Muscle activity adapts to anti-gravity posture during pedalling in persons with post-stroke hemiplegia

D. A. Brown, S. A. Kautz and C. A. Dairaghi

Rehabilitation Research and Development Center, V. A. Palo Alto Health Care System, Palo Alto, California, USA

Correspondence to: David A. Brown, Rehabilitation R and D Center (153), V. A. Palo Alto Health Care System, 3801 Miranda Avenue, Palo Alto, CA 94304, USA

Summary

With hemiplegia following stroke, a person's movement response to anti-gravity posture often appears rigid and inflexible, exacerbating the motor dysfunction. A major determinant of pathological movement in anti-gravity postures is the failure to adapt muscle-activity patterns automatically to changes in posture. The aim of the present study was to determine whether the impaired motor performance observed when persons with hemiplegia pedal in a horizontal position is exacerbated at more vertical anti-gravity body orientations. Twelve healthy elderly subjects and 17 subjects with chronic (>6 months) post-stroke hemiplegia participated in the study. Subjects pedalled a modified ergometer at different body orientations (from horizontal to vertical), maintaining the same workload, cadence, and hip and knee kinematics. Pedal reaction forces, and crank and pedal kinematics, were measured and then used to calculate the work done by each leg and their net positive and negative components. The EMG was recorded from four leg muscles (tibialis anterior, medial gastrocnemius, rectus femoris and biceps femoris). The main result from this study was that impaired plegic leg performance, as measured by net negative work done by the plegic leg and abnormal early rectus femoris activity, was exacerbated at the most vertical body orientations. However, contrary to the belief that muscle activity cannot adapt to anti-gravity postures, net positive work increased appropriately and EMG activity in all muscles showed modulated levels of activity similar to those in elderly control subjects. These results support the hypothesis that increased verticality exacerbates the already impaired movement performance. Yet, much of the motor response to verticality was flexible and appropriate, given the mechanics of the task.

Keywords: body orientation; hemiplegia; muscle activity; pedalling

Abbreviations: BF = biceps femoris; IEMG = integrated electromyography; MG = medial gastrocnemius; RF = rectus femoris; TA = tibialis anterior

Introduction

Co-ordinating muscular activity to produce a given movement in a variety of anti-gravity postures (e.g. supine, sitting and standing) is a complex motor control problem that requires appropriate integration of sensory information to adapt to the environment (Young, 1984). When a person with post-stroke hemiplegia moves in an anti-gravity posture the movement appears rigid and inflexible compared with movements in gravity-assisted postures. For this reason, clinicians prefer to begin exercise in gravity-assisted postures and progress persons with hemiplegia to anti-gravity postures as movement control improves. A major determinant of pathological movement in anti-gravity postures is the failure to adapt muscle-activity patterns automatically to changes in posture (Brunnstrom, 1970; Bobath, 1978). This failure to adapt may be due to the pathological release of inappropriate postural responses to anti-gravity body orientations (Bobath, 1978) and/or to an inability to coordinate muscle activity with changing sensory information so as to generate the movement task effectively under the altered environmental conditions (Carr and Shephard, 1987).

Traditional reflex models have been used to explain normal responses to static tilting into anti-gravity body orientations and to explain exaggerated postural responses in persons with cerebral damage. By tilting neurologically normal human subjects from horizontal to vertical and eliciting H-reflex responses in soleus motor neuron pools, Chan and Kearney (1982) (during quiet standing) and Brooke *et al.* (1995) (during stepping movements) have demonstrated decreased soleus motor neuron excitability at increased vertical positions. They proposed (in light of classical animal experiments (*see* Magnus, 1926) that tonic labyrinthine reflexes, elicited by otolith receptors, inhibit lower limb extensor motor neuron excitability during upright stance. Although reflex responses to static anti-gravity postures are demonstrable, they may not contribute to the functional integrity of a movement and may not be present in muscle groups other than the triceps surae. Further, responses to tilting in persons with hemiplegia, especially during a movement task, would demonstrate whether or not a supposedly exaggerated tonic labyrinthine reflex was operating as a pathological impairment to movement in anti-gravity postures.

Contemporary task-oriented models of control suggest that responses to anti-gravity postures are the product of the interaction of many components organized around a fundamental movement goal (Carr and Shephard, 1987). Although neural mechanisms, such as reflex pathways, play a major role in determining muscle-activity patterns, environmental forces and the musculoskeletal mechanics also contribute to the production of motor patterns to achieve cyclical and repetitive bipedal and quadrapedal locomotion (Hoy and Zernicke, 1985; Capaday and Stein, 1987). For example, gravity has the potential to influence the control of limb movements strongly, since it not only affects the sensory input (Young, 1984), but also alters the task mechanics (McMahon, 1984; Davis and Cavanagh, 1993). Therefore, with hemiplegia, degradation of movement capability occurring at anti-gravity postures may be partly the result of an inability to adapt an already impaired muscle-activity pattern to the altered task mechanics. Unless the task mechanics are understood, identifying the impairments associated with the pathological movement is impossible.

Pedalling provides a controlled locomotor movement in which the task mechanics are well understood (Gregor *et al.*, 1991). Therefore, the relative contribution of a post-stroke neurological impairment to the movement deficit observed during anti-gravity postures can be assessed by observing pedalling behaviour. In a pedalling protocol, the task mechanics can be maintained while body orientation with respect to gravity is altered because potentially confounding postural requirements can be eliminated (Brooke *et al.*, 1995) and the anti-phased movement of the two limbs which are coupled by a crank can minimize the net contribution of gravity (Brown *et al.*, 1996a). Since challenges to balance can be minimized by stabilization of the trunk and upper body, the effects of altered orientation on the control of lower limb, reciprocal, propulsive movements can be isolated. Because pedalling is functional, safe, and accessible to patients with a wide range of ambulatory function, the bicycle ergometer has been used to study bilateral movement patterns in several patient populations, including stroke (Brown and DeBacher, 1987; Rosecrance and Giuliani, 1991; Boorman *et al.*, 1992). For example, Benecke *et al.* (1983) found significant relationships between disordered leg muscle activation patterns during pedalling and indices of spasticity. More recent work has established direct functional relationships between the degree of abnormal muscle activation and poor pedalling performance (Brown *et al.*, 1996b).

By studying pedalling and modelling the task mechanics, our previous work with neurologically normal subjects showed systematically changed joint torque and muscle activation patterns as body orientation was altered (Brown *et al.*, 1996a). Changes in muscle activation patterns were consistent with a neural control strategy that integrates sensory information and produces steady-state pedalling trajectories consistent with some internal model of the movement.

Mechanical measures of pedalling performance can be used to characterize impairment in persons with hemiplegia. In order to pedal at a given cadence and workload the combined mechanical work done by the two limbs must be sufficient to overcome the resistive load. The total mechanical work done by the plegic leg (total work) represents the net contribution of the plegic limb and can be positive (the limb assists crank propulsion), negative (the limb resists crank propulsion) or zero. Since the total work provides a net measure for a cyclic movement, and pedalling typically includes periods of assistance and resistance (Gregor *et al.*, 1991), it is useful to assess the assistance and resistance provided by the leg independently. The net positive work done by the plegic leg represents the component of the mechanical work that propels the crank against the load and is typically dominated by muscular contributions in neurologically normal subjects (Kautz and Hull, 1993). The net negative work done by the plegic leg represents the component of the mechanical work that resists crank propulsion. Negative work typically occurs in the upstroke for neurologically normal subjects and represents the combined effect of muscular activity with gravity and inertial forces (Kautz and Hull, 1993). In the research reported here, we calculated the net work, the positive work and the negative work done by the plegic limb in order to quantify and compare changes in pedalling performance at different body orientations. In addition, EMG activity was measured in order to understand how individual muscle groups respond to altered body orientation, especially during inappropriate periods of the crank cycle.

