit to time the vibration continuously or during one of four different phases of gait. The hip and knee angles were recorded from the joint-angle sensors of the Lokomat system, with cadence adjusted to ~28 strides/min. Note that small differences in measured and actual joint angles could occur due to motion of the subject within the Lokomat and minor misalignment of the joints. The phasing circuit monitored the hip angle and a custom-timing circuit detected when the hip rotated past 10º flexion. The timing circuitry produced a command signal to activate the vibrator during swing, transition to swing, stance, or transition to stance. Swing phase began 1.8 s after the trigger, and stance phase began 0.8 s after the trigger (see Fig. 1). The transition to swing and stance began 1.4 and 0.4 s after the trigger, respectively. All of the phased vibration signals lasted 40% of the gait cycle (0.85 s).

Test paradigm

The experiment consisted of six different vibration scenarios, all tested in a single test session. Subjects stepped continuously during the test session, with vibration turned on and off in different scenarios. Each scenario was tested for 20 s (~8 strides), with >1 min between tests. The first scenario was a no vibration test ("NoVibe"), which was recorded before each of the other vibration scenarios. This NoVibe measurement helped to reduce any bias produced by a preceding test scenario by serving as a baseline measurement for each test. There were four different phased vibration scenarios including: (1) swing phase, ("Swing"), (2) transition to swing phase, ("Tran2Swing"), (3) stance phase, ("Stance"), and (4) transition to stance phase ("Tran2Stance). The phased vibration scenarios were

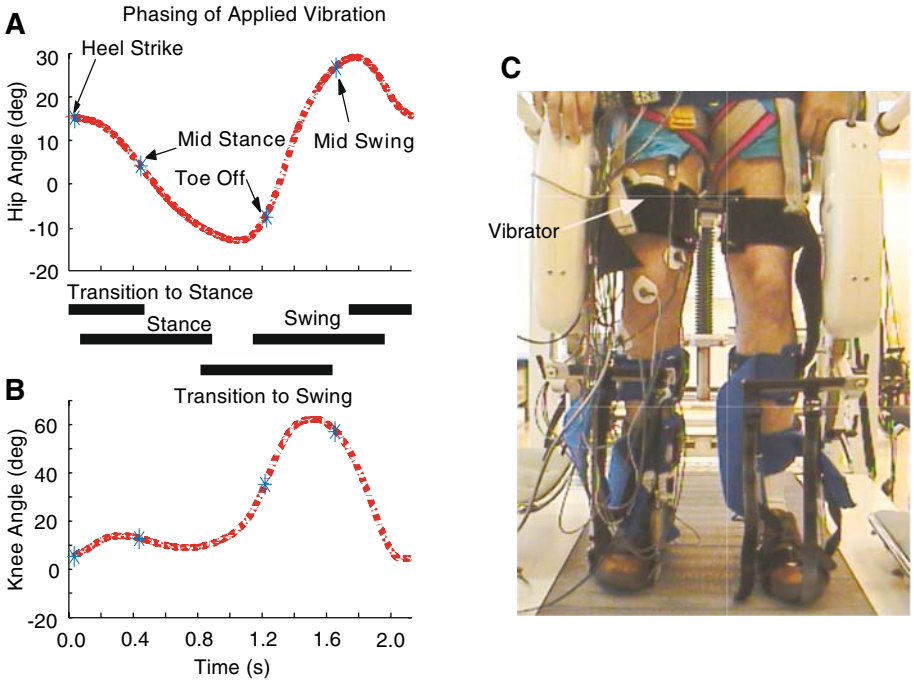

Fig. 1 **a** Depiction of hip and knee sagittal plane kinematics (as determined from the sensors of the robotic orthosis) and the timing of the different vibration conditions throughout the step cycle. **b** Positioning of subject within the robotic orthosis with placement of the vibrator under the thigh cuff

randomized four times to produce four different sets of tests in order to minimize history effects. At the end of each set, a sixth scenario was recorded with continuous vibration ("VibeAll"). NoVibe measurements (made before each vibration trial) were also used to assure that the amplitude and pattern of EMG did not change during the test session.

Surface electromyograms (EMGs) were recorded during each test. Surface electrodes (Suretrace 1800, ConMed Corp., Utica, NY) were placed on lightly abraded skin over the muscle belly of the tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF), vastus lateralis (VL), and medial hamstrings (MH, i.e., semitendinosus and semimembranosis) of each leg (see Fig. 2). Signals were amplified (1,000×), filtered (10–500 Hz) (Myosystem 1400A, Noraxon USA, Inc, Scottsdale, AZ), and collected at a sampling rate of 1,000 Hz using a 16-bit, 64-analog-input multifunction data acquisition card (PCI-6031E, National Instruments) on a PC with custom Matlab (The Math Works Inc., Natick, MA) software.

Analysis

EMG signals were compared between conditions to identify the effects of vibration on the locomotor muscle activity pattern. All EMG signals were high-pass filtered at 15 Hz to remove movement artifacts, and then band-stop filtered from 62–88 Hz to remove noise from the 80 Hz vibration. A fourth order Butterworth filter applied in the forward and backward directions was used for all filtering (using the *filtfilt* function in Matlab). This filtering algorithm was based on a preliminary spectral analysis of the EMG signals. Many of the EMG signals, regardless of vibration conditions, contained power in frequencies <15 Hz that appeared to correspond to relatively slow signal fluctuations in the time-domain plots at times where motion artifacts were expected. Vibration commonly added signal power selectively between 65 and 85 Hz to the RF and VL signals. As a result, these frequencies were removed from all EMG signals. The filtered signals were then plotted against time, and then rectified and smoothed with a low-pass filter at 20 Hz.

To evaluate the muscle-activity patterns during the gait cycle, each 20 s recording was divided into seven gait cycles. Cycles were identified by detection of the peak hip-angle signal recorded from the Lokomat during each cycle. A time delay of 17% of the subject's average gait cycle was added to the time of peak hip angle in order to align each gait cycle with the heel strikes (HS). This time delay was based on a preliminary measurement in which the timing of heel contact was measured relative to the hip angle. Because each vibration scenario was run four times during a single test session, each scenario was tested for a total of 28 gait cycles. The EMGs from these trials were then ensemble averaged across the gait cycles. Muscle-activation patterns for each scenario were characterized by the ensemble averages of the EMG signals for each muscle. The area beneath the rectified, smoothed EMG was used to compare the effects of vibration on the magnitude of the muscle activity across vibration sequences and test groups. The activation magnitude of each leg muscle (TA, MG, RF, VL, MH) was determined by calculating the average area under the EMG curve (using the *trapz* function in Matlab) during each gait cycle for each vibration scenario.

