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Alternating stimulation of synergistic muscles during functional electrical stimulation cycling improves endurance in persons with spinal cord injury

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ABSTRACT

Therapeutic effects of functional electrical stimulation (FES) cycling for persons with spinal cord injury (SCI) are limited by high rates of muscular fatigue. FES-cycling performance limits and surface mechanomyography (MMG) of 12 persons with SCI were compared under two different stimulation protocols of the quadriceps muscles. One strategy used the standard "co-activation" protocol from the manufacturer of the FES cycle which involved intermittent simultaneous activation of the entire quadriceps muscle group for 400 ms. The other strategy was an "alternation" stimulation protocol which involved alternately stimulating the rectus femoris (RF) muscle for 100 ms and the vastus medialis (VM) and vastus lateralis (VL) muscles for 100 ms, with two sets with a 400 ms burst. Thus, during the alternation protocol, each of the muscle groups rested for two 100 ms "off" periods in each 400 ms burst. There was no difference in average cycling cadence (28 RPM) between the two protocols. The alternation stimulation protocol produced longer ride times and longer virtual distances traveled and used lower stimulation intensity levels with no differences in average MMG amplitudes compared to the co-activation protocol. These results demonstrate that FES-cycling performance can be enhanced by a synergistic muscle alternation stimulation strategy.

1. Introduction

The effective use of functional electrical stimulation (FES) to produce stationary pedaling as an exercise modality and during rehabilitation programs for people with spinal cord injury (SCI) is severely limited by high rates of muscular fatigue (Petrofsky and Laymon, 2004). Sustained or repeated electrical stimulation of paralyzed muscle typically does not produce the muscular force necessary for rehabilitation or exercise (Petrofsky and Stacy, 1992). Several factors relating to muscle fatigue may contribute to reduced muscular force output during FES cycling. Following paralysis, muscles often become weak as a result of disuse atrophy (Gordon and Mao, 1994). Histochemical changes occur which result in a shift towards fast-twitch, readily fatiguable muscle fibers (Round et al., 1993). The effect of these changes may be exacerbated by a random or reverse recruitment of electrically activated muscle fibers (Gregory and Bickel, 2005). Enhanced fatigue may also be inherent to the stimulation parameters that produce and control FES cycling.

FES parameters that can affect force output include the timing of muscle group stimulation and characteristics of the stimulation train, such as stimulation intensity, frequency, and duration (Binder-Macleod, 1995; Binder-Macleod et al., 1995; Doucet and Griffin, 2008). Commercial FES cycles typically use constant frequency patterns of stimulation that co-activate a group of synergistic muscles. Some studies have found that manipulation of the characteristics of the stimulation train during FES cycling, including varying inter-pulse intervals (Eser et al., 2003; Janssen et al., 2004; Thrasher et al., 2005) and electrical pulse properties (Szecsi et al., 2007), can improve FES-cycling performance. Another approach to enhancing the stimulation pattern during FES cycling may be to employ a synergistic muscle activation strategy that produces variable load sharing among multiple muscles. Alternation of electrical stimulation across different sites has been shown to be effective theoretically (Livshitz et al., 2001) and empirically for standing posture and indwelling nerve stimulation (Mizrahi, 1997).

During volitional fatiguing contractions in able-bodied individuals, pairs of synergistic muscles have been shown to alternate between activation and deactivation or increased and decreased activation levels, during force exertion. This alternation pattern has been suggested to contribute to enhance local blood flow (Sjogaard et al., 1986; Kouzaki et al., 2003) and to increase the length of time a voluntary contraction can be sustained (Tamaki et al., 1998;

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Kouzaki et al., 2002; Shinohara and Kouzaki, 2006). The application of an alternation stimulation strategy may be advantageous to FES cycling because this strategy may distribute activation among a set of synergistic muscles and delay fatigue by allowing individual muscles to rest during the deactivation periods.

