Impaired foot-force direction regulation during postural loaded locomotion in individuals poststroke

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Liang JN, Brown DA. Impaired foot-force direction regulation during postural loaded locomotion in individuals poststroke. J Neurophysiol 110: 378–386, 2013. First published April 24, 2013; doi:10.1152/jn.00005.2013.—Following stroke, hemiparesis results in impaired motor control. Specifically, inappropriate direction of footforces during locomotion has been reported. In our previous study (Liang and Brown 2011) that examined poststroke foot-force direction during a seated, supported locomotor task, we observed that foot-force control capabilities were preserved poststroke. In this current study, we sought to better understand the mechanisms underlying the interaction of locomotor and postural control as an interactive mechanism that might interfere, poststroke, with appropriate foot-force generation. We designed an experiment in which participants performed biomechanically controlled locomotor tasks, under posturally challenged pedaling conditions while they generated mechanical output that was comparable to pedaling conditions without postural challenge, thus allowing us to monitor the strategies that the nervous system adopts when postural conditions were manipulated. We hypothesized that, with postural influence, individuals poststroke would generate inappropriate shear forces accompanied by inappropriate changes to muscle activity patterns when performing a mechanically controlled locomotor task, and would be exaggerated with increased postural loading. Sixteen individuals with chronic poststroke hemiparesis and 14 age-similar nonimpaired controls pedaled on a cycle ergometer under 1) seated supported and 2) nonseated postural loaded pedaling conditions, generating matched pedal force outputs of two effort levels. When we compared postural influence with seated pedaling, we observed increased magnitudes of forward-directed shear forces in the paretic legs associated with increased magnitude of leg extensor muscle activity, but not in controls. These findings provide evidence to support a model that describes independent controllers for posture and locomotion, but that the interaction between the two controllers is impaired poststroke.

postural control; locomotor control; poststroke hemiparesis; pedaling

POSTSTROKE HEMIPARESIS RESULTS in significant long-term disability and impairments in locomotor functions of the paretic leg (Chen et al. 2005; Olney and Richards 1996). In order for successful locomotion to occur, appropriate control of leg muscles to generate adequate foot-force magnitude and direction is essential to propel the center of mass and prevent limb collapse (Farley and Ferris 1998; Neptune et al. 2001).

For example, during steady-state upright walking, the body center of mass is accelerated by propulsive and decelerated by braking anteroposterior ground reaction forces. During steady-state, level, straight-path walking, leg muscles are coordinated to regulate these ground reaction forces such that they are bilaterally symmetric (Turns et al. 2007). In individuals with poststroke hemiparesis, however, significantly asymmetric ground reaction force profiles between limbs have been reported. Inappropriately greater braking forces and reduced propulsive forces have been observed in the paretic legs, and these force profiles were associated with slower walking speeds (Bowden et al. 2006; Hall et al. 2011; Olney and Richards 1996). Functionally, insufficient horizontal forces could impact propulsion during overground walking (Bowden et al. 2006; Turns et al. 2007), but excessive horizontal forces on low friction or slippery surfaces could lead to slips and falls (Lanshammar and Strandberg 1981; Redfern et al. 2001).

In accordance with the walking studies, evidence from pedaling studies also suggested an impaired ability to control reaction forces in the paretic leg. When pushing along with a motor-driven crank generating different magnitudes of forces, paretic limbs exhibited a shift in the force path orientation, rotating it anteriorly away from the hip (Rogers et al. 2004). This suggests that paretic legs tend to generate forces in a direction that tends to destabilize the center of mass, potentially implying impaired force vector direction control in the poststroke nervous system.

We recently extended the earlier pedaling work by utilizing a more controlled pedaling paradigm, and contrary to earlier reported studies we found that, when attempting to generate a target force normal to the pedal, individuals poststroke were able to generate ratios of shear-to-normal pedal forces comparable to that generated by nonimpaired legs, despite the use of inconsistent strategies to modulate paretic leg muscle activity (Liang and Brown 2011). In this recent experiment, participants were seated on a bike seat and reclined and strapped to a backboard, such that minimal active control for posture was required. With this setup, we were able to study locomotor control in isolation from postural control mechanisms, to examine the neural control mechanisms of poststroke locomotion without the confounding effects of posture. We observed a smaller maximal force output as well as comparable magnitudes of shear pedal forces by the paretic legs compared with the nonimpaired legs during a stationary leg push task and a locomotor task (Liang and Brown 2011). Our results suggested that, despite weakness, the stroke-impaired nervous system was able to achieve foot-force directional control during a constrained locomotor task with minimal postural control demands.
However, in a more functional locomotor task, such as walking, the interaction between posture and locomotion is important and must be accounted for (Massion 1992). Locomotion is a goal-directed, purposeful movement, and posture refers to the maintenance of the body in an upright position and suspensory control against limb collapse. It has been proposed that the nervous system uses two different control mechanisms for movement and posture (Massion 1992). The control of movement utilizes a feedforward control to produce accurate, goal-directed effortful movement, and the control of posture was hypothesized to use a feedback control based on the loading dynamics of the body relative to the base of support. Both the controls for movement and the controls for posture interact to act on locomotion networks to produce coordinated motor output (Massion 1992). Therefore, we hypothesized that the lack of a directional force deficit poststroke during seated, supported pedaling was due to the isolation of the locomotor control component from the postural control component.