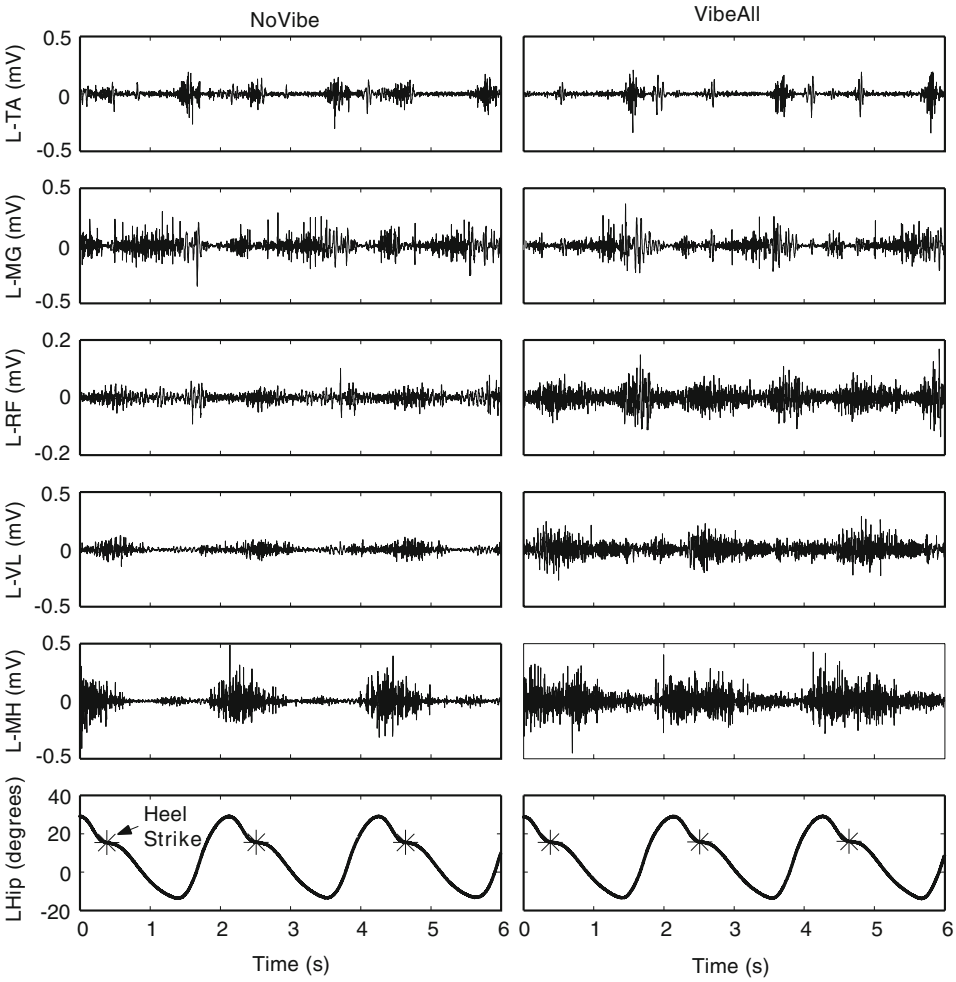

Fig. 2 Example of the selected lower extremity muscles and hip kinematics prior to and during continuous vibration (VibeAll) application for subject SCI-1. Little alteration was observed in TA and MG muscles. Enhanced and prolonged EMG activity was observed in both quadriceps muscles (RF and VL shown) and hamstrings (MH). The increase in EMG signal occurred throughout the step cycle during the VibeAll condition, although the increase appeared to be more pronounced during the pre-vibration timing of muscle activation

The NoVibe condition preceding each test condition was used for comparison. We observed that the NoVibe results showed no trend across the test session, and no correlation with prior vibration conditions, suggesting that there were no lasting motor-learning effects associated with any individual vibration test. Each EMG measurement was treated as a separate dependent variable and the NoVibe conditions were compared between the SCI and NI subjects using a *t* test to determine whether the baseline signals differed significantly. Then, each EMG dataset was entered into a mixed model ANOVA to determine whether the vibration had a different effect on NI and SCI subjects. Subject was entered as a random factor and the interaction of subject type (NI vs. SCI) × vibration (On/Off) and interaction of vibration (On/Off) × cycle (VibeAll/Stance/Trans2Swing/Swing/Trans2Stance) were tested. The preceding NoVibe measurement was used for quantifying the Off activity in each scenario. To display the results across the test groups, the average EMG areas were normalized by dividing each participant's average EMG area during a vibration scenario by its NoVibe counterpart for each of the muscles. SCI data were then analyzed separately using a repeated measures ANOVA with the interaction of gait cycle (VibeAll/Stance/Trans2Swing/Swing/Trans2Stance) × vibration (On/Off) to determine whether the timing of the vibration significantly affected the response in SCI subjects.

In order to determine if any of the vibration scenarios could alter the muscle phasing in the leg muscles, muscle activity during gait was quantified using a circular analysis (Batschelet 1981). Each rectified, smoothed EMG signal was plotted in polar coordinates for each gait cycle, with heel strike defined at 0°, stance ending with toe off (TO) at 206°, and swing following through to 360°. The sum of the *x*- and *y*-values from the polar plot were then used to calculate the *x*- and *y*-coordinates of a vector that represented the phasing of the EMG in a single gait cycle. The phasing of the EMG across cycles was then calculated. First, the mean vector for each gait cycle was set to a standard length (*r*) of one, effectively normalizing the EMG magnitude for each cycle. The mean *x*- and *y*-values from the end points of the normalized vectors for all cycles of each test condition were then used to calculate a mean vector. This vector was characterized in polar coordinates by an angle (θ) and a length (*r*), which were used in the subsequent circular

statistical analyses. A total of ten mean vectors were calculated for each muscle to represent each of the five vibration scenarios as well as the five preceding NoVibe scenarios. The Watson–Williams test was run for each individual to determine if the mean angle (θ) from each of the vibration scenarios was significantly different from the preceding NoVibe mean angles (Batschelet 1981). In order to test whether the EMG activity from each of the vibration scenarios demonstrated significant directedness, the Rayleigh test was conducted on the mean vector length (r) for each muscle, for each subject (Batschelet 1981). Note that a mean vector length near one would occur when phasing was consistent across cycles while the length would be closer to zero if the phasing was more variable.