Surface mechanomyography (MMG) amplitude is correlated to muscle force and surface EMG (Orizio, 1993) and has been found to be a reliable tool to study muscle fatigue in non-paralyzed muscles during cycle ergometry (Bull et al., 2000; Cramer et al., 2000; Perry et al., 2001a,b,c; Shinohara et al., 1997; Stout et al., 1997). MMG is a measurement of the sum of the mechanical contributions made by the lateral oscillations of the contracting skeletal muscle fibers (Orizio, 1993). The amplitude of the MMG signal depends on the magnitude of muscle fiber fluctuations under tension (Orizio, 1993) and it has been correlated with peak torque output during fatigue-induced fluctuations in muscle force (Gobbo et al., 2006).

The purpose of this study was to compare quadriceps FES-cycling performance during alternation and co-activation stimulation protocols in persons paralyzed from SCI. We hypothesized that a quadriceps muscle synergistic muscle alternation stimulation strategy would enhance FES-cycling endurance time compared to the standard co-activation protocol.

2. Materials and methods

2.1. Participants

Twelve individuals with clinical presentation of either paraplegia or tetraplegia resulting from SCI participated voluntarily in this study. None of the participants were trained FES users. Prior to participation, all individuals signed a written informed consent and provided a medical clearance form signed by their personal physician. Individuals were excluded from the study if they met any one of the following criteria: injury was within 1 year, skin breakdown which limited ability to sit for 30 min, diagnosed cardiovascular disease, use of cardiac pacemakers or defibrillators, vasomotor instability, severe loss of range of motion in joints, severe osteoporosis with risk of fractures, joint instability, heterotopic ossification, new fractures, pregnancy, epilepsy, frequent and severe bouts of autonomic dysreflexia, any other complications that would prevent them from participation as determined by their primary physician. The demographics of the study participants are shown in Table 1. All experimental procedures were approved by the Institutional Review Boards at the University of Texas at Austin and Seton Healthcare Network and all sessions were supervised by a licensed Occupational Therapist.

Table 1
Participant demographics.

Rider	Height (m)	Mass (kg)	Age (y)	Initial injury Level	Time from Initial injury (y)
1	1.85	71.4	41	T4	1.3
2	1.75	57.3	35	C6/7	1.7
3	1.91	93.2	49	C6/7	2.1
4	1.83	90.9	33	C3	14.8
5	1.63	52.3	38	T3/4	14.3
6	1.70	56.8	23	C6/7	5.1
7	1.83	100.0	53	T10	32.6
8	1.75	68.2	37	T6/7	10.6
9	1.83	90.9	31	C4/5	5.3
10	1.85	127.3	56	C3/4	14.7
11	1.78	90.9	23	C4	10.3
12	1.85	68.2	30	C4	10.7
Mean	1.80	80.6	37		10.3
SE	0.02	6.4	3		2.5

m: meters; kg: kilograms; y: years.

2.2. FES cycling

All participants performed FES cycling on an Ergys2 automated recumbent bicycle (Therapeutic Alliances, Dayton, OH, USA) with two different quadriceps stimulation protocols. The stimulation protocols were custom designed programs that were burned to a computer chip and placed in the hardware of the Ergys2 computer. One of two different stimulation protocols of the quadriceps muscles was used to produce continuous cycling motions during two experimental sessions, separated by at least 48 h. The order of the experimental stimulation protocols was randomly selected and balanced across subjects.

The quadriceps muscles were stimulated with co-activation and alternation stimulation protocols to produce the cycling motions. Fig. 1 shows a graphical representation of the protocols. The quadriceps muscles received electrical stimulation (50 Hz, sine wave, 500 μs pulse width) between crank angles of 314° and 30° (where 0° is top dead center). The periods of increasing and decreasing current (ramp up and down) normally used in the standard Ergys2 software were removed from both custom stimulation protocols. The Ergys2 software automatically increased the stimulus intensity up to a maximum of 140 mA to produce and maintain a target cadence of 32 RPM. FES cycling at this cadence resulted in intermittent stimulation sequences of 0.4 s that occur approximately every 1.88 s.

