Exaggerated interlimb neural coupling following stroke

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The patterns of interlimb coupling were examined in 10 stroke survivors with chronic hand impairment. In particular, the potential roles of postural state and motor tasks in promoting the flexed posture of the upper extremity were assessed. Through the use of electromyography analysis, joint angle measurements and a novel biomechanical apparatus to perturb the digits of the hand into extension, measurements of muscle activity and joint position were compared during multiple postural states, locomotion and voluntary muscle activity. The results demonstrated a significant increase in flexion of the digits (P < 0.001) and elbow (P < 0.005), during walking as compared with standing, sitting or laying supine. These results were indicative of an overall excessive activation coupling between the upper and lower extremities after stroke. Indeed both voluntary finger flexion and voluntary leg extension produced significant activity in the other impaired extremity, leg and arm, respectively, in the stroke as compared with the control subjects. Thus, rectus femoris in the impaired leg was active during finger flexion of the impaired hand in the stroke survivors and all four tested muscles in the impaired arm were active during extension of the legs (P < 0.05). These findings suggest an interlimb coupling related to active motor tasks, contributing to an upper extremity flexion bias following stroke.

Keywords: hemiparesis; stroke; posture; hand; muscle activity

Abbreviations: BIC = biceps; EDC = extensor digitorum communis; EMG = electromyography; FDS = flexor digitorum superficialis; MCP = metacarpophalangeal; PIP = proximal interphalangeal; TRIC = triceps


Introduction

Following stroke, a flexed posture of the upper extremity often develops in people with chronic hemiparesis. Our past studies suggest that this increased flexion is predominantly neurally mediated (Kamper et al., 2003) and as a result, the flexion posture is likely to be affected by the state of the sensorimotor system related to ongoing motor tasks. Specifically, the increased neural drive to the legs needed for functional motor tasks could increase uncontrolled flexion of the upper extremity. In accordance with this view, clinical observations suggest that involuntary flexion of the upper extremity is more pronounced during walking compared with sitting (Olsen, 1997).

Walking, which presents unique challenges to balance not present during static posture, might result in increased drive to anti-gravity muscles through vestibulospinal pathways. The vestibulospinal nuclei are active during locomotion, and the outputs of vestibular nuclei are modulated in phase with stepping in cats (Matsuyama and Drew, 2000). Vestibular input has also been observed to modulate activity in the upper (Britton et al., 1993; Guerraz et al., 2003), as well as lower extremities, even in the more distal muscles of the fingers in infants (Ronqvist, 1995). While the muscles extending the joints of the legs are considered as the anti-gravity muscles in the lower extremities, the muscles flexing the arm and hand are considered as the anti-gravity muscles in the upper extremities. Individuals grasp and pull on external objects, such as a pole on a bus, to help maintain an upright posture. Thus, an increase in the excitability of the vestibulospinal pathways after stroke could contribute to the flexed arm posture during walking.

Another possible mechanism for increased arm flexion during walking after stroke is an enhanced coupling between the upper and lower extremities associated with spinal networks that control walking. During walking, upper extremity
muscles are activated in a rhythmic fashion in synchrony with movement in the lower extremities (Jackson, 1983; Jackson et al., 1983). This coordination seems analogous to that employed in the coordination of all four limbs during quadruped locomotion (for review see Dietz, 2002) or reaching with postural compensation (Schepens and Drew, 2004). Reticulospinal pathways play a fundamental role in this coordination among the limbs (Drew et al., 2004). It is possible that following stroke, the corticoreticular pathways are interrupted (Matsuyama et al., 2004), leading to disinhibition of excitatory reticulospinal pathways and subsequent concurrent activation of muscles in multiple limbs.