The aim of the present study was to determine whether the impaired motor performance observed when persons with hemiplegia pedal in a horizontal position is exacerbated at more vertical anti-gravity body orientations. If further

impaired, are the changes in performance consistent with the release of tonic labyrinthine reflexes as a causative factor, or can the results be explained as an appropriate, yet limited, effort to deal with altered task mechanics? We compared the motor performances of a control population consisting of healthy, elderly subjects with those from persons with hemiplegia. It is generally believed that performance is degraded and muscle-activity patterns do not respond appropriately to changes in anti-gravity body orientations. Thus, we hypothesized that, in persons with hemiplegia, total work done by the plegic leg relative to the nonplegic leg would decrease because net positive work by the plegic leg would decrease and net negative work by the plegic leg would increase. Also, it was expected that inappropriate regions of EMG activity would show increased activity, contributing to performance degradation.

Methods

Experiments were performed using 12 healthy elderly subjects and 17 subjects with chronic (>6 months) post-stroke hemiplegia. The healthy elderly subjects showed no signs or symptoms of neurological disease or lower limb orthopaedic impairment. Subjects with hemiplegia were recruited from the surrounding community and were selected if they (i) had sustained a single, unilateral cerebral vascular accident with residual lower limb plegia, (ii) could tolerate sitting on a bicycle seat for ~1 h, and (iii) had no severe perceptual or cognitive deficits, no severe sensory deficits, no lower limb contractures, and no cardiovascular impairments that could prohibit pedalling activity. In addition, we selected individuals whose onset of stroke was >6 months before the study, since we wished to exclude subjects who were still undergoing significant neural recovery. All subjects gave informed consent, and the study was approved by Stanford University School of Medicine Ethics Committee. A summary of the clinical status of the subjects appears in Table 1.

All subjects with hemiplegia underwent the lower limb portion of the Fugl-Meyer Assessment (Fugl-Meyer *et al*., 1975) for assessment of global motor function. This assessment tool has been shown to be valid and reliable for this population (Duncan *et al*., 1983, 1992). Patient scores appear in Table 2 which also shows the scores obtained in selected subsections of the test. The reflex activity score shows that all subjects had some reflex activity present in the plegic limb and that more than half of the subjects (nine out of 17) had hyperactive responses in at least one muscle group. The synergy control sub-score shows that four subjects were unable to move out of abnormal synergy patterns, another four were able to combine abnormal flexor and extensor synergy patterns, and the remaining nine subjects demonstrated some isolated movements out of synergy. Finally, most subjects (12 out of 17) were unable to balance on the plegic leg. Although four of the subjects attained a 100% score, all subjects exhibited difficulty in walking.

Table 1 *Subject population characteristics*

Subject	Age (years)	Sex	Time since stroke (months)	Plegic side
M.I.	68	M	38	L
M.M.	61	F	18	R
M.U.	60	F	10	L
M.V.	65	F	56	L
N.C.	65	F	29	R
N.H.	57	F	7	L
N.M.	64	M	34	R
P.O.	60	M	19	R
P.P.	55	M	14	R
P.Q.	63	M	8	R
P.U.	74	M	23	L
P.V.	61	M	21	L
Q.A.	73	M	15	R
Q.G.	60	M	12	R
K.G.	61	M	10	L
K.V.	61	M	15	R
K.O.	65	F	12	L
Hemiplegic group (n = 17)	63.1 ± 5.0	11M, 6F	20.05 ± 5.2	8L, 9R
Healthy elderly group (n = 12)	72.9 ± 5.2	10M, 2F	–	–

Therefore, we feel the subjects represented a typical sub-population of rehabilitation candidates.

Pedalling set up

A standard ergometer with a frictionally loaded flywheel was modified so that body orientation with respect to gravity could be altered in a controlled manner (Fig. 1). Details of this set up are described elsewhere (Brown *et al*., 1996a, b). Reaction forces oriented normal and fore-aft to the pedal surfaces were measured using instrumented pedals (Newmiller *et al*., 1988). Feet were attached firmly on the pedal surfaces. Angular rotation of the crank and pedals were measured using optical encoders. The angular rotation and pedal force measures were used to calculate net torque about the crank axis (crank torque).

Surface EMGs were recorded from the tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF) and biceps femoris (BF) muscles of the right leg in healthy subjects, and of both legs in subjects with hemiplegia. EMG electrodes (Therapeutics Unlimited, Iowa City, Ia., USA) were positioned over the distal half of the muscle belly such that contact surfaces were aligned longitudinally to muscle fibres. Electrode sites were prepared by cleaning the skin with isopropyl alcohol and, when necessary, shaving the hair to insure good contact. Ag–AgCl electrodes (diameter 8 mm, interelectrode distance 22 mm) were attached using adhesive pads and electrode gel. The first stage preamplifiers provided a gain of ×35. A common reference electrode was placed on the distal end of the right tibia. Amplifier gain was selectable from ×500 to ×10 000 with a bandwidth of

Table 2 *Modified Fugl-Meyer scores broken down into selected subsections*

Subject	Reflex activity (max. = 6)	Synergy control (max. = 22)	Balance (max. = 10)	Total modified Fugl-Meyer Score (%)
M.I.	4	13	6	79
M.M.	6	22	9	98
M.U.	6	22	10	100
M.V.	3	20	8	94
N.C.	6	22	10	100
N.H.	3	7	6	83
N.M.	6	22	10	100
P.O.	2	17	8	81
P.P.	4	17	8	82
P.Q.	5	22	7	95
P.U.	4	18	6	88
P.V.	6	22	10	100
Q.A.	4	22	6	92
Q.G.	4	12	6	70
K.G.	6	17	6	89
K.V.	5	22	8	96
K.O.	4	10	5	59
Distibutions of hemiplegics' scores ($n = 17$)				
	0–1, $n = 0$	0–14, $n = 4$	0–4, $n = 0$	70–79, $n = 3$
	2–4, $n = 9$	15–18, $n = 4$	5–8, $n = 12$	80–89, $n = 5$
	5–6, $n = 8$	19–22, $n = 9$	9–10, $n = 5$	90–100, $n = 9$

The reflex activity score reflects either the absence of activity (0–1), or hyperactivity (2–4) or normal activity (5–6) during reflexes; the synergy control score reflects the ability to move within (0–14), to combine (15–18), or to move out of (19–22) primitive synergy patterns; the balance score reflects the ability to sit without support (0–4), or to stand with (5–8) or without (9–10) support. In general, higher scores reflect greater abilities.

20–4000 Hz. The common mode rejection ratio was 87 dB at 60 HZ and the input impedance was >15 MΩ at 100 Hz.

The experimental protocol, conducted in one 2-h period, consisted of measurement of pedal forces, pedal and crank kinematics, and EMG during pedalling at nine randomly ordered body orientations. Each subject pedalled against an applied friction of 15 N (workload of ~80 J) at each orientation (random presentations of 0°, 10°, 20°, 30°, 40°, 50°, 60°, 70° and 80° of tilt with respect to horizontal). Subjects were instructed to maintain a 50 r.p.m. cadence by following a metronome. Once a steady cadence was achieved, 15 s of EMG, pedal force, and encoder data were collected and stored at 1000 samples/s.

Since clear bursts of muscle activity are difficult to identify in subjects with hemiplegia, muscle-activity patterns were characterized in terms of activity present during four equal phases (90° duration) in the pedalling cycle (Fig. 1). Since 0° of the crank cycle was referenced relative to the seat tube, and hence the body axis, the four phases reflect the relationship between the movement of the crank and the body. Phase I is when the crank is moving away from the hip and anterior to the body axis; phase II is away and posterior; phase III is towards the body and posterior, and phase IV is towards the body and anterior. Muscle activity occurring during phases I and II is considered 'extensor' in nature (active during limb extension), while activity during phases III and IV is considered 'flexor' (active during limb flexion). Also, activity during phases IV and I or II and III is termed, respectively, 'top transition' and 'bottom transition'

activity because the muscles are active during the switch from either limb flexion to extension or vice versa.

EMG patterns were quantified to determine both absolute and relative activity during each phase. First, the signals were rectified and then integrated (IEMG) over each of the four phases. The absolute IEMG in each phase was used to quantify the changes in the magnitude of activity during each phase as the body orientation changed. Also, EMG was quantified for each muscle by expressing the IEMG of each phase as a percent of the total IEMG from the entire cycle. This allowed periods of relative activity and inactivity to be identified within the cycle and differences noted between the hemiplegic and the control population.