For the co-activation stimulation protocol, standard, two self-adhesive stimulating surface electrodes (oval, 2" by 4") were placed over the anterior thigh according to the Ergys2 manual (Fig. 2). The proximal electrode was shifted 2/3 of its width, laterally, from the midline of the quadriceps. The proximal edge of this electrode was no higher than the perineum. The distal electrode

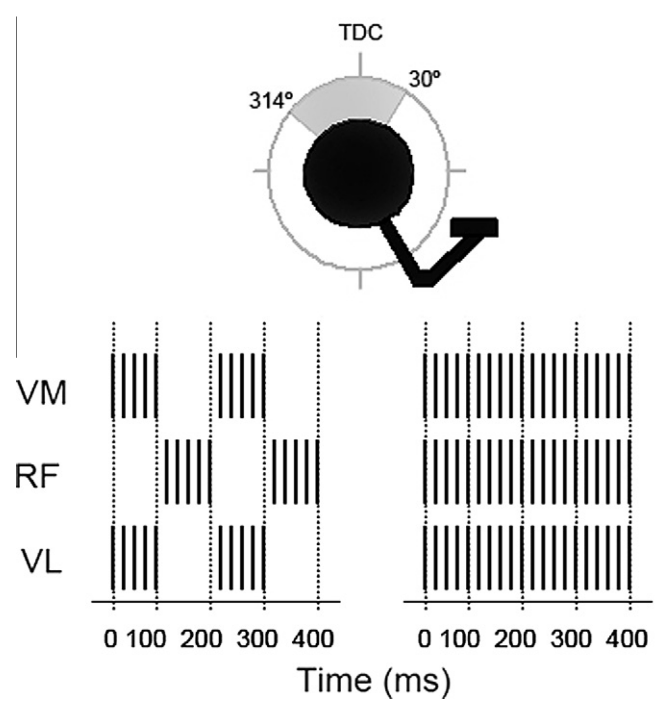

Fig. 1. Schematic of the stimulation trains (bottom) over the range of crank angles (top) used to make the alternation and co-activation stimulation protocols. Quadriceps stimulation started at 314° and lasted 400 ms. The co-activation pattern (bottom right) consisted of a constant frequency train (50 Hz, sine wave, pulse width of 500 μs) delivered simultaneously to the superficial quadriceps muscles (vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF)). The alternation stimulation pattern (bottom left) consisted of a series of four 100 ms constant frequency trains (50 Hz, sine wave, pulse width of 500 μs) where the VM and VL were stimulated 100 ms out of phase with the RF.

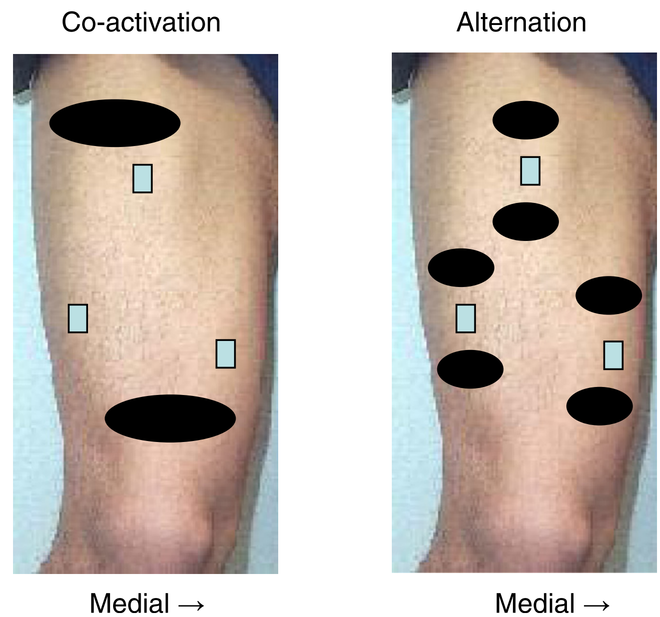

Fig. 2. Stimulating electrode (ovals) and accelerometer (rectangles) placements for the quadriceps muscles during the co-activation and alternation stimulation protocols.