Reduction of descending inhibition of reflex circuitry following neural injury, such as stroke, may also increase the influence of walking on activity in upper extremity muscles. For example, reciprocal inhibition between agonist–antagonist muscle pairs has been shown to be reduced after stroke (Baykousheva-Mateva and Mandellev, 1994; Crone et al., 1994; Crone et al., 2003). Reflex coupling between the upper and lower extremities has been shown to increase following cervical spinal cord injury, as demonstrated by recorded upper extremity muscle activity in response to electrical stimulation applied to the legs (Calancie, 1991; Calancie et al., 1996). Furthermore, the startle reflex, an interlimb reflex itself, is altered following neural injuries such as stroke and spinal cord injury (Jankelowitz and Colebatch, 2004a).

This study was conducted in order to quantify the effects of posture and motor task on the impaired upper extremity muscle activity and resulting joint position in individuals with chronic post-stroke hemiparesis. To accomplish this task, joint angles of the upper extremity digits and elbow of the impaired arm were measured along with muscle activity from muscles of both arms and legs. Subjects were instructed to remain as relaxed as possible in the uninvolved limbs while maintaining one of three postures (lying, sitting and standing) or performing one of three tasks (walking, finger flexion and leg extension) while kinematic and electromyographic data were recorded. We postulated that both posture and task would affect involuntary activity of the muscles in the upper extremity, and that interlimb activation would be exaggerated following stroke.

**Material and methods**

**Subjects**

Ten individuals (six men and four women) with chronic unilateral motor deficits subsequent to stroke participated in the study (see Table 1 for a summary of subject characteristics). Subjects (A–J) ranged in age from 41 to 83 (64 ± 12) years and were at least 1 year post-stroke (60 ± 35 months). Subjects were selected for the presence of an involuntary bias towards a flexed finger posture and for hand function characterized by Stages 2–5 on the Stage of Hand section of the clinical Chedoke-McMaster Stroke Assessment (Gowland et al., 1995) by an occupational therapist. This ordinal scale, which runs from 1 to 7 with 1 signifying greatest impairment, classifies hand motor control based on the performance of pre-specified tasks. The subjects for this study all had some finger flexor activity, but none had control of finger individuation. Five neurologically intact subjects (aged 41 ± 16 years) participated in the first part of the study (see Protocol), while 10 neurologically intact subjects (aged 39 ± 13 years) participated in the second experiment examining voluntary activation patterns. All subjects gave informed consent prior to participation. The study protocol was approved by the Institutional Review Boards of Northwestern University and Marquette University and complied with the ethical standards of the 1964 Declaration of Helsinki regarding the treatment of human participants in research.

**Protocol**

**Tasks**

The study consisted of two experiments. The first experiment examined the impact of body posture and activity on involuntary upper extremity posture and muscle activity. Measurements of hand joint angles were recorded with the subject in three different postural states (supine, sitting, standing), and while walking on a treadmill (GaitKeeper, Mobility Research, Tempe, AZ).

For the supine condition, the subject lay flat on his or her back on top of a therapy mat with arms out to the side. The subject was instructed to relax throughout the trial. For the sitting trials, the subject sat upright in a chair with the arms hanging out to the side, away from the chair. Again, the subject was told to relax his or her arms and hands throughout the trial. For the standing condition, the subject stood with eyes open, feet placed roughly shoulder-width apart, and the arms in a relaxed posture to the sides. Thus, the hand was rotated by 90° with respect to gravity for the supine condition as opposed to the other conditions.

For the walking condition, the subjects walked at a self-selected comfortable walking speed on a treadmill for the length of the trial. The mean gait cadence of the post-stroke subject population was 73 (±16) steps per min. The five neurologically intact subjects walked with a cadence of ~96 steps per min. No body weight support was provided; however, each subject was placed into a harness for safety (Litegait, Mobility Research, Tempe, AZ). The unimpaired hand lightly grasped the treadmill console while the subjects were instructed to relax the impaired arm.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Impairment side</th>
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<th>Gait cadence (steps/min)</th>
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Impairment side is defined as the side of the body with primary motor impairment; the glove was worn on this side of the body. Gait cadence denotes the subject-selected stepping frequency on the treadmill.
The order of the conditions was randomly chosen for each subject to remove any potential dependence on task history. In order to ensure that the hand was in a consistent starting position for each test, the subject was asked to flex the hand and then to open the hand as much as possible, immediately after positioning in the supine, sitting, standing and walking conditions. An investigator lightly contacted the impaired forearm during all tasks to ensure that the digits did not contact external objects or other body parts.