Data analysis

Mechanical work measures were calculated from the kinematic and kinetic data. Total work was calculated by integrating both (left and right) crank torques over the pedalling cycle. Single leg work was calculated by integrating only a single (left or right) crank torque over the cycle. Net positive work and net negative work were calculated by integrating all positive areas and negative areas of the single leg crank torque curve, respectively.

Individual, non-averaged crank kinematics, kinetics and EMG activity were examined for general trends. The kinetics were then characterized (total work, single leg work, net positive and net negative work by a leg) by calculating all measures for each revolution and averaging to get the mean
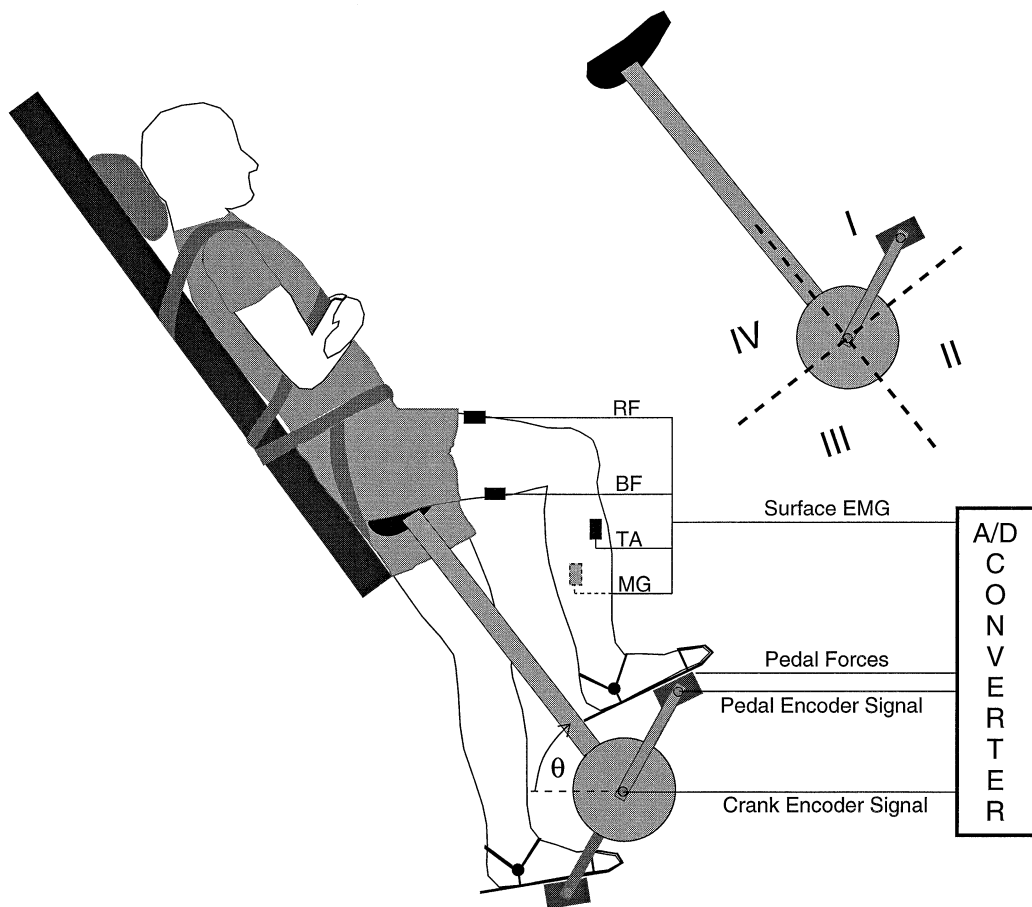
Fig. 1 Experimental set up showing the ergometer, subject and variables recorded. The tilt angle of the backboard (θ) which is used to define body orientation with respect to gravity (horizontal corresponds to 0° and vertical to 90°). The strapping system stabilized the subject so that a consistent body configuration was maintained throughout the experiment. Inset shows orientation of the four phases of the pedalling cycle that were used to analyse EMG activity (*see* text for further explanation).

values for each trial. A repeated-measures ANOVA model was used to identify overall differences at each of the nine body orientations while accounting for differences between subjects. Regression analyses were performed to test for associations between work done by the plegic leg and mean values of crank forces and relative EMG phasic distributions. Correlation coefficients were used to express the linear strength of association between dependent variables and work done by the plegic leg for the group means (Kleinbaum *et al.*, 1987).

Results

Mechanical work done by the leg

Impaired pedalling in subjects with hemiplegia was characterized by decreased assistive and increased resistive force generation in the plegic limb. Compared with control subjects, single leg crank torque trajectory was characterized by more net negative work (Fig. 2) and less net positive work (Fig. 3) in the plegic leg. Subjects with

hemiplegia performed less total work with the plegic leg than control subjects did with the non-preferred leg (defined as the leg generating <50% of the total work, namely 2.6 ± 31.9% versus 45.5 ± 4.1% of total work done by both legs, $P < 0.0001$) and produced a very large range of work values (41.0 − 37.8% of the total work done by both legs, $n = 17$).

For both study populations, assistive and resistive force generation increased with more vertical body orientations. In control subjects, net negative work became more negative as body orientation increased, becoming most negative at 50° of body orientation, and then leveling off at 60°, 70° and 80° (Fig. 2). Similarly, net positive work systematically increased until 50°, and then leveled off at 60°, 70° and 80° (Fig. 3). However, in subjects with hemiplegia, neither net negative nor net positive work values peaked at 50° but, rather, they continued to increase up to 80° (from −25.0 ± 24.4% to –39.2 ± 27.5% for net negative work; from 33.2 ± 19.8% to 43.7 ± 16.4% for net positive work) (Figs 2 and 3). As a consequence of the parallel increases in net
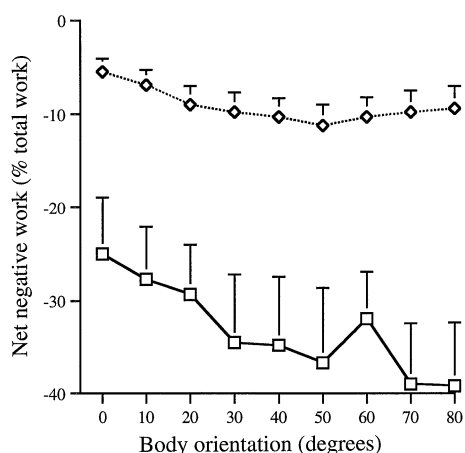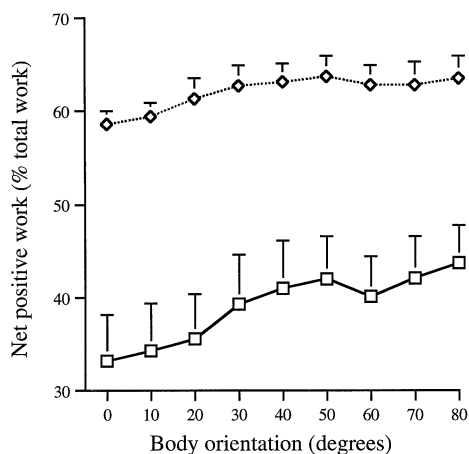
Fig. 2 Mean net negative mechanical work done by a single leg at each of the tested body orientations (0–80°) for all control subjects (*n* = 12; open diamonds) and the plegic leg of subjects with hemiplegia (*n* = 17; open squares). Values are expressed as percentages of the total work done by both legs in each subject. Bars represent the SE of the mean. Control subjects show less-negative values that peak at 50° of body orientation and then level off, while subjects with hemiplegia show more-negative values that continue to decrease to 80° of body orientation.

Fig. 3 Mean net positive mechanical work done by a single leg at each of the tested body orientations (0–80°) for all control subjects (*n* = 12; open diamonds) and the plegic leg of subjects with hemiplegia (*n* = 17; open squares). Values are expressed as percentages of the total work done by both legs from each subject. Bars represent the SEM. Control subjects show greater values that peak at 50° of body orientation and then level off, while subjects with hemiplegia show lesser values that continue to increase to 80° of body orientation.