In another earlier experiment that examined the pedaling locomotor task with different levels of weight-bearing, using a sliding backboard that allowed nonseated pedaling which involved postural control, it was reported that individuals poststroke exhibited reduction in coordination control, which worsened with increased weight-bearing but not with increased effort during seated pedaling (Burgess and Brown 2011). With respect to forces, at high levels of postural loading, the paretic leg was unable to produce comparable levels of pedal resultant forces compared with the nonparetic leg during weight-bearing pedaling. At lower levels of postural loading, appropriate force production by the paretic leg was observed, generating symmetric forces, similar to nonimpaired individuals, suggesting that postural control mechanisms interfere with symmetric force production in the poststroke nervous system. Thus the question arises as to whether some of the difficulties with appropriate foot-force direction during locomotor tasks experienced by individuals poststroke are attributable to an impaired interaction between control of locomotion and posture.

Therefore, we sought to investigate the ability of the stroke-impaired nervous system to control foot-forces during locomotion under demands of postural control. The main purpose of this experiment was to test whether the interaction between controllers for posture and the controllers for locomotion is affected in a poststroke nervous system. To test this, we took advantage of the two different pedaling tasks that our apparatus can produce: one that placed minimal demands on postural control during a locomotor task (i.e., seated pedaling), and the other that provided varying level of postural control during a locomotor task (nonseated pedaling). According to the idea that the interaction of posture and locomotion is impaired poststroke, we hypothesized that, in a stroke-impaired system, we would observe greater inappropriate shear pedal forces during the posturally involved pedaling task compared with the posturally supported task, which would not be present in nonimpaired individuals. We expected this inappropriate shear force to be further exaggerated with greater postural load levels and would be accompanied by inappropriately higher paretic leg extensor muscle activity during postural loading in the stroke-impaired group but not in nonimpaired individuals. Preliminary findings of this study were published as an abstract (Liang and Brown 2012).

METHODS

Subjects. Participants in this study were 16 individuals (14 men, 2 women; age = 59.5 ± 7.2 (SD) yr), who had sustained a single, unilateral, cortical, or subcortical stroke, more than 12 mo postictus [136.2 ± 76.9 (SD) mo] before the study and had residual lower limb hemiparesis without lower limb contractures. Fourteen age-similar nonneurologically impaired individuals (age = 57.6 ± 12.6 yr) were recruited as controls. Ambulatory ability of individuals poststroke ranged from independent ambulation without assistive devices to independent ambulation with cane/quad-cane/ankle-foot orthosis. Participants were excluded if they had other neurological conditions, severe cognitive or affective disorders, expressive or receptive aphasia, severe concurrent medical problems (e.g., severe cardiac disease, history of poorly controlled brittle diabetes, active cancer, etc.), orthopedic conditions affecting the legs, or history of hip or knee replacement. Each participant received written and verbal information about the experiment procedures before giving written consent. The protocol was approved by the Institutional Review Board at Northwestern University.

Experimental apparatus. A custom-made cycle ergometer (Fig. 1A) with instrumented pedals, a seat with backrest, and a motor-driven crank was used for this study (Rogers et al. 2004). Participants were seated on the bike with the torso stabilized against the backrest to maintain constant hip position. Optical encoders (BEI model EX116-1024-2), one at each pedal spindle and one coupled to the right crank, provided measurements of the pedal angles and the crank position with ±0.3° accuracy. Force transducers in each pedal measured the three-dimensional foot/pedal force vector (Delta 660, ATI-IA, Garner, NC). A custom-made boot with Velcro straps was attached to each pedal to fix the ankles at a neutral position during the pedaling tasks. Pedaling velocity was controlled by an electric motor (12:1 gear reducer, 3.7 hp, model MT506B1-SC1C, Kollmorgen, Radford, VA) and was kept constant at 40 revolutions/min (rpm) for all subjects during experiment. We adopted this particular velocity because a previous study in our laboratory reported, for an imposed velocity of 40 rpm, the motor can accurately regulate actual crank speed, despite large applied forces (40.5 ± 0.8 rpm) (Rogers et al. 2004).