Joint angles of the digits of the impaired hand were measured using electrogoniometers (F35, Biometrics, Ladsmith, VA) spanning joints on the dorsal side of the hand. The electrogoniometers were mounted on a glove, placed over the proximal interphalangeal (PIP) joint of the index and ring fingers, the metacarpophalangeal (MCP) joint of the middle and little fingers, and the carpometacarpal (CMC) joint of the thumb to obtain a representation of overall hand posture. Additionally, angle measurements of the elbow joint on the impaired side of the body and the knee of the unimpaired side of the body were also recorded with electrogoniometers. Prior to testing, the electrogoniometers were calibrated against measurements made at multiple joint angles with a manual goniometer. The calibration curve was computed for each electrogoniometer using linear regression.

Muscle activity was recorded using active bipolar surface electromyography (EMG) electrodes (Delsys Inc., Boston, MA). Electrodes were placed over the flexor digitorum superficialis (FDS), extensor digitorum communis (EDC), biceps (BIC) and triceps (TRIC) muscles of the impaired arm, FDS and EDC of the unimpaired arm and the rectus femoris (RF) of each leg.

In order to test the effect of finger stretch and finger joint angle on the flexion bias of the upper extremity during the different tasks, finger extension perturbations were applied to the digits during each trial. A novel pneumatic glove was developed to provide digit extension force through inflation of a single chamber air bladder (Fig. 1). The bladder was sewn onto the palmar side of a nylon/Lycra glove worn by the subject on the impaired hand. The glove itself is also likely to have altered sensory feedback from the hand, but this input was similar for all of the postural and motor tasks in this study (see later). The glove was placed over a wrist splint with palmar support (Futuro, Beiersdorf, Inc., Wilton, CT) intended to maintain wrist posture. Inflation of the air bladder provided a mechanical perturbation to the digits, pushing them further into extension. Servo control of air pressure in the bladder was provided through an electro-pneumatic valve (Pressure Control Valve, QB02005, Proportion-Air, McCordsville, IN), which connected a pressure reservoir (1104360, Jun-Air, Denmark) to the glove. The valve was controlled through a personal computer. A pressure relief valve (Halkey-Roberts, Saint Petersburg, FL) prevented over-inflation.

During each trial, the subject was asked to keep the impaired arm relaxed while the computer control system cycled the pressure in the glove between 0 and 34.5 kPa. Each pressure cycle consisted of 20 s of rest (0–20 s from start of cycle), 20 s of pressure (20–40 s from start of cycle), and 20 s of rest (40–60 s from start of cycle). Five pressure cycles were completed in each of the four test conditions: supine, sitting, standing and walking.

Data acquisition and pressure management were performed using custom software written in Visual Basic.NET (Microsoft, Redmond, WA) to control a PCI board (KPCL-3108, Keithley Instruments Inc., Cleveland, OH). The eight channels of EMG and seven channels of electrogoniometric data were sampled at 1 kHz. One analogue output channel was used to control the pressure set-point of the bladder, allowing computer control of glove inflation.

**Voluntary muscle activation**

A second experiment was performed to explicitly examine upper and lower interlimb activation during a voluntary task. Each subject performed sustained contractions of the upper and lower extremities. Both tests were performed with the subject in a seated posture with back support. First, the subject was instructed to maximally flex the impaired hand into a fist and to maintain activation in this posture for 3 s. The subject then relaxed for 3 s. This process was repeated three times for each trial. The hips and knees were maintained in a posture of roughly 90° of flexion.