Fig. 4 Correlation plot between single leg work done by the plegic leg and synergy score from the Modified Fugl-Meyer assessment. Group I represents those subjects (*n* = 4) who were only able to move within extensor/flexor synergy patterns. Group II represents those subjects (*n* = 4) who were only able to move with combined patterns of mass flexion and extension. Group III (*n* = 9) represents those subjects who were able to move out of synergy patterns. Note that all Group I subjects generated negative single leg work values and that all but one Group III subject generated positive single leg work values.

to complete the pedalling task at the 70° and 80° body orientations. Further, an even greater correlation was found for the synergy scores of the Fugl-Meyer and plegic leg work ($P = 0.0001$, $r^2 = 0.854$) (Fig. 4), indicating that the ability to move out of abnormal synergy patterns is a strong predictor of pedalling performance. As shown in Fig. 4, all subjects (*n* = 4) unable to combine, or move out, of synergy patterns (score 0–14) generated negative values of work done by the plegic leg. Also, with the exception of Subject Q.A., all subjects (*n* = 9) able to move out of synergy (score 19–22) generated positive values of work done by the plegic leg. Side of lesion did not appear to affect plegic leg performance ($-9.7 \pm 39.5\%$ for left sided plegia versus $11.8 \pm -26.3\%$ for right-sided plegia, $P = 0.4739$).

EMG activity

Increased resistive force generation was consistent with the EMG patterns of subjects with hemiplegia. On average, at 0° of tilt, subjects with hemiplegia exhibited differences in the percentage of total IEMG (%IEMG) that occurred in at least one phase of the pedalling cycle in all four muscles investigated when compared with healthy, elderly control subjects (Fig. 5). The RF and BF muscles showed changes in %IEMG (when compared with those of control subjects) that suggest changes in the temporal aspect of the EMG. As a group, the RF muscle in subjects with hemiplegia showed no clear absence of activity in any of the phases (activity in all phases exceeded 19.3%). RF activity in phase III was

negative and net positive work (and similar to control subjects), subjects with hemiplegia pedalled with no net change in total single leg work done over the range of body orientations ($P = 0.8619$).

Subjects with the lowest levels of motor recovery generated both the least assistive force and most resistive force. Work done by all plegic legs was correlated with motor recovery as measured by each subject's Fugl-Meyer score for the lower extremities ($P = 0.0001$, $r^2 = 0.775$). In fact, the subject with the lowest Fugl-Meyer score (K.O.) was unable
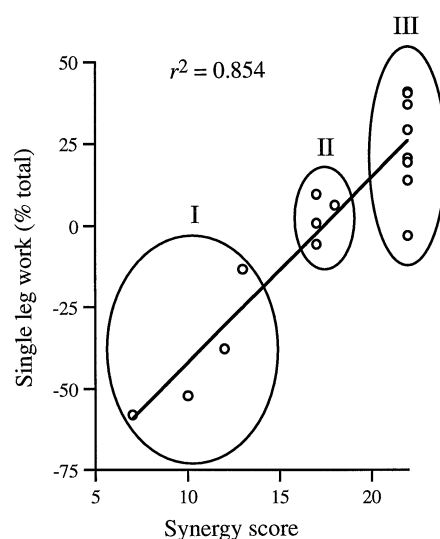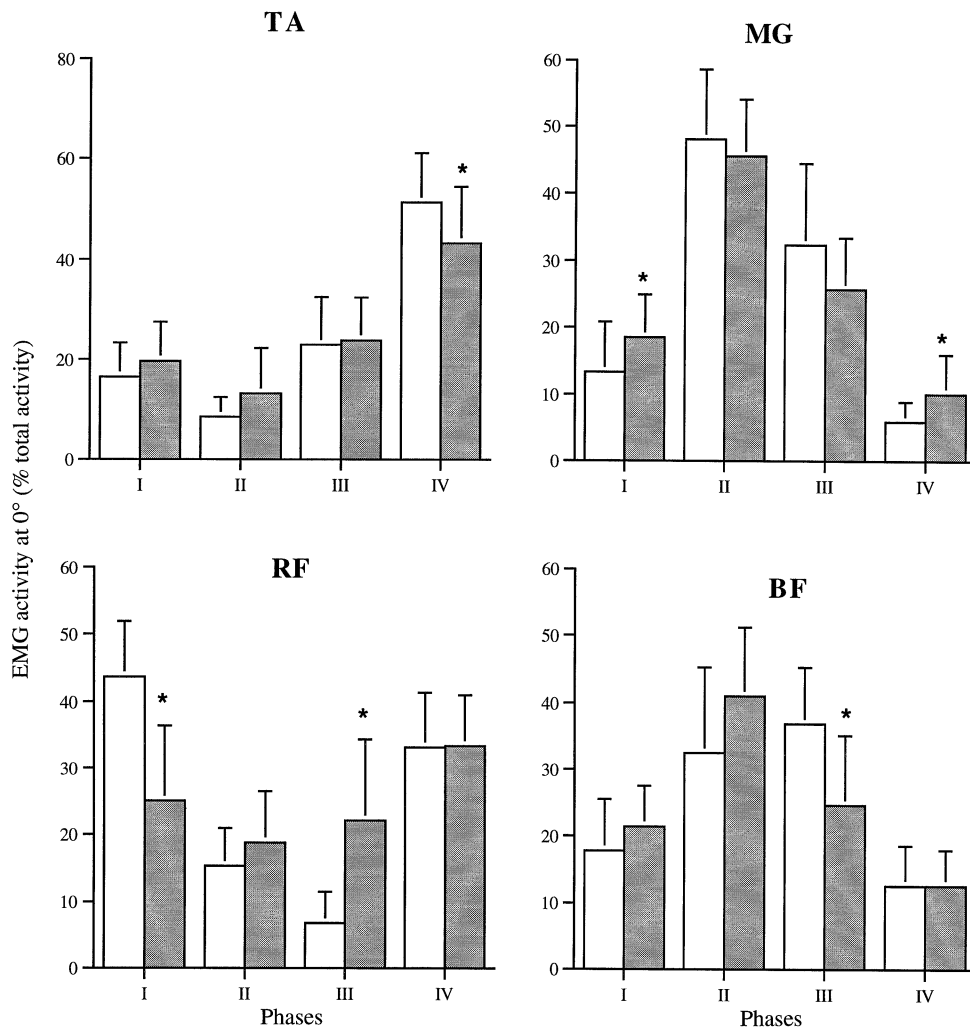
Fig. 5 IEMG activity in each of four phases of the pedalling cycle. Values are expressed as percentages of the overall integrated EMG activity for each muscle throughout the cycle for the control subjects ($n = 12$; open bars) and the plegic leg of the subjects with hemiplegia ($n = 17$; filled bars) when the body is at 0° tilt. Asterisks represent phases where the mean plegic leg activity is different ($P < 0.05$ level) using Student's two-tailed t test. Bars represent the SEM. Note the large increase in RF activity and large decrease in BF activity during phase III for the plegic leg.

strongly negatively correlated with work done by the plegic leg ($P = 0.0001$, $r^2 = 0.83$) thus implying some relationship between the two measures. In fact, RF activity in phase III was also strongly correlated with negative work done by the plegic leg, so that the greater the activity in this phase, the greater the negative work done by the plegic leg (Fig. 6). further, the RF muscle activity, which was normally very low during phase III (i.e. the muscle is usually off), showed greatly increased activity during this phase when compared with control subjects ($P = 0.0001$). With the BF muscle, the relative percent of phase III activity was reduced so that phase II activity was dominant. Therefore, in general, the plegic leg had a heightened amount of mistimed RF flexor activity, and when combined with abnormally low BF flexor muscle activity in phase III, this perhaps contributed to the exaggerated increase in negative work done during the upstroke phase of the pedalling cycle.

For the TA and MG muscles, similar levels of %IEMG occurred in all phases for subjects with hemiplegia and control subjects. The TA was mainly active as a flexor muscle (phases III and IV), and the MG was mainly active as a bottom transition muscle (phases II and III). The minor differences between plegic and control TA and MG activity were apparently the result of more diffuse plegic activity so that 'off' periods were less well defined.