Recordings. Bipolar silver surface electrodes (DeiSyS, 1 cm length, 1 mm wide, 1 cm interelectrode distance) were used to record the electromyograms (EMG) from five muscles on the test leg: vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), soleus (SOL), and tibialis anterior (TA). EMG signals were amplified with a gain of 10 at the electrode site before remote differential amplification (CMRR: 92 dB, gain range 100–1,000 times, frequency response 20–450 Hz) and were low-pass filtered (custom-designed filter, 500 Hz cutoff). The signals from the optical encoders and force-transducers were converted from digital to analog with a D/A converter module before sampling. All signals were then sampled at 1,000 Hz via a 12-bit A/D converter (National Instruments) and custom LabView software. The timing of EMG data collection was synchronized with the acquisition of position data from the crank optical encoder.

Experimental paradigm. While seated on the bike seat, and with the crank fixed at 90° in the middle of the downstroke (crank angle defined relative to top-dead-center), participants were instructed to generate three maximal force efforts with their test leg, which was the paretic leg for individuals poststroke and the dominant leg for nonimpaired individuals. The normal pedal force vector (Fp) magnitude generated from the maximal pushes (max Fp) was averaged across the three trials.

Each participant pedaled with the motor-driven crank moving at 40 rpm under two conditions: 1) seated, nonpostural loaded condition, and 2) nonseated, postural loaded condition. For each condition, there were two effort levels: 1) low-effort level (30% max Fp) and 2) high-effort level (50% max Fp).

For the seated, nonpostural loaded condition, participants, while seated on the bike seat, were instructed to assist the motor actively by pedaling
normal pedal force.

were required to sustain 30 s of continuous pedaling without collapse weight, while pedaling along with the moving crank. Participants to actively push away from the seat, supporting their own body based upon their particular body weights. Participants were instructed determination of the tilt angle was done on a subject-by-subject basis, tal. Bar graphs with real-time pedal FN were displayed as visual feedback conditions, the tilt angle of the backboard was kept at 40° from horizon-

yield a peak FN of 30% and 50% max FN, matching the target 30% (FS) with negative values indicate anteriorly directed shear, and FS values with negative values indicate forces directed downwards (Fig. 1B). For each pedaling trial, we averaged the FN profiles over the 20 cycles sampled. The crank angle position at which the peak target FN was achieved was used as an index to identify the corresponding FS value. For each task and effort level, we averaged the peak FN achieved across cycles and used independent t-test to compare the mean FN between seated and nonseated pedaling conditions for each effort level, to ensure we were able to match the target FN across conditions. To normalize the FN magnitude between subjects generating different force magnitudes, we expressed the forces as the ratio of FS per unit FN (FS/FN). Using independent t-test, we compared the mean FS/FN generated under each pedaling condition between groups for both 30% and 50% effort levels. A two-way repeated measures ANOVA was used to analyze the FS/FN generated for each group. The independent variables include load (seated, nonpostural loaded and nonseated, postural loaded) and effort (30% and 50%). No post hoc analysis was performed, as no interaction effect was observed.

EMG signals were rectified and integrated for the entire down-stroke phase (0° ~ 180° of the crank cycle with respect to top-dead-center) and averaged across cycles. All EMG profiles were smoothed with a fourth-order, zero-lag, low-pass Butterworth filter with a cutoff frequency of 25 Hz. The averaged EMG activity for each muscle was expressed as a percent change from the EMG amplitudes in the seated 30% effort condition, with a positive value indicating increased EMG amplitude, and a negative value indicating decreased EMG amplitude. A two-way repeated-measures ANOVA was used to analyze the EMG amplitude for each muscle for each group. The independent variables include load (seated, nonpostural loaded and nonseated, postural loaded) and effort (30% and 50%). A further post hoc analysis was conducted as there was an interaction effect. A P value less than or equal to 0.05 was considered statistically significant.

RESULTS

In individuals poststroke, we recorded lower maximum effort force output for the paretic leg compared with the nonimpaired group target leg. The maximum effort FS achieved by individuals poststroke (499.5 ± 22.6 N) were, on average, 27.7 ± 3.5% lower than that by nonimpaired individuals (690.9 ± 39.2 N) when the means were compared using independent t-test (P < 0.001).