Next, the subject performed isometric knee extension. With the hips and knees maintained in 90° of flexion, the subject was instructed to create maximal knee extension torque with both legs simultaneously. The subject was instructed to relax the arms and keep them at her or his sides. For both sets of trials, EMG was recorded at 1 kHz for the same muscles described previously.

**Data analysis**

**Tasks**

Data processing was performed using custom programs developed in Matlab (The MathWorks, Natick, MA) and statistical analysis was completed using SPSS (SPSS Inc, Chicago, IL). The calibrated joint angle data were filtered using a 30-point averaging finite impulse response filter in order to remove electrical noise at 60 Hz. The average angle of each digit was calculated for each of the 5 cycles over a 5 s window when the glove was not inflated (0 kPa) and when the glove was inflated to 34.5 kPa (5 psi). The average angle data when the glove was not inflated was determined from the period 10–15 s after the start of the pressure cycle. The window for average angle data when the glove was inflated was 30–35 s after the start of the pressure cycle. The initial pressure cycle for each condition was discarded because the values were substantially different from the last four pressure cycles.

![Fig. 1 Palmar view of glove. The bladder was sewn into the palmar surface of the glove. Inside the bladder open cell foam was used to maintain air flow when the fingers were in a flexed posture.](image-url)
The joint angles from each of the last four (of five) pulses were averaged for each subject and were used to determine the effects of posture and pressure on joint angle. Data for the dependent variables of finger joint angle of each digit (index, middle, ring, little, thumb) were analysed using a doubly multivariate ANOVA (α < 0.05) with task (4 levels: lying down, sitting, standing and walking) and pressure in the glove (2 levels: 0 and 34.5 kPa) serving as the independent factors for comparing within-subject trends. During ANOVA analysis, values that violated the assumption of sphericity (based on the results of Mauchly’s sphericity test) were adjusted using the Greenhouse–Geisser correction. For the independent variables in which the Wilks’s lambda value showed significance, an additional post hoc analysis, Tukey test, was conducted on each dependent variable to determine statistical significance of the main effects. For the data for elbow angle during the standing and walking conditions were compared using a repeated measure ANOVA (α < 0.05) during the no-pressure condition.

EMG signals were rectified and filtered using a 4th-order 10 Hz low-pass Butterworth filter with zero-phase shift to create envelopes for the activation patterns. These envelopes were then normalized by the peak value obtained for the corresponding muscle during the course of the experiment, either during purported maximum voluntary contraction or during any of the trials. EMG data collected for the impaired arm (EDC, FDS, BIC and TRIC) during each task were averaged over a 5 s window when the glove was not inflated (0 kPa, 10–15 s after the start of a pressure cycle), during inflation (20–25 s after the start of a pressure cycle), and after the glove was inflated (34.5 kPa, 30–35 s after the start of a pressure cycle). The upper extremity EMG data were compared using the four upper extremity muscles as the dependent variables. A multivariate ANOVA (α < 0.05) was run with task (4 levels: lying, sitting, standing, walking) and pressure state (3 levels: no-pressure, inflation and pressure) as the independent within-subject factors. For each of the independent factors, if Wilks’s lambda value showed significance, post hoc Tukey tests were performed to determine statistically distinct levels of the main effects.

Voluntary muscle activation tasks
EMG data from the second experiment were also rectified, filtered and normalized as described previously. For each contraction, EMG was calculated over a one-second window, calculated from 20 to 40 s was 34.5 kPa (with pressure referenced to atmospheric pressure). EMG data from the second experiment were also rectified, filtered using a 4th-order 10 Hz low-pass Butterworth filter with zero-phase shift to create envelopes for the activation patterns. These envelopes were then normalized by the peak value obtained for the corresponding muscle during the course of the experiment, either during purported maximum voluntary contraction or during any of the trials. EMG data collected for the impaired arm (EDC, FDS, BIC and TRIC) during each task were averaged over a 5 s window when the glove was not inflated (0 kPa, 10–15 s after the start of a pressure cycle), during inflation (20–25 s after the start of a pressure cycle), and after the glove was inflated (34.5 kPa, 30–35 s after the start of a pressure cycle). The upper extremity EMG data were compared using the four upper extremity muscles as the dependent variables. A multivariate ANOVA (α < 0.05) was run with task (4 levels: lying, sitting, standing, walking) and pressure state (3 levels: no-pressure, inflation and pressure) as the independent within-subject factors. For each of the independent factors, if Wilks’s lambda value showed significance, post hoc Tukey tests were performed to determine statistically distinct levels of the main effects.