In general, total muscle activity throughout the cycle changed as a consequence of verticality indicating some degree of tonic influence over motor neuron excitability (Fig. 7). In control subjects, there was an increase in the total TA muscle activity and decrease in the total MG muscle activity with verticality ($P = 0.001$). Although there was no statistically significant increase in plegic leg TA activity in response to verticality, five subjects with hemiplegia showed over-responsive increases, and four subjects showed unduly
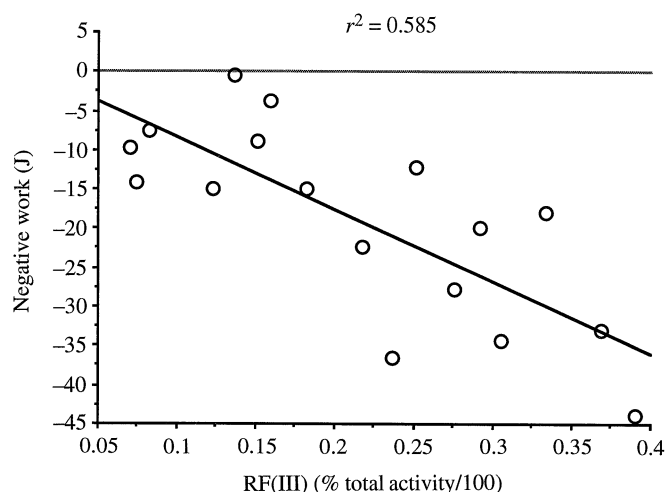
$r^2 = 0.585$

Fig. 6 Correlation between negative work done by the plegic leg and percent of total RF activity occurring in phase III. Since phase III is normally a silent period in the cycle, greater activity in this phase is strongly associated with increased resistive work during the upstroke.
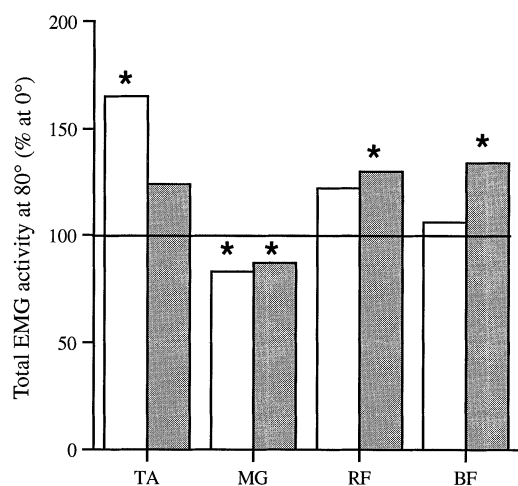

Fig. 7 Bar graph representation of the amount of total integrated EMG activity over an average cycle in control subjects (open bars) versus plegic muscles in subjects with hemiplegia (filled bars) at the most vertical tilt angle (referenced to the most horizontal orientation). Stars represent activity levels that significantly exceed ($P > 0.05$) the level of activity at 0° of tilt (horizontal). Note that, except for the decreased MG activity, plegic activity increased with verticality in the RF and BF muscles while control leg activity increased with verticality in the TA muscle.

large decreases (compared with control subjects) in TA activity during phases IV and I (the predominant phases of TA activity) (Fig. 8). Also in the plegic legs, MG muscle activity decreased ($P = 0.001$) while there was an overall increase in the RF and BF muscles ($P = 0.001$) which was not present in controls.

The inappropriate activity in the RF muscle in the plegic limb characterized above at 0° was exacerbated by increased verticality. The increased RF IEMG activity in phase III, on average, was different at 70° and 80° of body orientation

when compared with 0° ($P = 0.0001$) (Fig. 9). Most subjects (12 out of 16) showed some level of increase in phase III activity with the RF muscle at the most vertical body orientation, when compared with control subjects (Fig. 10).

Figure 11 shows the changes in absolute IEMG amplitude that occurred in each muscle during its predominant two phases (as defined by control subject activity) of the pedalling cycle. With the notable exception of the BF muscle, all muscles demonstrated changes in EMG activity as a result of tilting that were similar to those in the control subjects. Whereas the average absolute IEMG of BF remained relatively unchanged with body orientation in control subjects, it increased during the bottom transition phase at the 70° and 80° body orientations in the plegic limb of subjects with hemiplegia. Figure 12 demonstrates, in one representative subject (N.M.), how plegic leg activity tended to increase in the TA, RF and BF muscles, and decrease in the MG muscles, when tilted from a horizontal to a vertical orientation. The responses in the plegic subject can be seen to parallel the responses, on average, in control subjects.

EMG activity in the nonplegic leg

Although, on average, the nonplegic leg demonstrated EMG patterns and responses to tilting that were similar to the leg of control subjects, the nonplegic RF muscle activity showed compensatory alterations in its timing. Percentage of RF muscle activity in the nonplegic leg was dominant in phases I ($50.1 \pm 10.2\%$) and II ($21.3 \pm 8.5\%$), the extension phases, rather than phases IV ($33.5 \pm 8.1\%$) and I ($44.9 \pm 12.4\%$), the top transition phases, as in control subjects. This phasic shift in activity is probably the result of increased demand for extensor activity required to counteract the increased negative work produced by the plegic leg during the upstroke. Similar to control subjects, subjects with hemiplegia, on average, exhibited nonplegic TA and RF activity that increased, MG activity that decreased and BF activity that did not change at the most vertical body orientations. Therefore, the nonplegic muscles performed very much like control muscles except that they were forced to generate more positive work to compensate for the decreased work done by the plegic leg.

Discussion

The main finding from this study was that plegic leg performance, as measured by net negative work done by the plegic leg and poorly timed RF activity, was exacerbated at the most vertical body orientations. However, net positive work increased appropriately and EMG activity in most muscles showed modulated levels of activity, contrary to the belief that muscle activity cannot adapt to anti-gravity postures. Therefore, it appears that even though the patterns of muscle activity are abnormal in hemiplegia, and even though these abnormal patterns tend to degrade with verticality, the ability to modulate agonist and antagonist
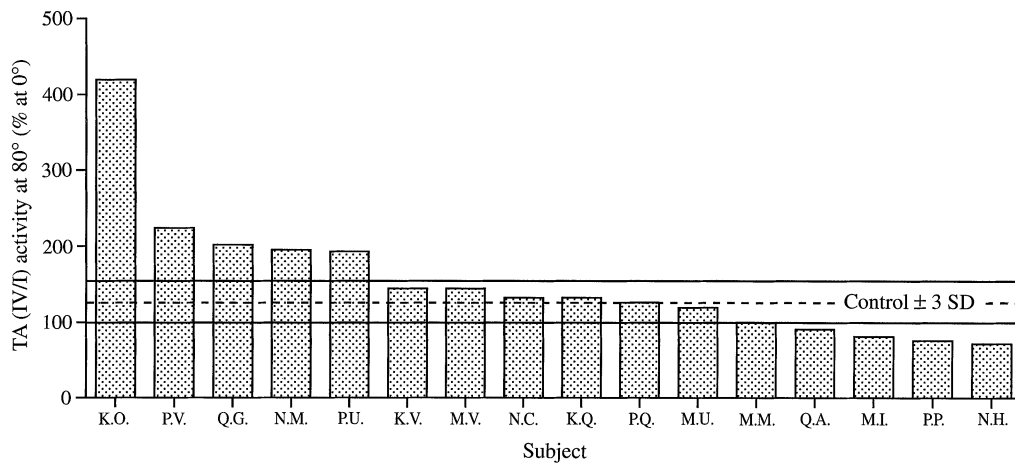
Fig. 8 Bar graph representation of the relative increase in TA integrated EMG activity during its predominantly active phases (IV and I) in subjects with hemiplegia as they were tilted from a horizontal to the most vertical body orientation. Included is the mean value for the control subject population and lines representing ±3 SD. Note that five subjects (K.O., P.V., Q.G., N.M. and P.U.) showed over-responsive increases to verticality, while four subjects showed significant decreases as a result of verticality.
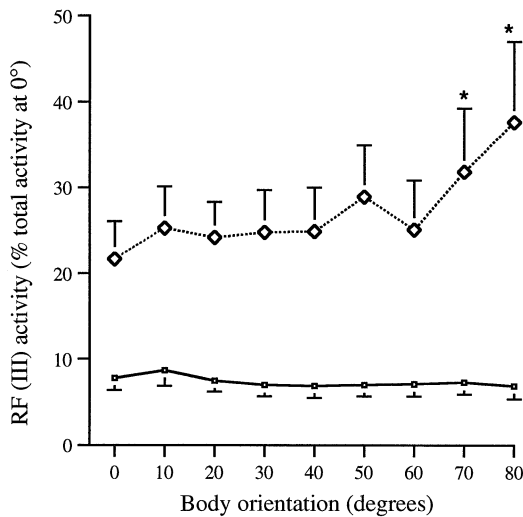