After calculating the mean vector for each muscle, for each subject, the groups were tested for effects of vibration on the phasing of the muscle activity. The mean vector length for each subject from each scenario was normalized to one. The x- and y-values from the end point of each standardized mean vector for each subject were then used to establish a mean vector for the group (SCI or control) in each scenario. The Watson–Williams and Rayleigh tests were conducted with the group data sets to determine statistically significant ($\alpha = 0.05$) trends in muscle phasing across the group. The Hotelling one-sample test was used to determine significant changes in mean vector angles, between the vibration scenarios and NoVibe values, across the group using a paired comparison.

Results

EMG area results

Individual data trials suggested that the vibration applied over the quadriceps increased the thigh muscle activity but had minimal effect on the ankle muscles. EMGs of the TA, MG, RF, VL, and MH were compared for the VibeAll scenario to the NoVibe condition for each test subject across several gait cycles in order to identify trends in the data. Representative data from subject SCI-1 are shown in Fig. 2, demonstrating the difference between VibeAll and NoVibe conditions. The TA and MG showed no consistent differences between the VibeAll and NoVibe response, while the RF, VL, and MH appeared to have greater responses when vibration was applied. The MH also appeared to remain active for a longer duration.

Similar trends were observed in the rectified, smoothed EMG signals. Figure 3 demonstrates the average EMG signal of SCI-1 during VibeAll (thin black signal) and the preceding NoVibe intervals (thick gray signal). Again, an increase in EMG amplitude was seen for the RF, VL, and MH. The bars at the bottom of each EMG graph in Fig. 3 are included as a reference to signify "appropriate" muscle timing during the gait cycle, based on Lokomat stepping in healthy subjects (Hidler and Wall 2005). Note that the anticipated EMG patterns are affected by spinal cord injury (Pepin et al. 2003) and body-weight support (Ferris et al. 2001; Ivanenko et al. 2002).

The application of phased vibration also appeared to increase the muscle activity, although primarily during the vibratory stimuli. For example, the EMG signals during the different vibration phases are shown in Fig. 4 for Subject SCI-3. In Fig. 4, the shaded region delineates the different timed vibration scenarios and the bar at the base of each graph is used to represent the appropriate muscle timing during gait. One noticeable effect, common across the SCI group, was a phasic increase in EMG for the RF and VL at the onset and termination of the vibration. The responses were variable and the effects were not observed in all subjects, for all muscles, and sometimes depended on the phase of gait in which the vibration was applied. Anecdotal observations suggested that the onset and termination bursts were not present in the NI, however, the bursts were not quantified for statistical comparison. A spectral analysis of the EMG signals with transient increases and decreases indicated no difference in low-frequency power, as would occur with motion artifact.

The group data demonstrated higher EMG for the ipsilateral thigh muscles with vibration, compared to NoVibe, while no differences were observed in the ankle muscles. These results are summarized in Fig. 5a, in which the average area of the smoothed, rectified EMG signal for each specific vibration scenario was normalized and then averaged across the group. RF, VL, and MH activities were higher during all vibration scenarios, with VibeAll showing the largest increase in EMG activity (43% increase in RF, 42% in VL, and 28% in MH). The timed vibration scenarios ranged from 14 to 31% increase in average EMG activity (Table 2). Note that Subject SCI-2 was removed from the analysis because 80% of the participant's EMG values were statistically defined as outliers (i.e., mean EMG area >3 SD above group mean). Subject SCI-2 showed a dramatic increase in EMG signals during the vibration across all muscles (including distal muscles) and the response subsided within two steps after termination of vibration. The reason for the high sensitivity to vibration in this subject could not be identified.

The muscles that increased activity with vibration were common for the NI and SCI groups, with means of the group EMG areas shown in Fig. 5. For NI subjects, there were no consistent differences in the TA and MG EMG areas with and without vibration, but the RF, VL, and MH were larger for all vibration scenarios (96% increase in RF, 211% in VL, and 41% in MH). The timed vibration had slightly lower effects, with increases in the mean EMG area