Results
Tasks
The first experiment investigated the effects of body posture and locomotion on finger flexion angles and muscle activity in the impaired arm of 10 hemiplegic subjects. The data for the impaired arm of these subjects was compared with the dominant arm of 5 neurologically intact subjects to ensure that any observed trend was due to factors resulting from hemiplegia rather than as a result of typical changes in motor performance during testing.

Individual finger joint angle data suggested that walking results in a more flexed finger posture than the other tasks (see example data for Subject A in Fig. 2). The multivariate analysis on the finger joints showed that there was a significant difference in joint angle associated with posture (Wilks’s lambda, P < 0.001) across all subjects. Post hoc analysis on each of the digits showed that the increased finger flexion angle was significant across all impaired subjects during walking compared with standing, sitting and supine for the index PIP joint (ANOVA, P < 0.005), middle MCP joint (ANOVA, P < 0.05) and thumb CMC joint (ANOVA, P < 0.05). There was a significant increase in finger flexion during walking compared with sitting and standing for the ring finger PIP joint (ANOVA, P < 0.05) and little finger MCP joint (ANOVA, P < 0.05).

Pressure also had a significant effect on finger posture (Wilks’s lambda, P < 0.003), although the task × pressure interaction term was not significant. The mean difference in angle from standing to walking during the no-pressure condition (mean ± standard deviation) was 25.4 ± 6.1° for the index PIP joint, 12.7 ± 4.1° for the middle MCP joint,
17.2 ± 6.7° for the ring PIP joint, 10.1 ± 3.3° for the little finger MCP joint and 13.7 ± 4.6° for the thumb CMC joint (see Fig. 3). Similar changes were seen during the pressure perturbation. The mean difference in angle from standing to walking during the inflated pressure condition (34.5 kPa) (mean ± standard deviation) was 20.8 ± 5.8° for the index PIP joint, 12.4 ± 4.0° for the middle MCP joint, 13.0 ± 7.4° for the ring PIP joint, 8.6 ± 2.9° for the little finger MCP joint and 11.2 ± 5.5° for the thumb CMC joint. No significant difference in the digit flexion angles during walking compared with the stationary postures for the control subjects for either pressure condition was observed (P > 0.4).

The elbow angle was also affected by the walking task in stroke survivors. Subjects with hemiplegia showed a mean increase of elbow flexion of 28.3° during walking as opposed to standing. This difference was statistically significant (ANOVA, P < 0.005). The control subjects, meanwhile, exhibited an insignificant (P > 0.40) decrease of elbow flexion of 1.6° during walking as opposed to standing (Fig. 4).

Similar results were observed in the normalized EMG data. Activity was typically greater in the walking task across EDC, FDS and BIC of the upper extremity, while TRIC activity was only minimally different between the conditions (see Fig. 5). Across subjects, task had a significant effect on the upper extremity muscle activity (Wilks’s lambda, P < 0.001), as did the pressure condition (Wilks’s lambda, P < 0.05). Comparing walking with the standing condition with no glove pressure, the mean difference of the average EMG activity, expressed as a percentage of MVC over a period of 5 s, was 22 ± 18.3% (ANOVA, P < 0.05) for the EDC muscle, 21 ± 11.4% (ANOVA, P < 0.005) for the FDS, 45 ± 19.2% (ANOVA, P < 0.001) for the BIC, and 21 ± 20.0% (ANOVA, P < 0.05) for the TRIC (Fig. 6A).