Fig. 9 Mean RF activity during phase III for all control subjects (open squares) and all subjects with hemiplegia (open diamonds) at each body orientation. Values represent total RF IEMG for phase III as percentages of overall RF IEMG during pedalling at 0° of body orientation for each subject. Bars represent the SEM. Increases in activity were observed at the 70° and 80° body orientations for the plegic leg, but not for the control leg. *$P < 0.05$.

muscle activity in response to a gravitational stimulus remained intact.

Although some net negative work is a normal, expected component of pedalling performance (Redfield and Hull, 1986), large amounts of resistive force may interfere with smooth and efficient pedalling. Control subjects do a small amount of negative work during the upstroke phase of pedalling, mostly due to the weight of the leg opposing the upward rotation of the crank. In this study, control subjects

demonstrated a leveling off of the net negative work at 50° presumably because the negative contribution of gravity peaks at that body orientation. Therefore, in the hemiplegic population, the continued decrease in net negative work after 60° of tilt suggests that impaired muscle activity, rather than gravity, has produced the negative functional effect. Furthermore, increased net extensor muscle activity must be contributing to the observed over-generation of net negative work since the majority of net negative work is generated during upstroke (Redfield and Hull, 1986), when the limb is flexing.

The abnormal RF activity in phase III is strongly correlated with net negative work as would be expected given the mechanical contributions of the RF. This activity was exacerbated at the most vertical body orientations. Even though RF activity during phase III is occurring as the hip is flexing, the RF has been shown to be lengthening during this period due to the knee concurrently flexing and, hence, it would contribute negative mechanical work (Hull and Hawkins, 1990). However, the RF is not generally considered a major power-producing muscle during steady-state pedalling so it is likely that the major power-producing muscles, not measured in this study, also contributed to the increased level of negative work (e.g. prolonged vastus group activity).

The result that MG activity is reduced at more vertical body orientations is similar to that shown in H-reflex studies of static tilting in humans, where soleus motor neuron excitability was found to decrease with more vertical body orientations (Chan and Kearney, 1982; Brooke et al., 1995). In the current study, decreased MG activity was observed both in control subjects and those with hemiplegia which suggests that effects of otolith stimulation (via static tilting) on lower motor neuron excitability may still be operating
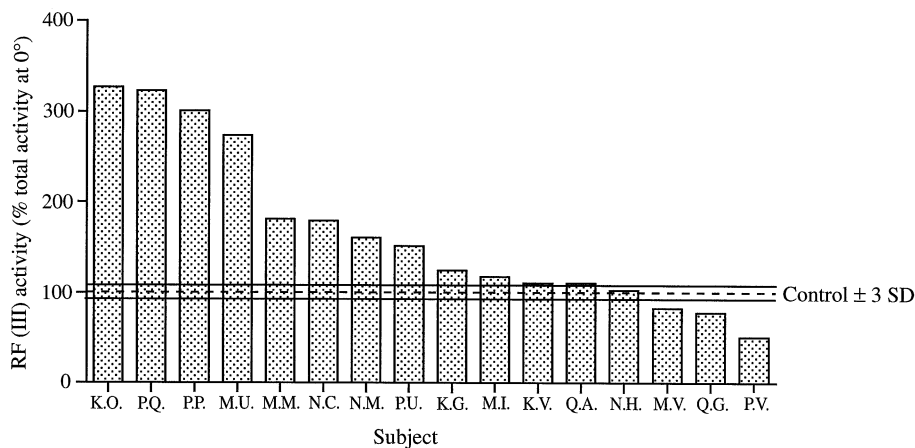
Fig. 10 Mean change in RF activity during phase III for each individual subject with hemiplegia at the most vertical (either 70° or 80°) body orientation when compared with its value at horizontal orientation. Values represent total RF IEMG for phase III at 70° or 80° as percentages of phase III RF IEMG occurring at the 0° body orientation. Twelve out of 16 subjects demonstrated some increase in activity at the most vertical body orientation when compared with mean ±3 SD in control subjects. One subject was unable to obtain either the 70° or 80° body orientation.

after cerebral damage, even during movement tasks. Although responses to static tilting in TA muscles have not been studied because of the inherent difficulties associated with eliciting an H-reflex in this muscle group (Teasdall *et al.*, 1951), the strong reciprocal activation pathways between the TA and MG motor neuron pools (Tanaka, 1974) suggest that the increase in TA activity at more vertical body orientations should be expected. Although the strong increase in TA activity with verticality observed in control subjects is expected and has been reported in an earlier paper (Brown *et al.*, 1996a), some subjects with hemiplegia (Fig. 8) showed greater than expected increases that are consistent with observations of released inhibition, by higher centres, of TA motor-neuron excitability in this population (Yanagisawa *et al.*, 1976). Yet other subjects with hemiplegia showed a greater than expected decrease in TA activity, possibly as a result of an inability to recruit motor units in response to a gravitational stimulus. Similar to the TA muscle, the RF muscle activity, also present during limb flexion, is increased with verticality, although the increase occurs during a period in the pedalling cycle which is considered pathological (contributing to negative work). However, the BF muscle, shifting its timing of activity more toward the extension phases of pedalling, indicates an increase in extensor bias with increased verticality.

During vertical pedalling, with the exception of the MG muscle, a general picture emerges of enhanced extensor activity during the downstroke phase of pedalling, and enhanced flexor activity during the upstroke phase of pedalling. Indeed, the results from this study show an overall increase in muscle activity results from verticality in persons with hemiplegia (*see* Fig. 7). This non-biased increase in activity is contrary to the extensor bias that is expected with lesions of the corticospinal tract (Lawrence and Kuypers, 1968a, b). The expression of flexor bias may be unique to

our experimental set up since bilateral pedalling allows flexion movements to occur even though flexor muscle activity may be low and overwhelmed by extensor activity.

Since subjects with vestibular response deficits were excluded from this study, we may suppose that the vestibulospinal system remained intact, thus allowing tonic influences to remain operational. Otolith contributions to postural control in humans are difficult to demonstrate but, in one study, has been shown to have strong gravito-inertial influences on lower extremity motor-neuron pools (Fries *et al.*, 1993). These influences appear to increase the excitability of anti-gravity muscles with changing posture in a systematic manner. Also, the vestibulospinal tract has been shown to exert a strong facilitatory influence over flexor reflex afferent pathways (Bruggencate *et al.*, 1969) that may be participating in the reciprocal flexor/extensor muscle activation during pedalling (Brown and Kukulka, 1993). Flexor reflex afferent pathways are known to be released from inhibition in persons with spasticity (Meinck *et al.*, 1985). This release, in addition to an increased afferent flow that can accompany upright posture (e.g. increased seat and foot pressure) can contribute to some of the exaggerated responses in proximal musculature that occurred with increased verticality.