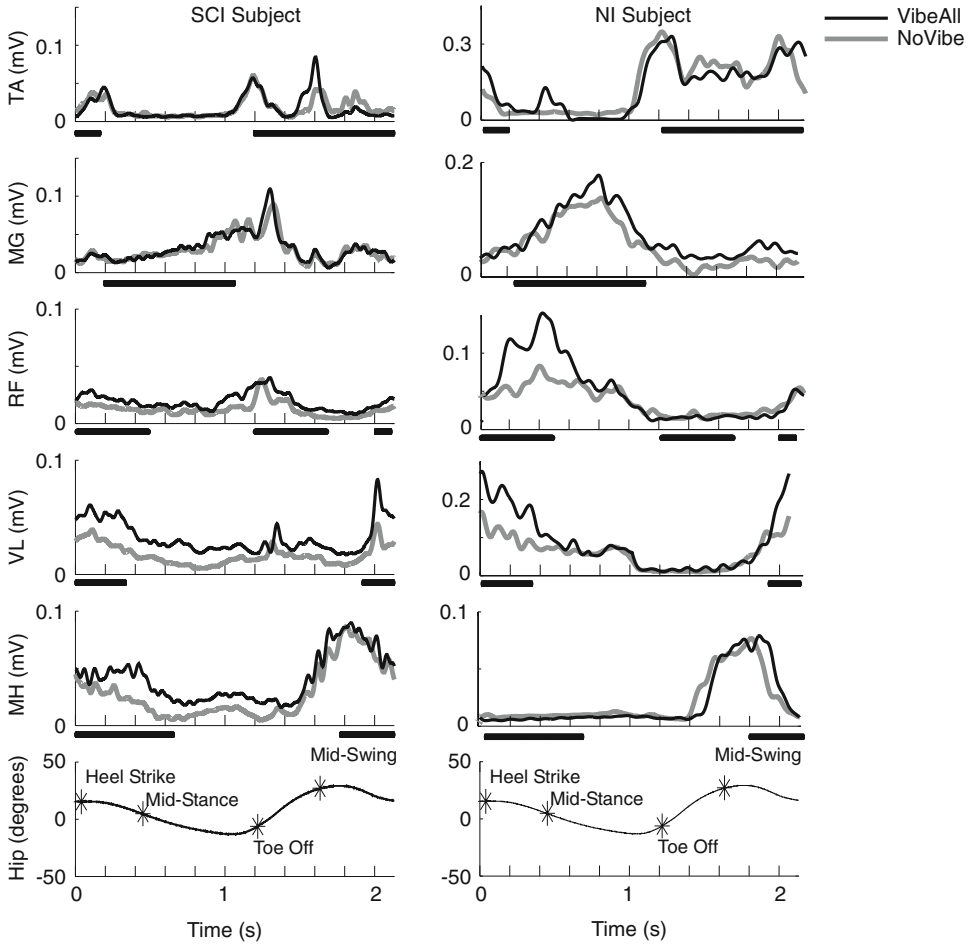

Fig. 3 Rectified, smoothed EMG activity during NoVibe and VibeAll conditions for subject SCI-1. *Solid line* indicates time period during which the EMG activity is typically observed in intact subjects during robotic-assisted walking (estimated from NI data and data of Hidler and Wall (2005)). Note that the EMG for the MG activity peaked in early swing, consistent with typical patterns of treadmill walking in incomplete SCI (Israel et al. 2006). VibeAll appeared to increase the EMG of the thigh muscles. Increases in muscle activity were typically largest during the "normal" muscle activation times, although the VL was higher throughout the gait cycle in this subject

(across Swing, Stance, Tran2Swing, and Tran2Stance) for the RF of 38, 55, 44, 47% and an increase of 123, 86, 94, 100% for the VL, respectively, with MH demonstrating a smaller increase. In the data for both groups (NI and SCI), the vibration produced a statistically significant increase in muscle activity for the RF, VL, and MH (ANOVA, $P < 0.001$). The NI subjects had greater effects of vibration than SCI subjects for the MH (ANOVA, vibration × group interaction $P = 0.012$) and VL (ANOVA, vibration × group interaction $P < 0.001$). The RF was not significantly different between the SCI and NI groups ($P = 0.058$). The baseline activity of the NI and SCI groups was not significantly different ($P > 0.05$). The vibration appeared to have little effect on the other muscles ($P > 0.05$).

Although the magnitude of the change in muscle activity with vibration differed between the SCI and NI groups, the vibration still produced a statistically significant increase in EMG activity for the thigh muscles of the ipslilateral leg in the SCI group. A one-way repeated measures ANOVA indicated a significant difference between the EMG activities for the vibration condition (compared to preceding NoVibe) for the RF, VL and MH ($P < 0.001$). No significant effect of vibration was observed for the ipsilateral TA and MG or for muscles of the contralateral leg (ANOVA, $P > 0.05$). There was no significant difference in the EMG area for the different vibration conditions for any of the muscles (ANOVA, $P > 0.05$). Thus, vibration appeared to increase muscle activity of the ipsilateral thigh muscles. A phase analysis was conducted to determine whether the timing of muscle activity was altered by the vibration.

EMG phase results

Muscle phasing was modified by vibration in individual subjects. In Fig. 6, the resultant vectors are shown for RF, VL, and MH in SCI-3 for the following vibration scenarios: Stance, Tran2Stance, VibeAll and each preceding NoVibe scenario. Zero degrees corresponds to the heel strike (HS) and toe off (TO) is indicated at 206°. The gray arc plotted inside the circular graph is used to demonstrate expected muscle timing during the gait cycle based on Lokomat stepping in noninjured individuals (Hidler and Wall 2005). The NoVibe vectors showed a consistent phase angle of preferred activation. Although the MH did not change its phasing appreciably in this subject, the RF and VL showed