was shifted 2/3 of its width medially, from the midline of the quadriceps muscles. The distal edge of this electrode was at least 1 in. superior to the patella. The co-activation protocol consisted of a 400 ms constant frequency train delivered simultaneously to the superficial quadriceps muscles (vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF)).

The alternation stimulation protocol consisted of a series of four 100 ms constant frequency trains where the VM and VL were stimulated 100 ms out of phase with the RF. A pair of self-adhesive stimulating surface electrodes (oval, 2" in by 2.5") was placed over the muscle belly of each superficial quadriceps muscle (Fig. 2). Smaller electrodes were used for this protocol so that the stimulation would be focused on specific muscles. The proximal edge of the RF electrode was no higher than the perineum. The distal edges of the VM and VL electrodes were at least 1 in. superior to the patella.

2.3. Experimental protocol

After a 1-min warm-up with assisted passive movements of the pedals, the stimulation intensity was increased automatically by the Ergys2 computer to assure the rider's cadence matched the target cadence of 32 RPM. If the cadence lowered during cycling, the stimulation intensity automatically increased up to a maximum of 140 mA. If the cadence dropped below 20 RPM, despite of the quadriceps muscles receiving the maximum stimulation intensity, the machine automatically started a 2-min cool-down mode with passive pedaling produced by the experimenter. All cycling was performed with zero resistance on the flywheel, so the work performed consisted primarily of the acceleration of the limbs. The participants had a total of five attempts to cycle for a total of 30 min during each experimental session with 5 min of rest between attempts. Each attempt was termed a "ride," thus there were a maximum of five rides performed during each session to attain the maximum 30 min ride time.

2.4. Mechanomyography

Uni-axial accelerometers (Entran EGAS, weight: 1 g, 2–150 Hz, gain of 25) were used to measure the surface MMG amplitudes of the VM, VL, and RF muscles. The rectangular accelerometers were placed on the muscle bellies between the stimulating electrodes and the length was aligned with the long axis of the femur (Fig. 2). The transducers measured accelerations perpendicular to its length where positive and negative deflections indicated medial and lateral muscle oscillations, respectively. MMG signals were sent online to an A/D board (Micro 1401, Cambridge Electronics Design, Cambridge, UK) and stored on a personal computer for further analysis. Offline, the root-mean-square (RMS, 50 ms bins) of the MMG signal for each muscle was calculated with Spike2 software (Cambridge Electronics Design, Cambridge, UK) for each 400 ms stimulation on-time during each participant's first FES-cycling ride for each stimulation condition.

2.5. Data analysis

The ride time and virtual distance traveled were summed over all FES-cycling attempts during each experimental session. These values were manually recorded from the digital output of the Ergys2 computer and averaged across participants. One-way repeated measures ANOVAs were used to contrast the effect of stimulation strategy on the total FES-cycling ride time, virtual distance traveled and cycling cadence across all rides. Given the wide range of subject characteristics, and the fact that muscle fatigability has been shown to increase after the initial injury (Gordon and Mao, 1994), Pearson correlations were implemented to determine the relationship between the time from injury and the total ride time.

Since the first ride was not 30 min for every participant, the data were partitioned in time into 10 segments, each representing 10% of the total ride duration. The stimulation intensity was manually recorded during each 30 s of ride time and then averaged over the 10% segments across participants. A two-way repeated measures ANOVA with Bonferroni post hoc tests was used to determine if the two stimulation strategies had a different stimulation intensity over each 10% segment of ride time and whether the stimulation intensity during the first 10% segment of ride time was lower than the other segments of ride time within each stimulation protocol.