During the nonseated conditions, we were successful in matching FS with the targeted FS that was generated during the seated conditions for a comparable mechanical output. For both groups, we did not observe a statistical significant difference in FN values for seated vs. nonseated conditions during both low (30%) and high (50%) effort levels (P > 0.05). Since FS magnitude is partially dependent upon overall push effort and foot orientation relative to the pedal (Liang and Brown 2011), similar FS values generated when the foot was locked in a neutral ankle position allowed us to make valid comparisons of foot force direction under the seated and nonseated conditions within and between each group. Furthermore, on average, due to the weaker force-generating capability of individuals poststroke, the tilt angle was lower compared with nonimpaired individuals for the 50% effort (20.9 ± 3.1° for stroke-impaired and 26.1 ± 4.6° for nonimpaired) (P < 0.01), but were comparable between groups for the 30% effort (15.6 ± 2.6° for stroke-impaired and 16.9 ± 2.6° for nonimpaired) (P > 0.05).

Foot force direction: seated vs. nonseated. When we com-
pared, using independent t-test, the mean FS/FN generated

along with the moving crank in the forward direction, generating a target FS of 30% and 50% max FS with visual feedback. For all seated pedaling conditions, the tilt angle of the backboard was kept at 40° from horizontal. Bar graphs with real-time pedal FN were displayed as visual feedback on a monitor. Each pedaling trial lasted 30 s.

For the nonseated, postural loaded condition, the bike seat was lowered, and the backboard was unlocked to slide for an additional postural component to the locomotor task. B: pedal force orientation. SOL, soleus; VM, vastus medialis; TA, tibialis anterior; RF, rectus femoris; BF, biceps femoris; FS, shear pedal force; FN, normal pedal force.

Fig. 1. A: custom-made cycle ergometer with instrumented pedals, a seat with backrest, and a motor-driven crank was used. The backrest was locked in place to allow a locomotor task with minimal demands for postural control, or was unlocked to slide for an additional postural component to the locomotor task. B: pedal force orientation. SOL, soleus; VM, vastus medialis; TA, tibialis anterior; RF, rectus femoris; BF, biceps femoris; FS, shear pedal force; FN, normal pedal force.

Data processing and analysis. All force and EMG data were processed using custom MATLAB programs. Force profiles were expressed in the pedal coordinate system where shear pedal forces (FS) with negative values indicate anteriorly directed shear, and positive values indicate posteriorly directed shear, and FS values with negative values indicate forces directed downwards (Fig. 1B). For each pedaling trial, we averaged the FS profiles over the 20 cycles sampled. For each task and effort level, we averaged the FS achieved across cycles and used independent t-test to compare the mean FS between seated and nonseated pedaling conditions for each effort level, to ensure we were able to match the target FS across conditions. To normalize the FS magnitude between subjects generating different force magnitudes, we expressed the forces as the ratio of FS per unit FN (FS/FN). Using independent t-test, we compared the mean FS/FN generated under each pedaling condition between groups for both 30% and 50% effort levels. A two-way repeated measures ANOVA was used to analyze the FS/FN generated for each group. The independent variables include load (seated, nonpostural loaded and nonseated, postural loaded) and effort (30% and 50%). No post hoc analysis was performed, as no interaction effect was observed.

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Foot force direction: seated vs. nonseated. When we com-
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under each pedaling condition between groups, we found comparable ratios of $F_S/F_N$ between groups for both 30% and 50% effort levels during the seated pedaling condition ($P > 0.05$), which was in accordance with the findings from our earlier study that indicated foot force direction was not altered poststroke during seated pedaling (Liang and Brown 2011). When asked to pedal in the nonseated condition, we observed that individuals poststroke generated large magnitudes of forward shear force, which was in stark contrast to nonimpaired subjects who generated small amounts of posteriorly directed shear forces. For the nonseated pedaling conditions, individuals poststroke generated significantly greater forward-directed shear forces (30% effort = 0.11 ± 0.04; 50% effort = 0.20 ± 0.04) than nonimpaired individuals (30% effort = −0.10 ± 0.03; 50% effort = −0.004 ± 0.03) under both 30% and 50% effort levels ($P < 0.05$). For individuals poststroke, when comparing the $F_S/F_N$ ratio generated for each task, a two-way ANOVA yielded a statistically significant main effect for seated vs. nonseated condition ($P < 0.01$), such that the average $F_S/F_N$ ratios were significantly higher in the anterior direction for nonseated pedaling (0.16 ± 0.013) compared with seated pedaling (0.10 ± 0.013). However, for nonimpaired individuals, at both effort levels, during the nonseated conditions, the average $F_S/F_N$ ratio was less anteriorly directed and more posteriorly directed (−0.05 ± 0.009), compared with the seated conditions (0.16 ± 0.009) ($P < 0.001$) (Fig. 2).