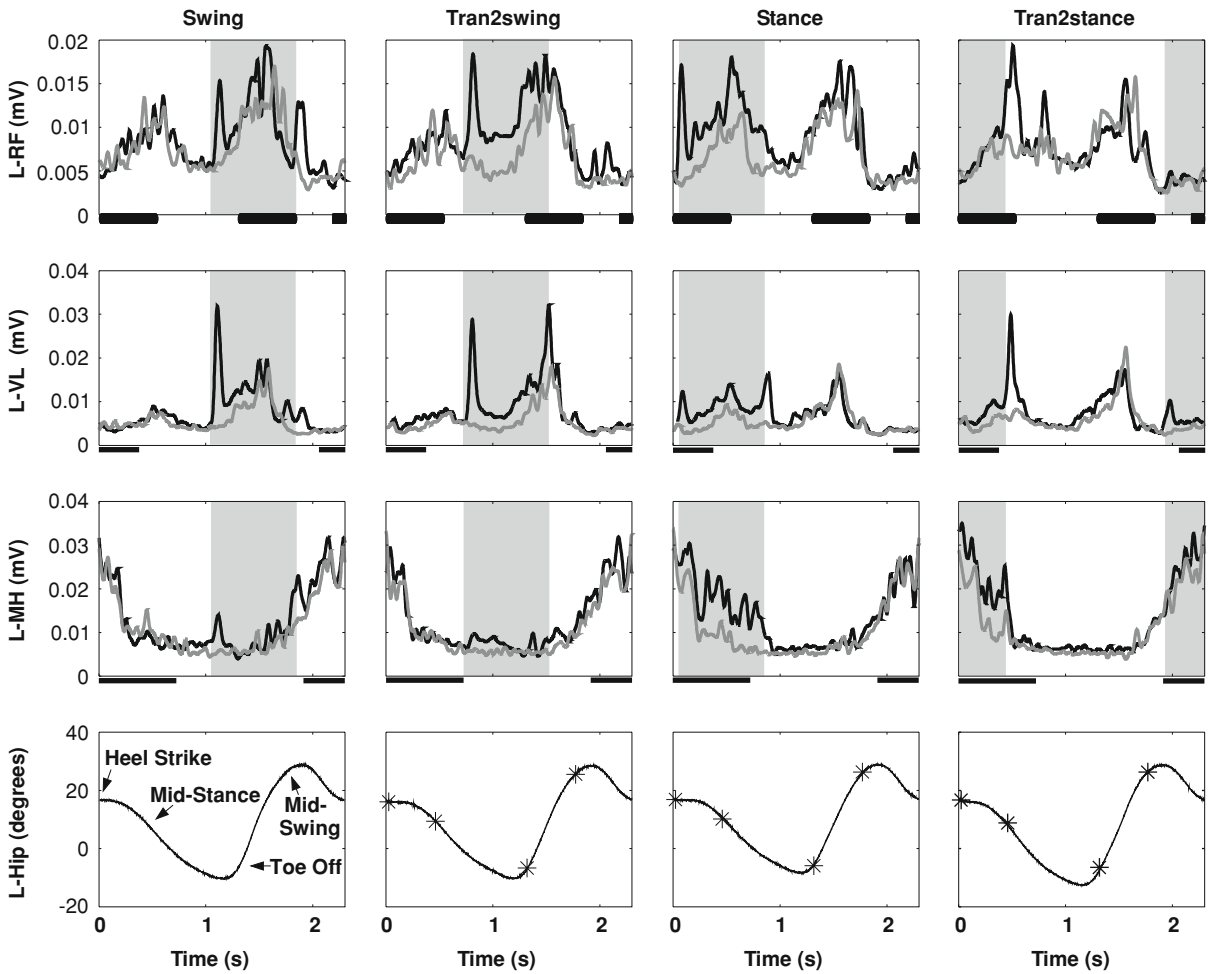

Fig. 4 Rectified smoothed activity of upper leg muscles during vibration at different phases of the gait cycle (shaded in gray) for subject SCI-3. *Gray traces* indicate pre-vibration, *dark lines* indicate vibration conditions. *Solid dark lines* below each trace indicate the time period in which muscle are typically active during robot-assisted treadmill stepping (Hidler and Wall 2005). EMG signals were increased during the time of muscle vibration, with little carryover into the nonvibrated time. The onset and offset of vibration sometimes produced a burst of muscle activity, as shown for this subject

a shift away from toe off and more towards mid-stance during the Stance and Tran2Stance scenarios.

The group data also demonstrated a net shift in the phasing of the muscle activity with vibration. The VL resultant vectors are shown in Fig. 7 for each of the four vibration scenarios in both the SCI and NI groups. The gray arc plotted inside the circular graph is used again to demonstrate appropriate muscle timing and the black arc is used to display the vibration timing. The NoVibe resultant vector, from the NI group demonstrated an appropriate phasing near heel strike for the VL in all four graphs and the different vibration scenarios tended to pull the resulting vector toward the timing of the vibration. The NoVibe resultant vector from the SCI group was much shorter because the SCI group tended to have a less-consistent muscle phasing than the NI group. The NoVibe vectors demonstrate an inappropriate phasing of the VL, occurring near toe off in all four graphs, consistent with the phenomenon of stiff-knee gait observed in subjects with neurological injury (Fung and Barbeau 1989; Kerrigan et al. 1991). The Stance and Tran2Stance vibration scenarios appeared to draw the resulting vector toward the timing of the vibration, which was closer to the normal phasing of the activity. This effect is consistent with the increase in EMG observed during vibration, including both the transient and overall shift in EMG (e.g. Fig. 4).

Circular statistics applied to the muscle phasing in each individual subject indicated that the vibration produced significant alterations in the overall phasing of muscle activity. The Rayleigh test was used to determine whether each muscle demonstrated a preferred phase direction. In muscles with phasic activity (Rayleigh test, $P < 0.05$), the Watson-Williams test was used to determine whether the mean angle from each of the vibration scenarios was significantly

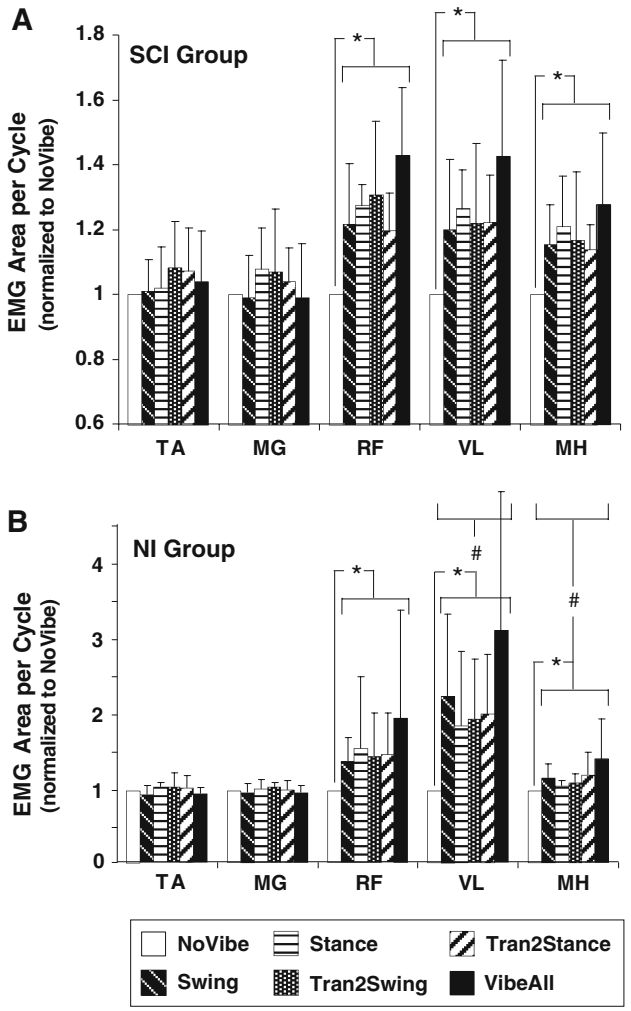

Fig. 5 Alterations in EMG area during no vibration (NoVibe) versus different vibration conditions in SCI subjects (**a**) and uninjured subjects (**b**). The EMG area for each condition was normalized by the EMG area in the NoVibe condition. Statistically significant increases in muscle activity with vibration are indicated by $*P < 0.05$ and significant differences in the effects of vibration on muscle activity between the SCI and NI groups are indicated by $\#P < 0.05$