The average MMG RMS amplitude measured for each muscle during both stimulation strategies was calculated across participants over the 10% segments of ride time. Analysis of the average MMG data changes over time were performed on the last nine segments, since during the first 10% of a ride, the data included start-up features unrelated to endurance. During this segment, the experimenter intermittently assisted the participant's cycling motions to level off fluctuations in stimulation intensity and cycling cadence. Thus, the fluctuations in the MMG amplitude often reflected features external to the rider. A three-way repeated measures ANOVA was used to determine the influence of ride time segment, muscle, and stimulation strategy on the average RMS MMG amplitudes. This procedure consisted of seven significance tests: a test for each of the three main effects (pattern, muscle and ride time segment), a test for each of the three two-way interactions (strategy $*$ muscle, pattern $*$ ride time segment, muscle $*$ ride time segment) and a test of the three-way interaction (strategy $*$ muscle $*$ ride time segment). Additional 1 or 2 way repeated measures ANOVA with Bonferroni post hoc tests were performed where appropriate.

All statistical calculations were performed with SPSS software package (SPSS Inc., Chicago, IL) and employed an alpha level of $p \leqslant 0.05$. The ANOVA procedures utilized Huynh–Feldt corrections for violations of sphericity where appropriate. All data are presented as mean ± standard error.

3. Results

There was no statistical difference in cycling cadence between the two stimulation strategies (co-activation: 27.7 ± 0.92 RPM;

alternation: 28.8 ± 0.80 RPM) ($F_{1,11}$ = 3.934, p = .073). Group means and standard errors for the total ride time and virtual distance traveled during FES cycling with co-activation and alternation stimulation protocols are presented in Table 2. The participants rode on average 2.36 min longer ($F_{1,11}$ = 6.830, p = .024) and 0.4 miles further ($F_{1,11}$ = 10.299, p = .008) when performing FES cycling with the alternation stimulation protocol compared to the co-activation protocol. The time from injury was negatively correlated with the total ride time for both the co-activation (r = −.599, p = .039) and alternation protocols (−.600, p = .039).

Group means and standard errors of the stimulation intensity used with co-activation and alternation stimulation protocols over 10% segments of ride time are presented in Fig. 3. The stimulation intensity was influenced by ride time segment ($F_{2.25,24.78}$ = 54.848, p < .001) and stimulation strategy ($F_{1,11}$ = 5.761, p = .035) but a statistical interaction between ride time and stimulation protocol was not found ($F_{2.30,25.26}$ = .988, p = .396). Post hoc testing revealed that FES cycling with the co-activation protocol required significantly greater stimulation intensity levels than the alternation protocol during the first three segments of ride time (first 30%) (all p < .05). Within each stimulation protocol, the first 10% segment of ride time had a statistically lower stimulation intensity level compared to all other ride time segments (last 90%) (all p < .05).

Group means and standard errors of the average MMG RMS data calculated over 10% segments of ride time during FES cycling with co-activation and alternation stimulation strategies are presented in Fig. 4. Average MMG RMS values were different across muscles but these values were not statistically influenced by the ride time segment or stimulation protocol (Table 3). Further, there were no statistically significant 2 or 3-way interactions between ride time segment, muscle or stimulation protocol (all p > 0.05). Collapsing the average MMG RMS data across the independent variables of ride time segment and stimulation protocol revealed

Table 2
FES-cycling variables with co-activation and alternation stimulation protocols.