Significantly greater activity of the TRIC muscle was also observed when comparing standing and supine tasks (ANOVA, P < 0.05). On the unimpaired side, no difference was observed in FDS activity when walking or standing (P > 0.48), but EDC activity was greater during walking than standing (P < 0.05).

Similar task-dependent effects on EMG were observed during inflation and with the glove fully inflated. A post hoc analysis on the differences in muscle activity showed an increased muscle activity during the dynamic condition (inflation) compared with the activity during the static pressure conditions (0 and 34.5 kPa) for the FDS muscle
ANOVA, $P < 0.05$) during walking, whereas the other recorded upper extremity muscles did not show a significant increase. Although not significant, during walking there was a noticeable increase in activity in the EDC during the stretch, and this activity remained high throughout the duration during which pressure was applied to the glove (20–40 s) (e.g. see Fig. 5).

Control data showed a minimal difference in muscle activity for EDC and BIC during the different tasks (Fig. 6B). The TRI did exhibit a task dependence (ANOVA, $P < 0.05$), while changes in FDS approached statistical significance (ANOVA, $P = 0.07$), with greatest activity during walking. However, over the 5 s interval in all tasks, normalized muscle activity for the controls was never greater than 1% (compared with the 6–10% of MVC that was seen in the stroke survivors) for all arm muscles monitored, suggesting that there was minimal activity during all tasks (Fig. 6B). The increase in FDS and TRI activity during walking was <0.3% of MVC. The BIC and TRIC, especially, activity present in the controls during locomotion did show phasic characteristics corresponding to the gait cycle. Only two hemiplegic subjects (K and M) demonstrated phasic activation in the upper extremity; this activity was similar in phase to that observed in control subjects.

Voluntary muscle activation tasks
The second experiment was designed to determine whether interlimb muscle activation occurs when muscles of the upper or lower extremity are activated voluntarily. Voluntary finger flexion often resulted in activity in other muscles of the arms and legs (Fig. 7A). Typically, activity in all muscles of the upper extremity on the impaired side increased during the finger flexion task.

Comparing the EMG signals between baseline and during finger flexion across all subjects showed a significantly larger EMG in the non-targeted muscles (Wilks’s lambda, $P < 0.01$) for the within-subject factor (finger flexion), and for the between-subject factor (subject group) (Wilks’s lambda, $P < 0.001$). Individual ANOVA results for the muscles showed a significant increase in activity in the hemiplegic subjects compared with control subjects for the BIC ($P < 0.05$) and impaired RF ($P < 0.05$) (see Fig. 8A). No significant difference in activity was seen in the ipsilateral TRIC ($P = 0.7$).

Volitional leg muscle contractions appeared to produce activation of the arm muscles in individuals post-stroke (Fig. 7B). Across subjects, muscle activity was significantly increased for the impaired upper extremity muscles of the stroke survivors in comparison with controls during leg extension (Wilks’s lambda, $P < 0.001$). (The data from one control subject were removed from the analysis for the leg extension task because it was greater than two standard deviations outside the norm, and were thus classified as an outlier.) Individual ANOVA results for each muscle showed significantly higher activity for the EDC ($P < 0.002$), FDS ($P < 0.001$), TRIC ($P < 0.05$), and BIC ($P < 0.001$) during leg extension in the hemiparetic subjects (Fig. 8B).

Fig. 5 EMG data from the impaired arm muscles including EDC, FDS, BIC, and TRIC. This is a representative sample during one pressure pulse for subject J. The bar represents the time at which pressure was applied to the glove.
Discussion
The results from this study demonstrate the role of lower extremity activation in promoting a flexed posture of the upper extremity. There was no change in flexion of the digits during the static tasks (laying supine, sitting and standing). Increased flexion of the impaired upper extremity, however, was present during locomotion and volitional leg extension in a seated posture, as confirmed by both joint angle measurements and electromyographic recordings.