Table 2 Percent increases in EMG during vibration for SCI subjects

Scenario	RF (%)	VL (%)	MH (%)
Swing	22	20	15
Stance	27	26	21
Tran2Swing	31	22	16
Tran2Stance	19	22	14

different from the preceding NoVibe mean angle (Batschelet 1981). Eight of 11 SCI subjects had significant changes in RF phasing, with 6/11 in the VL, and 7/11 in the MH during Stance vibration (Watson–Williams test on each individual, $P < 0.05$). During Tran2Stance and VibeAll vibration, muscle timing of the quadriceps (RF and VL) was altered significantly in over half of the SCI subjects (6/11). Changes in muscle phasing occurred in the RF and VL during Swing and Tran2Stance vibration, in 3/5 NI subjects. These results are summarized in Table 3.

Similar results were observed across the groups. For the group analysis, the Hotelling one-sample test was used to make paired comparisons between the mean EMG phase angle of each vibration scenario and the EMG phase angle in the NoVibe condition. The NI group showed no significant change with vibration, but the SCI group displayed significance in at least one of the two quadriceps muscles measured (RF or VL) for every vibration scenario (Hotelling test, $P < 0.05$). Stance vibration produced the greatest change, with significant changes in the RF and VL (Hotelling test, $P < 0.01$), as well as the MH (Hotelling test, $P < 0.025$).

Discussion

The aim of this study was to evaluate the effects of unilateral quadriceps vibration on SCI individuals during robotic-assisted treadmill stepping. The main results were: (1) a significant increase in EMG activity of the RF, VL, and MH in both NI and SCI subjects when vibration was applied over the RF; and (2) a phase shift in muscle activity for the RF and VL towards mid-stance during the Stance and Tran2Stance vibration scenarios in the SCI group. The combined results suggest that the muscle vibration might be useful for improving phasing of muscle activity during pathologic gait. In particular, the muscle vibration could prove effective in enhancing quadriceps activity during stance, allowing patients to bear more weight during treadmill training, thereby further increasing locomotor muscle activity and the subsequent training effect (Visintin and Barbeau 1989; Harkema et al. 1997).

The magnitude of the increase in EMG activity associated with vibration in the current study was comparable to previous reports in NI subjects. Verschueren et al. (2003) found an increase in EMG of 69% in RF EMG and 65% in VL EMG during continuous vibration of the quadriceps (at the knee) in NI subjects during normal walking. In the current study, increases observed in NI subjects ranged approximately 40–200% during the VibeAll conditions, and 40–120% during phased vibration. In the SCI population, increases in EMG were lower (43%); however, even the subjects with motor complete SCI (SCI-10, SCI-11) demonstrated increased muscle activity, indicating a potential spinal mechanism underlying the observed responses. As a percentage of the pre-vibration muscle activity, the effects of vibration in motor complete subjects were within the range of the other subjects. Overall, the baseline activity in the two motor complete SCI subjects was 36% of the

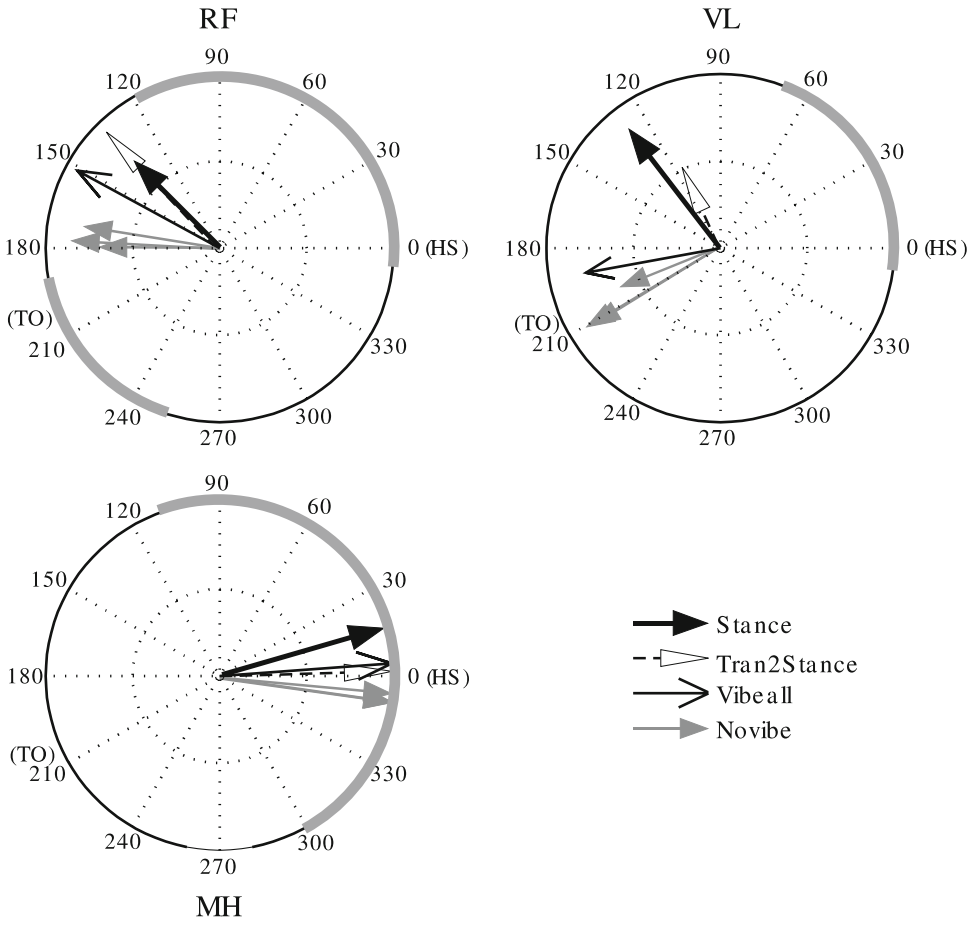

Fig. 6 Single subject example of muscle phasing changes during different vibration periods for three thigh muscles (VL, RF, and MH). *TO* marks toe off, *HS* marks heel strike. Since the result vector is the normalized vector sum of all step cycles, the size of the vector reflects the step-to-step consistency of the timing. For example, when the phasing is identical from step-to-step, the vector length will be one. When the phasing from step-to-step is random, the vector length approaches zero. Vibration throughout the gait cycle (VibeAll) produced only modest changes in timing of muscle activation. Conversely, the vibration during stance (Stance) and transition to stance (Tran2Stance) generally produced more "corrective" changes in muscle activity phasing. The circle diameter designates a magnitude of one, with vector length indicating the repeatability of phasing

level in the incomplete subjects, however, the phasic muscle activity was still evident in the EMG signals and increases were noticeable with vibration.