	Ride time (min)*		Distance (min)*	
Rider	Co-activation	Alternation	Co-activation	Alternation
1	30.0	30.0	3.5	3.4
2	30.0	30.0	2.7	3.4
3	30.0	30.0	3.1	3.3
4	7.6	15.6	0.2	1.3
5	7.6	8.1	0.2	0.2
6	11.6	17.3	1.2	1.8
7	6.3	6.9	0.1	0.3
8	28.0	30.0	3	3.4
9	14.1	16.3	1.3	1.6
10	21.6	30.0	2	2.8
11	30.0	30.0	3.4	3.4
12	28.8	30.0	3.2	3.4
Mean	20.5	22.8	2.0	2.4
SE	2.9	2.7	0.4	0.4

min: minutes; mi: miles.
* Statistically significant (p < .05).

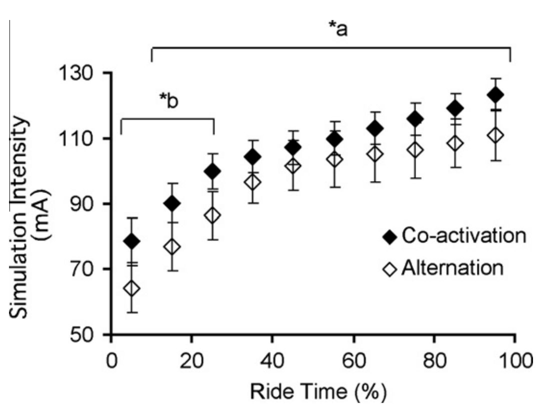

Fig. 3. Means and standard errors of the average stimulation intensity level for each stimulation pattern over 10% phases of FES-cycling ride time. *a: Stimulation intensity was larger in the last 90% of ride time compared to the first 10% for both stimulation patterns (p < .05). *b: Stimulation intensity levels were larger with the co-activation pattern compared to the alternation pattern during the first 30% of ride time (p < .05).

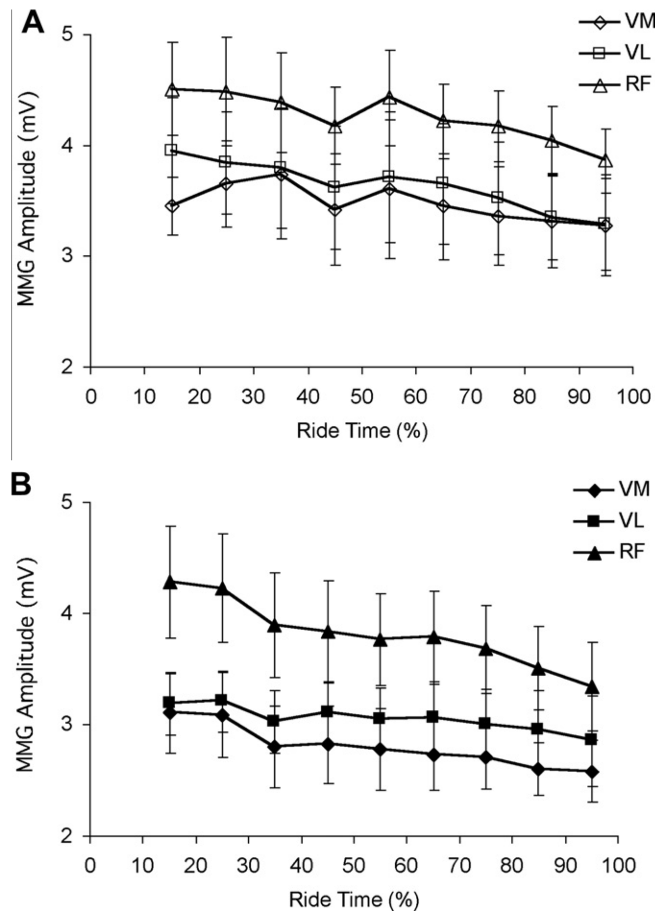

Fig. 4. Means and standard errors of the average MMG RMS amplitude for the vastus medialis (VM), vastus lateralis (VL) and rectus femoris (RF) muscles over 10% phases of ride time for with the (A) alternation stimulation pattern and the (B) co-activation stimulation pattern. No statistical differences were found between patterns or within a pattern over ride time.