Foot force direction: low- vs. high-effort levels. Furthermore, not only did the paretic legs generate greater anteriorly directed shear during nonseated conditions, the anteriorly directed shear was exaggerated with increased effort levels, as demonstrated by a statistically significant main effect for effort level ($P < 0.01$), such that the average $F_S/F_N$ ratios were significantly higher in the anterior direction for the 50% effort (0.162 ± 0.013) than for the 30% effort (0.094 ± 0.013) level. For nonimpaired individuals, a two-way ANOVA yielded a statistically significant interaction effect ($P < 0.001$) such that, during nonseated conditions only, there was a greater ratio of posteriorly directed $F_S/F_N$ at 30% effort levels (−0.104 ± 0.03) than for 50% effort levels (−0.004 ± 0.03) ($P < 0.001$) (Fig. 2).

Muscle activity: seated vs. nonseated. We expressed the EMG activity of each muscle as a percent change from the seated, low effort (S30) condition, a positive value indicated an increase from S30, while a negative value indicated a decrease compared with S30 condition. In the paretic leg extensors, a two-way ANOVA yielded a significant interaction effect ($P < 0.05$), such that the overall effect for paretic VM amplitudes was significantly higher during nonseated conditions (146.7 ± 15.8%) than seated conditions (44.6 ± 15.8%) only for the 50% effort level ($P < 0.001$). With the nonimpaired VM, a two-way ANOVA yielded a significant interaction effect ($P < 0.01$), such that the VM amplitudes during nonseated conditions (31.2 ± 14.0%) were significantly lower than during seated conditions (112.1 ± 15.8%) only during the 50% effort level ($P < 0.01$) (Fig. 3A).

With the SOL in nonimpaired individuals, we did not observe a statistically significant difference in amplitude between seated and nonseated conditions ($P > 0.05$). With paretic SOL, simple main effects for load showed that, during nonseated conditions (79.2 ± 15.4%), SOL amplitudes were significantly greater than seated conditions (13.2 ± 15.4%) ($P < 0.01$) (Fig. 3A).

With the leg flexor muscles, nonimpaired individuals showed reduced activity during nonseated compared with seated conditions, whereas in the paretic flexors, we observed inappropriate nonchanging flexor muscle activity during nonseated compared with seated conditions. With the paretic RF, no interaction or main effects for load were observed. With nonimpaired RF, simple main effects showed that RF amplitude during nonseated conditions (−3.7 ± 31.3%) were significantly lower than during seated conditions (110.7 ± 31.3%) ($P < 0.05$). With the paretic BF and TA, there were no significant main effects for load ($P > 0.05$). With nonimpaired BF, the interaction effect was significant ($P < 0.05$), such that the BF amplitude during nonseated conditions (−2.7 ± 9.5%) were significantly less than during seated conditions (55.7 ± 9.5%) ($P < 0.001$) only during 50% effort levels. With nonimpaired TA, the interaction effect was significant ($P < 0.05$), such that the TA amplitude during nonseated conditions (2.9 ± 12.1%) was significantly less than during seated conditions (80.8 ± 12.1%) ($P < 0.001$) only at the 50% effort level (Fig. 3B).

Muscle activity: low- vs. high-effort level. With the leg extensors, nonimpaired VM amplitude appropriately increased with greater effort during seated pedaling only, and remained unchanged with effort during nonseated conditions. This was a different response compared with the paretic VM, which re-

![Fig. 2. Shear force per unit normal force ($F_S/F_N$) generated under each pedaling condition and effort level. A positive value of $F_S/F_N$ ratio indicates forward-directed $F_S$, and a negative ratio indicated backwards-directed $F_S$. Black bars represent nonimpaired individuals (top); gray bars represent individuals poststroke (bottom). Statistical significance at $P < 0.05$ level: *between effort, §between conditions.](http://jn.physiology.org/doi/abs/10.1152/jn.00005.2013?journalCode=jn)
Fig. 3. Change in EMG activity expressed as a percent change from seated pedaling conditions at 30% effort level for leg extensors (SOL, VM; A) and leg flexors (TA, RF, BF; B). Black bars represent nonimpaired individuals; gray bars represent individuals poststroke. Statistical significance at $P < 0.05$ level: *between effort, §between conditions.
mained inappropriately unchanged during seated conditions and inappropriately increased with effort during nonseated conditions. With the ankle extensors, both the nonimpaired SOL and the paretic SOL increased amplitude with increased effort under both seated and nonseated conditions. In individuals poststroke, as a result of the significant interaction effect \((P < 0.05)\), paretic VM amplitudes were significantly higher for 50% effort levels \((146.7 \pm 15.8\%)\) compared with 30% effort levels \((38.0 \pm 15.8\%)\) \((P < 0.001)\) in the nonseated condition only, with unchanged activity during seated conditions. Contrarily, in nonimpaired individuals, as a result of the significant interaction effect \((P < 0.01)\), VM amplitudes, on average, were significantly higher for 50% effort levels \((71.6 \pm 6.9\%)\) than 30% effort levels \((-8.4 \pm 6.9\%)\) \((P < 0.01)\) for both seated and nonseated conditions. With the SOL muscle, simple main effects showed that the paretic SOL amplitudes at 50% effort level \((74.9 \pm 15.4\%)\) were significantly greater than those at 30% effort level \((17.5 \pm 15.4\%)\) \((P < 0.05)\). For nonimpaired individuals, simple main effects showed that, with SOL, amplitudes at 50% effort level \((62.5 \pm 8.6\%)\) were significantly greater than those at 30% effort level \((6.2 \pm 8.6\%)\) \((P < 0.001)\) (Fig. 3A).