Contrary to expectations, the hemiplegic population in this study were able to increase net positive work output and systematically change much of the abnormal muscle activity in response to increased verticality. Control subjects maintained the percentage of work done by the dominant leg at all body orientations by systematically increasing the positive work done as the negative contribution from gravity similarly increased. Subjects with hemiplegia also maintained the percent work done by the plegic limb. However, they also had an exaggerated concomitant increase in both positive and negative work at the most vertical body orientations. The plegic leg was already performing a reduced level of
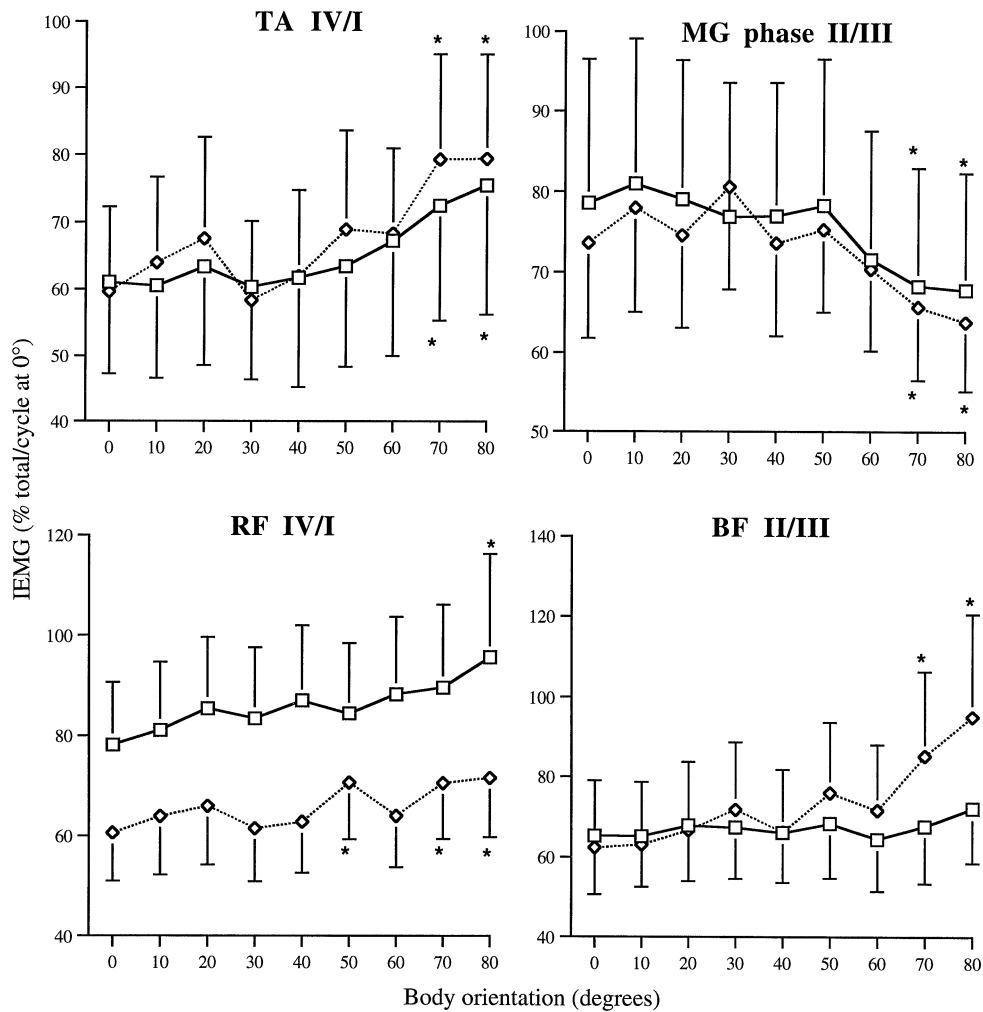
Fig. 11 Mean activity during the two contiguous phases with greatest total activity (based on control subjects), for all control subjects (open squares) and all subjects with hemiplegia (open diamonds) at each body orientation. Values represent total IEMG for these phases as percentages of overall IEMG occurring at 0° of body orientation for each muscle. Bars represent the SEM. Asterisks represent values that were different when compared with 0° using a *post hoc* Tukey comparison with a $P < 0.05$ corrected for multiple comparisons. In the cases of TA, MG, and RF activity, the changes in plegic legs paralleled the changes in control legs. However, BF muscle activity was altered at vertical body orientations for plegic legs, but not for control legs.

positive work, so the exaggerated increase in positive work represented additional force-generation capacity that was not used at more horizontal orientations.

The biomechanics of tilted pedalling dictate that increased extensor activity is necessary during the downstroke in more vertical postures to overcome the increased torque produced by the weight of the opposite limb during its upstroke (Brown *et al.*, 1996*a*). With hemiplegia, extensor tone may already dominate the control of the pedalling task and, if prolonged into the upstroke, may be responsible for the increased negative work that is exacerbated by verticality. Therefore, increased extensor tone with verticality could be a major contributor to the parallel increase in both net positive and net negative work done.

The observation that verticality results in an increased overall excitability in flexors and extensors of persons with hemiplegia can be used therapeutically to grade the level of activity during exercise, as well as to affect the relative balance of activity between ankle flexors and ankle extensors. For example, if the goal of therapy is to minimize overall muscle tone during movement, then pedalling can be practised in less vertical body orientations. However, if the goal of therapy is to increase muscle force generation, then more vertical body orientations would be indicated. Also, the more vertical orientations would enable preferential excitation of TA muscle activity over MG muscle activity, thereby reducing the frequently observed coupling of weak TA tone with strong MG tone. Concurrent studies of the therapeutic effects of gravitational stimuli on the recovery of motor function after stroke have yielded promising results (Brown *et al.*, 1994).
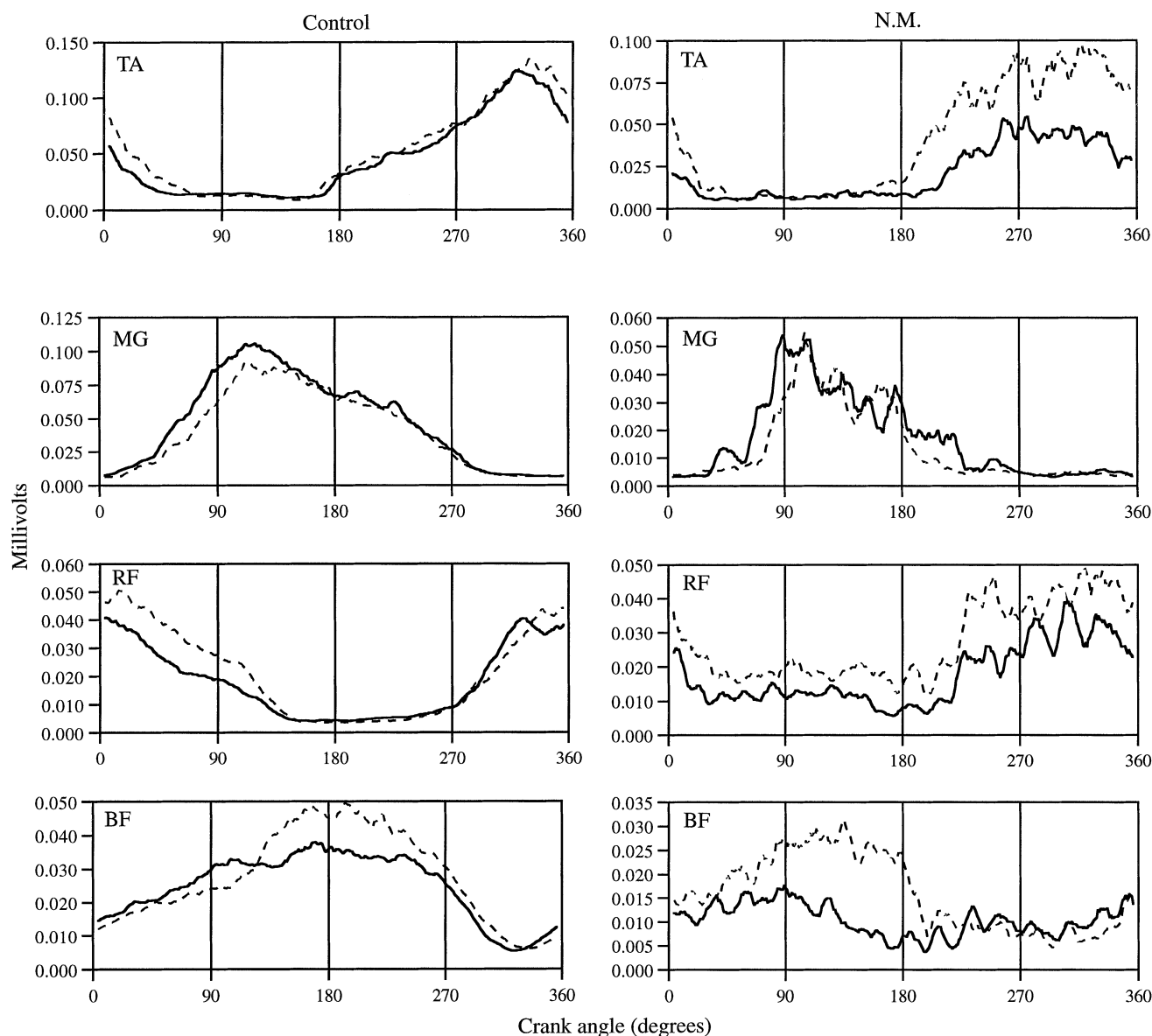
Fig. 12 The EMG, averaged, rectified and smoothed with a 9-ms moving average window, from TA, MG, RF and BF muscle groups at the 0° (continuous line) and 80° (broken line) body orientations. Left panels represent average curves generated from all control subjects. Right panels represent average curves from N.M., a representative subject with hemiplegia. EMG profiles for all plegic muscles demonstrate systematic changes in the timing and amplitude as a function of body orientation, with N.M. also showing pathological early RF muscle onset and early BF muscle onset.