Absence of influence of static posture
The different static tasks appeared to have had little effect on hand posture or upper extremity muscle activity in people with stroke. The hand remained relatively relaxed with absolute joint flexion typically remaining <20°. While some low-level muscle activation was observed, the amount of activity did not change with posture. Presumably, the vestibulospinal drive to the spinal cord varied for these different postures. Indeed, a decrease in the soleus H-reflex during standing as compared with sitting has been shown to occur in neurologically intact subjects (Chan and Kearney, 1982). H-reflex changes are also apparent during static and dynamic whole body tilt (Knikou and Rymer, 2003). While these studies examined only lower extremity muscles, the impact of vestibular input on the upper extremities has also been documented (Britton et al., 1993; Ronnqvist, 1995; Guerraz et al., 2003).

**Fig. 6** Integrated EMG averaged across (A) all stroke survivors and (B) all control subjects. Mean data collected with no inflation (0 Pa) of the bladder for upper arm muscles on the impaired side during the four tasks. Data display the mean value (column) with standard error bars. Asterisks indicate significant difference between groups at $P < 0.05$ (Tukey test) for comparisons where the ANOVA indicated a significant effect of posture ($P < 0.05$).
While it has been proposed that the response to vestibular input following stroke is compromised (Carr and Shepherd, 1987; Bobath, 1990), we observed no effect of body posture on hand muscle activation in this study. These proposed alterations in vestibular drive are also inconsistent with an appropriate adaptation of muscle activity with body tilt during a pedalling paradigm in individuals with chronic stroke (Brown et al., 1997). Recently, galvanic stimulation has been used to directly evaluate the reflex response from vestibular input following stroke. This method shows no change in the reflex amplitude or latency to stimulation between the impaired and unimpaired soleus muscles in people with stroke (Jankelowitz and Colebatch, 2004b). Although further study is warranted, the absence of an effect of a change in static posture on the upper extremity muscle suggests that the vestibulospinal pathway is unlikely to explain an increased hand flexion in people with stroke.

**Effects of dynamic tasks**

In contrast to the influence of the different static postures, the voluntary contractions and walking had significant impact on the posture of untargeted upper extremity joints and activity of upper and lower extremity muscles. In accordance with clinical observations (Olsen, 1997), we found that upper extremity flexion increased during locomotion. Changes in finger and elbow joint angles of >20° were routinely observed when the stroke survivor was instructed to walk rather than stand. The introduction of the pressure perturbation to the digits failed to alter the magnitudes of these changes, thereby suggesting that the greater flexion of the digits was being actively maintained. This interpretation is supported by the significantly greater activity in upper extremity muscles during walking, reflected by the EMG signals in this study. The increased activity in the anti-gravity (flexor) muscles of the upper extremity mirrors the increased activity after stroke in the anti-gravity (extensor) muscles of the lower extremity previously described (for review see Lance, 1980); however, increased activation of finger and elbow extensors were also observed in this study.

The unintended increases in activity of upper extremity muscles during walking were not observed in the control subjects. As expected, the joint angles of the control subjects during walking were similar to those recorded during the static postural tasks. While some elbow flexion was observed, elbow displacement was oscillatory, modulated in phase with the gait cycle. This phasic elbow flexion during locomotion

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**Fig. 7** Rectified and filtered EMG for all muscles recorded during (left) (A) one finger flexion task for subject H and (right) (B) one leg extension task for subject I. (Un RF: unimpaired RF, Un EDC: unimpaired EDC, Un FDS: unimpaired FDS).
has previously been shown to result from muscle activity in the upper limbs, rather than passive pendular movement (Jackson, 1983; Jackson et al., 1983). In accordance with this view, we detected a slight increase in TRIC activity during walking in control subjects. This increase, however, was an order of magnitude smaller than the changes seen in the finger and elbow muscles of the hemiplegic subjects. Additionally, control subjects did not show a significant increase in the EDC, FDS or BIC muscle activity during walking, as occurred in the stroke survivors.