Neural mechanisms of vibration-initiated muscle activity during walking

The increased muscle activation could be explained largely by reflex activation of the vibrated (i.e., homonymous) muscle. Tendon and muscle vibration strongly activate muscle Ia afferents, as evidenced in spinalized cats (Bianconi and van der Meulen 1963; Matthews 1966) and humans (Burke et al. 1976; Roll and Vedel 1982; Roll et al. 1989). In the current study, increases in muscle responses during walking were largely constrained to the vibrated muscles of the thigh during the vibration, consistent with activation of homonymous reflex pathways. Further, the muscle activation associated with the vibration appeared to have a strong "phasic" component consisting of a burst activity at onset of vibration (Fig. 4), consistent with the observations of an enhanced phasic component of tendon vibration in people with SCI at rest (Dimitrijevic et al. 1977). Overall, these observations suggest that the local increase in muscle activity associated with vibration during walking could occur through stretch-reflex pathways.

Vibration also produced an increase in MH activation, although to a slightly lesser extent than the quadriceps muscles. The MH activation may have been caused by several factors. The strap used to secure the muscle vibrator to the leg was relatively stiff and was not designed specifically to isolate the vibration to the quadriceps. As a result, there was likely to be substantial spillover vibration to the MH, which could have resulted in a stretch reflex excitation of the hamstrings. Alternately, the MH could have been activated through reciprocal-reflex pathways, which may become facilitatory in spinal spastic paresis (Crone et al. 2003; Xia and Rymer 2005), however, the MH activation during vibration was similar for the SCI subjects and the NI controls (compare MH in Fig. 5a and b). Antagonist activity has also been noted during vibration as a component of the response to illusory sensations of movement (Feldman and Latash 1982; Calvin-Figuiere et al. 1999, 2000). Although beyond the scope of the current study, investigation of vibratory stimuli applied directly to the hamstrings may be warranted, especially given the problems with knee extension during swing in spastic gait (Kerrigan et al. 1991) and the potentially stronger effects of hamstrings vibration on locomotion (Ivanenko et al. 2000).

Another potential mechanism underlying the observed increase in muscle activity during vibration may be the

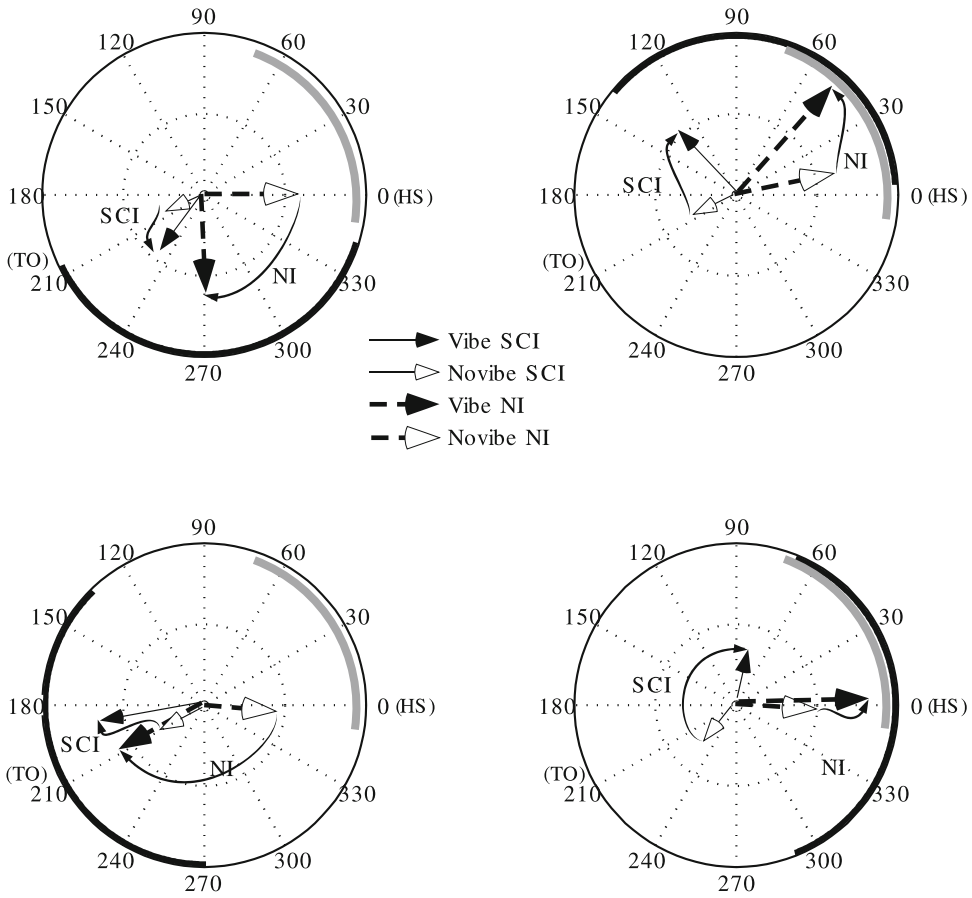

Fig. 7 Grouped VL EMG data all SCI subjects and NI subjects during walking with and without thigh vibration, applied during four different phases of the step cycle, indicated by the *black arc*. Normal timing of the VL is indicated by the *gray arc*. *TO* marks toe off, *HS* marks heel strike. In addition to a change in direction, the size of the resultant vector increased during vibration, indicating a more-consistent timing of VL EMG across subjects during vibration

Table 3 Number of SCI subjects (out of 11) showing significance of phase changes with vibration (Raleigh and Watson–Williams tests)