References

Benecke R, Conrad B, Meinck HM, Hohne J. Electromyographic analysis of bicycling on an ergometer for evaluation of spasticity of lower limbs in man. In: Desmedt JE, editor. Motor control mechanisms in health and disease. New York: Raven Press, 1983: 1035–46.

Bobath B. Adult hemiplegia: evaluation and treatment. London: Heinemann Medical Books, 1978.

Boorman G, Becker WJ, Morrice BL, Lee RG. Modulation of the soleus H-Reflex during pedalling in normal humans and in patients with spinal spasticity. J Neurol Neurosurg Psychiatry 1992; 55: 1150–6.

Brooke JD, Cheng J, Misiaszek JE, Lafferty K. Amplitude modulation of the soleus H-reflex in the human during active and passive stepping movements. J Neurophysiol 1995; 73: 102–11.

Brown DA, DeBacher GA. Bicycle ergometer and electromyographic feedback for treatment of muscle imbalance in patients with spastic hemiparesis. Phys Ther 1987; 67: 1715–9.

Brown DA, Kukulka CG. Human flexor reflex modulation during cycling. J Neurophysiol 1993; 69: 1212–24.

Brown DA, Burgar CG, Kautz SA, Dairaghi CA, Dunn-Gabrielli S. Improving lower extremity force symmetry in individuals with hemiplegia. In: Taguchi K, Igarashi M, Mori S, editors. Vestibular and neural front. Amsterdam: Elsevier, 1994: 263–6.

Brown DA, Kautz SA, Dairaghi CA. Muscle activity patterns altered during pedaling at different body orientations. J Biomech 1996a; 29: 1349–56.

Brown DA, Kautz SA, Dairaghi CA. Prolonged muscle activity versus premature onset during pedaling in persons with post-stroke hemiplegia. Society for Neuroscience 26th Annual Meeting. 1996b: 2041.

Bruggencate G ten, Burke R, Lundberg A, Udo M. Interaction between the vestibulospinal tract, contralateral flexor reflex afferents and Ia afferents. Brain Res 1969; 14: 529–32.

Brunnstrom S. Movement therapy in hemiplegia: a neurophysiological approach. Hagerstown: Harper and Row, 1970.

Capaday C, Stein RB. Difference in the amplitude of the human soleus H reflex during walking and running. J Physiol (Lond) 1987; 392: 513–22.

Carr JH, Shepherd RB. A motor relearning programme for troke. 2nd ed. London: Heinemann, 1987.

Chan C, Kearney R. Influence of static tilt on soleus motoneuron excitability. Neurosci Lett 1982; 33: 333–8.

Davis BL, Cavanagh PR. Simulating reduced gravity: a review of biomechanical issues pertaining to human locomotion. [Review]. Aviat Space Environ Med 1993; 64: 557–66.

Duncan PW, Propst M, Nelson SG. Reliability of the Fugl-Meyer assessment of sensorimotor recovery following cerebrovascular accident. Phys Ther 1983; 63: 1606–10.

Duncan PW, Goldstein LB, Matchar D, Divine GW, Feussner J. Measurement of motor recovery after stroke. Outcome assessment and sample size requirements. Stroke 1992; 23: 1084–9.

Fries W, Dieterich M, Brandt T. Otolith contributions to postural control in man: short latency motor responses following sound stimulation in a case of otolithic Tullio phenomenon. Gait Posture 1993; 1: 145–53.

Fugl-Meyer AR, Jaasko L, Leyman I, Olsson S, Seglind S. The post-stroke hemiplegic patient. 1. A method for evaluation of physical performance. Scand J Rehabil Med 1975; 7: 13–31.

Gregor RJ, Broker JP, Ryan MM. The biomechanics of cycling. [Review]. Exerc Sports Sci Rev 1991; 19: 127–69.

Hoy MG, Zernicke RF. Modulation of limb dynamics in the swing phase of locomotion. J Biomech 1985; 18: 49–60.

Hull ML, Hawkins DA. Analysis of muscular work in multisegmental movements: application to cycling. In: Winters JM, Woo SL-Y, editors. Multiple muscle systems: Biomechanics and movement organization. New York: Springer-Verlag, 1990: 621–38.

Kautz SA, Hull ML. A theoretical basis for interpreting the force applied to the pedal in cycling. J Biomech 1993; 26: 155–65.

Kleinbaum DG, Kupper LL, Muller KE. Applied regression analysis and other multivariate methods. 2d ed. Boston (MA): PWS-Kent, 1987.

Lawrence DG, Kuypers HG. The functional organization of the motor system in the monkey. I. The effects of bilateral pyramidal lesions. Brain 1968a; 91: 1–14.

Lawrence DG, Kuypers Hg. The functional organization of the motor system in the monkey. II. The effects of lesions of the descending brainstem pathways. Brain 1968b; 91: 15–36.

Magnus R. Cameron prize lectures on some results of studies in the physiology of posture. Lancet 1926; 2: 531–6, 585–8.

McMahon TA. Muscles, reflexes, and locomotion.Princeton (NJ): Princeton University Press, 1984.

Meinck HM, Benecke R, Conrad B. Spasticity and the flexor reflex. In: Delwaide PJ, Young RR, editors. Clinical neurophysiology in spasticity. Amsterdam: Elsevier, 1985: 41–54.

Newmiller J, Hull ML, Zajac FE. A mechanically decoupled two force component bicycle pedal dynanometer. J Biomech 1988; 21: 375–86.

Redfield R, Hull ML. Prediction of pedal forces in bicycling using optimization methods. J Biomech 1986; 19: 523–40.

Rosecrance JC, Giuliani CA. Kinematic analysis of lower-limb movement during ergometer pedaling in hemiplegic and nonhemiplegic subjects. Phys Ther 1991; 71: 334–43.

Tanaka R. Reciprocal Ia inhibition during voluntary movements in man. Exp Brain Res 1974; 21: 529–40.

Teasdall RD, Park AM, Porter WE, Magladery JW. Electrophysiological studies of nerve and reflex activity in normal man. VI. Excitation and inhibition of 2-neuron reflexes by impulses in other nerves. Bull Johns Hopkins Hosp 1951; 88: 538–48.

Yanagisawa N, Tanaka R, Ito Z. Reciprocal Ia inhibition in spastic hemiplegia of man. Brain 1976; 99: 555–74.

Young LR. Perception of the body in space: mechanisms. In: Brookhart JM, Mountcastle VB, Darian-Smith, editors. Handbook of physiology, Sect 1, Vol 3, Pt 2. Bethesda (MD): American Physiological Society, 1984: 1023–66.