With the leg flexors, nonimpaired individuals increased flexor activity with increased effort, whereas paretic flexors remain inappropriately unchanged with increased effort. With the nonimpaired RF, simple main effects showed that RF amplitudes were significantly higher for 50% effort levels \((119.9 \pm 31.3\%)\) than 30% effort levels \((-12.9 \pm 31.3\%)\) \((P < 0.01)\). In contrast, with paretic RF, no main effects or interaction effects were observed \((P > 0.05)\). With the nonimpaired BF, as a result of the interaction effect \((P < 0.05)\), BF amplitudes were significantly higher for 50% effort levels \((55.7 \pm 9.5\%)\) than 30% effort levels \((0.0 \pm 9.5\%)\) \((P < 0.001)\) during the seated condition only, whereas, in individuals poststroke, simple main effects showed that the paretic BF amplitudes were significantly higher for 50% effort levels \((34.2 \pm 4.6\%)\) than 30% effort levels \((0.8 \pm 4.6\%)\) \((P < 0.001)\). With the nonimpaired TA, due to the significant interaction effect \((P < 0.05)\), TA amplitudes were significantly higher for 50% effort levels \((80.8 \pm 12.1\%)\) than 30% effort levels \((0.0 \pm 12.1\%)\) \((P < 0.001)\) during the seated condition only, whereas with paretic TA, no interaction or main effects with effort level were observed (Fig. 3B).

**DISCUSSION**

With this study, with respect to pedal forces, first, we validated our earlier study where we reported comparable magnitudes of anteriorly directed \(F_S\) in both stroke and nonimpaired individuals during seated pedaling (Liang and Brown 2011). Second, we found, in the stroke-impaired individuals, that these anteriorly directed shear forces were inappropriately increased with nonseated pedaling, and further exaggerated with increased postural loads. These results supported the main hypothesis that, in a stroke-impaired system, we would observe greater inappropriately directed \(F_S\) during the posturally involved pedaling task (i.e., nonseated pedaling) when compared with a nonposturally involved pedaling task (i.e., seated pedaling), and that these forces would be exaggerated with greater postural load levels. This result was also in contrast with nonimpaired individuals, where we observed during the nonseated pedaling conditions that the \(F_S\) were directed posteriorly. With respect to muscle activity, the hypothesis that we would observe inappropriately exaggerated paretic leg extensor muscle activity during postural loaded pedaling conditions was also supported. It is also important to emphasize that, since we were successful in matching the \(F_N\) magnitudes during seated and nonseated pedaling, the differences in shear force direction under postural loaded pedaling conditions that were observed between subject groups, occurred under comparable magnitudes of \(F_N\) values.

Previously, we showed that, during a seated pedaling task, the \(F_S\) generated is influenced by magnitude of \(F_N\) as well as posture of the ankle joint (Liang and Brown 2011). In this present study, we designed the experimental task such that, during nonseated pedaling, the ergometer was tilted at an angle to yield a \(F_N\) that matched the target \(F_N\) values achieved during the seated pedaling condition. We also visually examined the pedal angle trajectories data to ensure that the ankle joint was well constrained in a neutral posture by our custom-designed boot during pedaling. Furthermore, the effects of gravity on the backboard orientation could also affect the shear forces. However, first, the differences in tilt angle, on average, between the poststroke and nonimpaired individuals were relatively small \((-1°\text{ at 30% effort}, -5°\text{ at 50% effort})\). Second, our data analysis compared changes in shear forces under different conditions within subject, so that significant differences were calculated relative to each person’s response. Third, theoretically speaking, a mechanical linkage system that matched the mechanics of this pedaling task (4-bar linkage) that was placed at a greater verticality would be expected to have reduced backward shear forces, and at some verticality would generate greater forward shear force. In contrast, we observed that individuals poststroke who were, on average, tested at lower tilt angles expressed greater forward shear forces so that the slight differences in tilt angle between the two groups do not contradict the interpretation of our results. By controlling these factors, we were able to eliminate the confounding factors of force magnitude, foot posture, and backboard inclination, which enabled us to determine whether differences in footforce direction were due to an impaired postural/movement interaction poststroke. The comparable magnitude of \(F_S\) per unit \(F_N\) observed during the seated pedaling conditions under both effort levels for both groups was in line with our earlier findings (Liang and Brown 2011). \(F_S/F_N\) is a valid indicator of foot force direction because it expresses the ratio of two orthogonally directed force values.