Interaction between upper and lower extremity muscles was observed during voluntary contractions as well. During finger flexion, significant activity was present in not only untargeted muscles in the ipsilateral arm, but also in the ipsilateral leg RF. This coupling was not present in the control subjects. Similarly, leg extension led to a significant increase in muscle activity of all four muscles tested in the impaired upper extremity for stroke survivors, in contrast with control subjects.

A potential mechanism for the increase in flexor activity during locomotion and volitional commands in chronic stroke is an increased drive from the reticulospinal system. The reticulospinal system has been shown in cats and non-human primates to be primarily involved in the regulation of anti-gravity tone, both during locomotion and reaching (Mori, 1987; Werner et al., 1997; Schepens and Drew, 2004). It is possible that following stroke, the input from the corticospinal tract to the reticular formation (Matsuyama et al., 2004) is altered such that the excitatory commands from the pontine reticular formation are disinhibited. Indeed, the BIC response to the reticularly mediated startle reflex has been shown to increase following stroke (Voordecker et al., 1997; Jankelowitz and Colebatch, 2004a).

The presence of a flexion position bias in the upper extremity after stroke also points to a role for the reticulospinal system. Stimulation of the reticular formation in non-human primates has been shown to produce preferential facilitation of arm flexor muscles (Davidson and Buford, 2004). Furthermore, stimulation of reticular nuclei often produces coactivation of agonist–antagonist pairs (Drew and Rossignol, 1990). Excess coactivation of these muscle pairs is common after stroke (Kamper, 2001; Crone et al., 2003). Due to the greater joint torques produced by the flexors as compared with the extensors, this coactivation could produce a net flexion.

The neurological coupling between muscles of the arms and legs during voluntary contractions suggests disinhibition of propriospinal pathways. Propriospinal activity has been shown to be enhanced after stroke, possibly due to enhanced reticuloantispinal influence (Mazevet et al., 2003). Increased propriospinal excitability could also account for the functional grouping of muscle activity into synergy patterns within the arms and legs.

Alternatively, the excessive motor impairment may arise from alterations in direct descending command from the cortex. Interruption of the pyramidal tract in the monkey severely impairs individual movements of the digits (Lawrence and Kuypers, 1968), akin to the situation following stroke (Ada et al., 1996). This suggests that signalling through the direct corticospinal pathway to motoneurons and interneurons involved in voluntary movement is disrupted following stroke; however, this does not fully explain the changes that are seen in evoked reflexes following stroke (Olsen, 1997; Schmit et al., 1999; Kamper, 2000). Cortical damage and subsequent reorganization may contribute to the non-specific coactivation of muscles of the upper and lower extremities, but this process has not been shown directly. The damage to brain tissue resulting from a stroke is likely to have effects on other descending tracts from the brain.

**Clinical implications**

Many measurements of post-stroke motor impairment of the upper extremity have been made; however, these measurements have been limited largely to conditions in which the subject’s legs and torso are in a static posture.
For example, the spastic response to stretch (Levin and Feldman, 1994; Schmit et al., 1999; Kamper, 2000), measurements of muscle synergy patterns (Dewald et al., 1995; Roby-Brami et al., 2003) and kinematics of volitional reach (Velicki et al., 2000; Wu et al., 2000; McCrea and Eng, 2005) have all been used to evaluate sensorimotor dysfunction with people that have had a stroke, while the patient is in a seated posture with the legs nominally at rest. Although these methods provide a description of the impairment severity in static postures, actual functional performance during movement might be quite different due to a change in the sensorimotor coupling of the arms and legs resulting from the neural injury. Additionally, our results suggest that the impact of arm hypertonicity on function may have been underestimated in the past due to its assessment while the subject assumed a relaxed static posture. Significant muscle activity was observed in the arm during either walking or voluntary isometric leg extension. We suggest that the effects of tone on hand function should also be considered while patients are walking, or contracting leg muscles.

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