Table 3
Source table for the 3-way RM ANOVA contrasting the average MMG RMS amplitude across stimulation pattern, muscle and time.

Source	SS	df	MS	F	Sig
Pattern	52.26	1	52.26	3.24	0.099
Error (pattern)	177.46	11	16.13		
Muscle	94.91	2	47.45	7.56	0.003
Error (muscle)	138.08	22	6.28		
Time	20.03	[a]2.36	8.50	2.55	0.09
Error (time)	86.33	[a]21.39	4.04		
Pattern by Muscle	1.48	2	0.74	0.06	0.942
Error (pattern by muscle)	270.83	22	12.31		
Pattern by Time	1.45	[a]2.51	0.58	0.29	0.799
Error (pattern by time)	55.53	[a]27.58	2.01		
Muscle by Time	1.72	16	0.11	1.44	0.128
Error (muscle by time)	13.16	176	0.08		
Pattern by muscle by time	1.80	16	0.11	1.22	0.254
Error (pattern by muscle by time)	16.17	176	0.09		

[a] Indicates Huynh–Feldt adjustment.

the average RF MMG RMS amplitude to be greater than the average VM and VL MMG RMS amplitudes (both $p < .05$).

4. Discussion

Functional electrical stimulation cycling has been shown to increase muscle volume (Skold et al., 2002), strength (Belanger et al., 2000), and endurance (Gerrits et al., 2000) and to decrease metabolic risk factors (Mahoney et al., 2005; Griffin et al., 2009). However, these improvements are largely dependent upon the length of time that an individual performs FES cycling at a given workload. Enhancing FES-cycling ride time and virtual distance traveled during the initial period of rehabilitation may be a means to enhance the efficacy of this exercise modality and promote functional improvements in persons with SCI. The results of the current study show that a stimulation strategy that alternates activation across synergistic muscles produces greater FES-cycling ride times and virtual distances traveled than the standard co-activation protocol.

The standard mode of maintaining FES-cycling performance is to progressively increase the stimulation intensity to promote muscle fiber recruitment (Crago et al., 1980) and utilize the whole cross-sectional area of the muscle. However, the stimulation intensity cannot be increased indefinitely. In addition, greater muscle fatigue is induced by higher compared to lower stimulation intensity levels. The alternation stimulation strategy in the present study utilized 10% lower average stimulation intensity levels with longer endurance times while producing the same average mechanical output from all muscles.

Cycling cadence was statistically equal during both stimulation conditions and suggests that the net quadriceps muscle force to accelerate the limb was also equal. However, researchers have shown that synchronous stimulation (co-activation) of the quadriceps muscles during isometric contractions produced greater net force output than asynchronous stimulation (Binder-Macleod and McLaughlin, 1997). The tension exerted by a group of synchronously activated muscles has to first overcome the compliance of the series elastic component prior to contributing to the net force. The benefit of asynchronous stimulation over synchronous stimulation is that the tension exerted by each muscle does not have to overcome most of the compliance of the muscle (Lind and Petrofsky, 1979). Thus, the alternation protocol utilized in the current study may have required smaller stimulation intensity levels because the alternating activation history influenced the engagement of the tendon, which permitted each muscle to exert their tension without having to overcome all of the elastic damping. Force preservation from this stimulation strategy would also be heightened by the additional intermittent rest periods and to an increase in blood flow from enhanced muscle pump activity which increases during intermittent bouts of electrical stimulation delivered to paralyzed leg muscles (van Beekvelt et al., 2000). These postulates are consistent the results of other studies that have found alternating muscle activation strategies to reduce fatigue in non-paralyzed leg muscles (Tamaki et al., 1998; Sjogaard et al., 1986; Kouzaki et al., 2002, 2003; Shinohara and Kouzaki, 2006).