To the best of our knowledge, this study is the first to compare a mechanically constrained paradigm that allowed execution of a locomotor task with minimal postural control demands (i.e., seated, nonpostural loaded pedaling) as well as a locomotor task with different levels of postural control (i.e., nonseated, postural loaded pedaling) to investigate the interaction of control for posture and locomotion, comparing nonimpaired and stroke-impaired systems. The controlled pedaling setup that we used allowed us to control variables that are very difficult to control during body weight-supported walking studies. For example, in upright walking studies that used body weight support (Aaslund et al. 2012; Hesse et al. 1999), loading on the legs is adjusted by a harness that reduces biomechanical and equilibrium walking constraints. While such studies were able to study the effect of lower limb loading
on locomotion, the nature of the suspension could not constrain postural control mechanisms from interacting with locomotor control, where, even under full suspension, there remained complex and unconstrained interaction between gravity and the suspension harness.

Similar to other studies, including our previous study, we observed impaired force output capabilities in individuals poststroke. In the paretic legs, maximum effort pushes yielded lower $F_N$ output compared with the test leg of nonimpaired individuals. This observation was in accordance with our earlier study (Liang and Brown 2011), as well as earlier experiments documenting weakness in the affected limbs following cerebral stroke (Canning et al. 1999; Harris et al. 2001), possibly resulting from disrupted descending commands (Adams et al. 1990; Schneider and Gautier 1994). However, despite the commonly observed force deficit, we observed greater magnitudes of $F_S$ that were directed anteriorly, which suggests a disorder related to foot force direction.

In our previous experiment (Liang and Brown 2011) and with this current, more controlled version of the seated pedaling experiment, we found comparable magnitudes and direction of $F_S$ when both nonimpaired and stroke-impaired participants pedaled while seated on the bike seat, suggesting that, under minimal demands for postural control, the stroke-impaired nervous system is able to direct foot forces appropriately during pedaling. The effects of limb unweighting on paretic limb motor performance have been more extensively studied in the upper extremities. Reduced loading of the paretic arm during a reaching task results in improved motor performance with respect to greater range of motion as well as independent joint control (Beer et al. 2004, 2007), supposedly attributable to alleviation of the increased influence from less specific, intact bulbospinal pathways, as the availability of corticospinal resources are sufficient at less challenging levels of effort (Sukal et al. 2007).

In nonimpaired individuals, during the nonseated pedaling condition, there was posteriorly directed shear force, accompanied by less extensor muscle activity, primarily the VM muscle. This result can be explained by the posteriorly directed gravitational influence on the weight of the limbs and by the assistance from gravity on the center of mass during the downstroke of the pedaling cycle that aided virtual propulsion of the crank and required less VM muscle activity to accomplish this task. In contrast, with the paretic leg extensors, we observed an exaggerated anteriorly directed shear forces accompanied by increased activation of VM activity during nonseated pedaling. This was observed even in the ankle extensors SOL, whose activity was not required to contribute to force output (due to the locked boot that fixed the ankle in a neutral position). We observed greater paretic ankle extensor activity during nonseated compared with seated pedaling, but not in nonimpaired individuals, which was expected because the ankle extensors were not required to contribute in our task, since the ankles were stabilized at neutral by the custom-made boots. However, we observed an inappropriate increase in paretic SOL activity with postural loading, and further exaggerated with higher postural loads, suggesting a synergistic neural coupling of muscle activity between two different extensor muscle groups. This unnecessary obligatory coupling of extensor muscles has been documented in individuals with chronic poststroke hemiparesis both in the upper (Dewald and Beer 2001) and lower extremities (Cruz and Dhaher 2008), and kinematic data suggest this unnecessary coupling negatively impacts movement performance, such as upper limb reaching tasks in the presence of gravity or external loads (Beer et al. 2004, 2007; Sukal et al. 2007). In the stroke-impaired nervous system, there may be a disrupted integration of sensory afferent input with descending motor commands, resulting in the underlying observed inappropriate paretic limb joint and reflex coupling (Finley et al. 2008; Kloter et al. 2011). Furthermore, earlier studies, mostly on upper extremities, postulated that, after injury to the corticospinal tract, there is increased dependence on the reticulospinal tract (Dewald and Beer 2001; Jankowska and Edgley 2006; Sukal et al. 2007). In terms of locomotor control, the reticulospinal tract has been suggested to be responsible for the coordination of postural and locomotor control (Mori et al. 1998). This increased and altered reliance on the reticulospinal tract in the poststroke nervous system can possibly impact the integration of postural and locomotor control.