	TA	MG	RF	VL	MH
Swing	3	2	4	6	5
Stance	3	4	8	6	7
Tran2Swing	5	5	5	5	4
Tran2Stance	3	2	6	6	5
VibeAll	7	2	9	6	5

onset of spastic reflex responses. Spasticity, defined as a velocity-dependent increase in the reflex response to passive muscle stretch, and the presence of multijoint flexor or extensor spasms are often observed in responses to specific afferent cues applied to the lower limb. While abnormal reflex responses certainly account for altered muscle activity patterns in SCI, their contribution to the observed effects of vibration presented here may be limited. In particular, the lower leg muscle activity is often observed during multijoint flexor and extensor spasms, while there was no apparent change in ankle muscle EMG during vibratory stimuli in the current study. Secondly, the augmented EMG of the RF and VL was observed during the vibratory stimuli, which occurred during phases of the gait cycle in which both quadriceps lengthening and shortening were occurring, with differences in EMG changes between phases. With the exception of SCI-2, increases in muscle activity were constrained to the time of applied vibration, without evidence of prolonged "spastic" activity. Consequently, the enhanced muscle activity was likely due directly to the vibratory stimuli affecting primarily Ia pathways, and not the result of spastic reflexes.

The vibratory stimuli used in the current study did not appear to affect ongoing activity of a presumed spinal locomotor generator. Central pattern generator (CPG) control of locomotion, including modulation by sensory afferents, has been studied extensively in animals, with some suggestion that the spinal locomotor generators also exist in humans (Bussel et al. 1988; Calancie et al. 1994; Dimitrijevic et al. 1998). In the current study, the increased activation was limited to the vibrated muscles or those muscles near the vibratory stimulus, whereas distal muscles below the knee were not affected. One subject, Subject SCI-2, responded very strongly to the vibration, including activation of the ankle muscles in a pattern that strongly enhanced locomotion. The data from this subject were removed from the group analysis because the increase in activation was so large (subject data were >3 SD above the mean) that the

data point was classified as an outlier. It is possible that the vibratory stimulus activated larger pools of spinal interneurons, including a presumed CPG in this subject, although this is difficult to assess in humans. This effect was only observed in a single subject in this study, however, the sample size of the current study was only eleven, and the effect might be more prevalent than observed in our sample. In addition, the vibratory stimuli with greater forces, at other sites and for longer durations might increase the likelihood of a stronger facilitatory effect on locomotion. Gurfinkel et al. (1998) observed cyclical movement of the leg in half of subjects tested when vibration was applied to multiple sites on the leg under gravity eliminated, relaxed conditions. The sensory cues and motor programs initiated during stepping could reduce the effect of vibration on activation of central pathways related to gait, however, the widespread increases in muscle activity observed in subject SCI-2 warrant further investigation using different vibratory inputs.

Functional implications

The increase in selective muscle activity produced by vibration may be useful in augmenting locomotor training performed on a treadmill using body weight support (BWS). While such training performed on a treadmill typically provides load support in order to enable walking, animal studies in which the initiation of swing phase can be blocked by maintaining load on the ankle extensors (Duysens and Pearson 1980; Timoszyk et al. 2005) stress the importance of minimizing load support. Consequently, the locomotor training with BWS in SCI patients provides only partial support and maintains some unsupported weight in order to activate load receptors that are critical for leg-muscle activation during stepping movements (Dietz et al. 1995, 2002; Harkema et al. 1997). Vibration might prove useful in further reducing the amount of BWS needed for gait, however, the absence of facilitation of ankle planar flexor activity by vibration might limit any reduction in BWS. In addition, the co-activation produced by increasing antagonist activity may increase stiffness to the point of reducing desired knee mobility, which could have a detrimental effect on swing. As a result, the vibration might be best applied during the late swing/early stance phase of locomotion, thereby increasing quadriceps activity and assisting stance phase control. The resulting increase in knee stiffness would be expected to reduce the risk of knee buckling, and consequently allow greater weight to be supported during stance. If effective, this would allow treatment using less BWS, thereby potentially increasing the therapeutic effect of treadmill training.

Vibration may also be useful in shifting the timing of muscle activity during locomotion. Note that the vibration of the quadriceps selectively enhances the EMG bursts of the RF and VL during the stance phase of walking more than the swing phase in NI subjects (Verschueren et al. 2003). The increased sensitivity during stance may be related to a phasic increase in quadriceps stretch-reflex excitability during stance (Dietz et al. 1990a, b), which would be activated by Ia drive from the tendon vibration. In the current study, there was no apparent reduction of muscle activity in the off phase during the different vibration scenarios, and the targeted stance vibration produced an increase in muscle activity that was comparable to the increase when vibration was applied during other phases of gait. As a result, the change in phasing observed in the current study was more likely attributed to the increase in muscle activity associated with the vibration itself, rather than a fundamental alteration in locomotor timing. Regardless, muscle vibration might be effective in altering the net timing of thigh-muscle activity during gait.

The implications of the vibration on overground walking and on treadmill walking without robotic assistance needs to be examined in people with SCI. In NI individuals, the vibration of the quadriceps has little or no effect on forward locomotion (Ivanenko et al. 2000) based on measures of stepping frequency and stride length. Conversely, Verschueren et al. (2003) observed an earlier onset of TA EMG activity after vibration of the RF during unassisted walking, although the magnitude of the EMG did not change appreciably. This EMG change produced changes in the remote ankle joint during vibration of the quadriceps, including a decreased plantar flexion at toe-off and an increased dorsiflexion during swing. In the current study, the kinematics of the gait were largely controlled by the Lokomat, with vibratory effects observed by increases in EMG in the RF, VL, and MH. The lack of changes in ankle muscle activity in this study could be accounted by the constraint of the metatarsal straps on the Lokomat that provide toe clearance. We expect that the EMG changes during vibration would produce corresponding changes in gait kinematics in people with sufficient walking ability.

Conclusion

This study demonstrated that the quadriceps vibration triggers an increase of EMG signals from the RF, VL, and MH, while not affecting the TA and MG in people with SCI. In addition, the vibration could be selectively timed to shift the phasing of the quadriceps activity. As a result, the vibration might prove useful for increasing quadriceps activity during BWSTT thereby decreasing the amount of required weight support and improving the efficacy of the treatment.

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