Kouzaki et al. (2002) investigated alternating muscle activation patterns during sustained voluntary isometric quadriceps contractions at 2.5%, 5%, 7.5% and 10% MVC. The alternating muscle activation patterns were found to occur at the 2.5% and 5% MVC contraction intensity levels and the frequency of the alternations increased with time. Kouzaki and colleagues next performed a series of studies that investigated alternating muscle activation patterns of the knee extensor muscles during a 1 h sustained isometric contraction at 2.5% MVC (Kouzaki et al., 2003, 2004; Shinohara and Kouzaki, 2006). These studies found that the frequency of alternating muscle activation patterns was associated with local blood flow (Kouzaki et al., 2003), force steadiness (Kouzaki et al., 2004) and fatigue attenuation (Shinohara and Kouzaki, 2006). The distribution of muscle activation or electrical stimulation to synergistic muscles may be a robust biological fatigue prevention mechanism during fatiguing contractions.

The surface MMG signal is primarily generated by gross lateral movement of the muscle and dimensional changes of the active muscle fibers (Orizio, 1993). Fatiguing contractions typically reduce the MMG signal (Gobbo et al., 2006) because of increases in fluid content, muscle thickness, and intramuscular pressure (Sjogaard and Saltin, 1982; Sjogaard et al., 1986) that collectively decrease the compliance of the muscle by restricting muscle fiber oscillations and attenuating the pressure waves that generate the MMG signal (Perry-Rana et al., 2002). However, incremental increases of stimulation intensity (Orizio, 1993) or volitional effort (Shinohara et al., 1997) during fatiguing contractions can enhance the MMG amplitude. In the present study, the MMG amplitude decreased throughout ride time but this reduction failed to reach statistical significance. Similarly, Housh et al. (2000) found no changes in the VM and VL MMG amplitudes in able-bodied individuals during continuous cycle ergometry at 80% effort. These findings may be attributed to a balance between factors that increase and decrease MMG amplitude. In the present study, an increase in muscle fiber recruitment with increases in stimulation intensity and variation between the SCI participants may have masked the reductions in MMG amplitude caused by weaker contractions during fatigue.

The RF muscle was found to produce the largest MMG amplitude. This is in agreement with the results of other studies (Shinohara et al., 1998; Kouzaki et al., 1999). The RF composes 15% of the total cross-sectional area (CSA) of the quadriceps muscle group and the VM and VL is 60% of the total CSA (Kouzaki et al., 2004). Thus, greater RF MMG activation during the alternation stimulation pattern may have been due to increased mechanical stress when this muscle was being stimulated out of phase with the VM and VL muscles. Larger RF MMG amplitude during the co-activation pattern may have resulted from a disproportional stimulation level delivered to this muscle due to the proximal location of the large, proximal stimulating electrode. The disparity of muscle fiber type distributions of the superficial quadriceps muscles may have also contributed to higher RF MMG amplitudes (Johnson et al., 1973). In able-bodied individuals, the RF muscle has the largest percentage of the highly fatigable muscle fibers followed by the VL and VM muscles (Johnson et al., 1973). Muscles with the greatest percentage of highly fatigable fibers demonstrate the largest MMG amplitudes (Beck et al., 2007).

In summary, this study has revealed improvements in FES-cycling performance by a stimulation strategy which alternates activation of synergistic muscles in individuals with SCI. The synergistic muscle alternation protocol produced greater ride times and virtual distances traveled with lower stimulation intensity levels and no difference in mechanical output compared to the co-activation protocol. Future applications of an alternation stimulation strategy for FES training may be a means to promote longer training sessions and greater rehabilitative outcomes. For example, an increase in endurance time by modulating the stimulation protocol could also greatly improve FES walking systems since at present, average walking times of trained individuals are low. However, parameters such as overlap and intensity ramping of stimulation trains should be investigated to maximize overall force steadiness during FES training.

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