One important feature in our experimental design when using the two different types of pedaling was that two different modes of load were provided, that is, an seated, nonpostural loaded pedaling task, and an nonseated, postural loaded pedaling task. The seated, nonpostural loaded task required participants to generate targeted magnitudes of $F_N$ given visual feedback by pushing with their legs with different efforts, whereas for the nonseated, postural loaded task, participants were required to actively push their body off from the seat and maintain a percentage of their body weight while pedaling without collapsing. Thus there were comparable magnitudes of loading input to the load receptors under both conditions. Therefore, differences in response between seated and nonseated pedaling are less likely due to the loads that were experienced by peripheral receptors of the lower extremity. Another possible underlying control scheme was proposed by Massion (1992), where two different motor controllers, one for movement and one for posture, are thought to be involved in posturally dependent movement tasks, such as walking and (we suggest) nonseated pedaling. The control of posture utilizes a feedforward model of limb dynamics to produce cortically driven purposeful movements (Georgopoulos 1988; Ghez and Krakauer 2000), whereas the control of posture utilizes continuous and discontinuous feedback regulation from visual, labyrinthine, and proprioceptive inputs (Massion 1992, 1998). Both of these controllers act on locomotion networks to produce effective, goal-directed motor output. Neurophysiological data supporting this theory indicated that these two separate controllers are interdependent at different levels of the central nervous system. In the cat, this interaction between pathways controlling posture and gait has been proposed to exist at the hypothalamus and brain stem level, as well as at the level of spinal premotor interneurons (Jankowska and Edgley 1993), where stimulation could trigger changes in posture before gait initiation (Mori 1983, 1987, 1989).

In this present study, we attempted to examine the force control capabilities with and without postural control demand, and our results of exaggerated anteriorly directed shear forces in individuals poststroke only during nonseated, postural loaded pedaling, but not during seated, nonpostural loaded pedaling suggested that the posture component possibly interferes with the foot force control capabilities in the stroke-

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impaired system, and that the interaction between the two controllers is impaired, resulting in misdirected foot-forces when locomotion and posture are required to be regulated simultaneously.

Limitations and future studies. The nonseated pedaling task was a challenging task for stroke-impaired individuals who, as described, were much weaker than the nonimpaired, age-similar control subjects. To ensure that all subjects were able to complete the task, we limited the recruitment of stroke-impaired individuals to relatively higher functioning individuals. Despite the relatively higher functional status, the poststroke participants were still impaired in motor function and force output; therefore, we set the target force levels as a percentage of each individual’s maximum output capacity. We also recognize that we must use caution when using a pedaling paradigm to study locomotion, as generalizing these results to walking could be limited. However, a line of research in our laboratory has used this paradigm with much success (Alibiglou and Brown 2011a, 2011b; Alibiglou et al. 2009; Brown and Kautz 1998; Brown et al. 1996; Burgess and Brown 2011, 2010; Schindler-Ivens et al. 2004), and the results were consistent with current models of locomotor control that explained forward and backward pedaling (Raasch and Zajac 1999; Schindler-Ivens et al. 2004; Ting et al. 1999), bilateral reciprocal control (Alibiglou and Brown 2011b; Alibiglou et al. 2009; Ting et al. 1998, 2000), and central vs. peripheral control of muscle activation patterns (Rogers et al. 2011b). The pedaling paradigm has also been used to study spinal reflex behavior during constrained locomotor behavior (Fuchs et al. 2011; Rogers et al. 2011a; Schindler-Ivens et al. 2008), and we plan to study postural control/locomotion interactions, as a way to determine whether the control of reflex stiffness is altered poststroke. Also, we would like to take advantage of new robotic systems that can be used to control postural mechanisms so that we can examine the interaction between control for posture and locomotion under walking conditions over the treadmill with a novel body weight support robotic system (Burgess et al. 2010; Capo-Lugo et al. 2012; Patton et al. 2008).

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AUTHOR CONTRIBUTIONS

Author contributions: J.N.L. and D.A.B. conception and design of research; J.N.L. performed experiments; J.N.L. analyzed data; J.N.L. and D.A.B. interpreted results of experiments; J.N.L. prepared figures; J.N.L. drafted manuscript; J.N.L. and D.A.B. edited and revised manuscript; J.N.L. and D.A.B. approved final version of manuscript.

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