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Foot force direction control during leg pushes against fixed and moving pedals in persons post-stroke

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Abstract

The component of foot force generated by muscle action (F_m) during pedaling in healthy humans has a nearly constant direction with increasing force magnitude (J Mot Behav 32 (2000) 151–62; Exp Brain Res 148 (2003) 50–61). The present study investigated the effect of stroke on the control of foot force. Ten individuals with hemiparesis secondary to a cerebral vascular accident performed pushing efforts against translationally fixed and moving pedals on a custom stationary cycle ergometer. We found that while F_m direction remained constant with increasing effort in both the fixed- and moving-crank conditions for both limbs, the orientation of that force component differed between limbs. The non-paretic limb produced the same F_m orientation as seen previously in healthy humans. However, relative to the non-paretic limb, the paretic limb force line-of-action was shifted away from the hip and closer to the knee in the sagittal-plane for both pedal motion conditions. In the frontal plane, the paretic limb force line-of-action was shifted laterally, closer to parallel to the midline, for both pedal motion conditions. These shifts were consistent with previously reported lower limb muscle weakness and alterations in muscle activation observed during pedaling tasks following stroke. The finding of similar orientations for static and dynamic pushing efforts suggests that limb posture could be a trigger for relative muscle activation levels. The preservation of a constant direction in F_m with increasing force magnitude post-stroke, despite an orientation shift, suggests that control of lower limb force may be organized by magnitude and direction and that these two aspects are differentially affected by stroke.

Keywords: Force; Direction; Pedal; Stroke; Leg

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1. Introduction

The force humans exert on the environment is a useful measure in the study of motor control because this force represents the final common output of the motor system in its effort to perform many common tasks. Environmental interaction forces with a particular direction are required to successfully complete some tasks, and control of that force direction must often fulfil many requirements simultaneously. For example, upright walking requires foot forces that must propel the body forward, while maintaining balance and adjusting to external perturbations (uneven pavement, a strong gust of wind, etc.). On the other hand, there are also tasks that do not strictly require a particular force direction. Opening a door and pedaling a bicycle are examples of tasks that demand only some component of force have a particular direction for successful task completion. The second component of force, and thus the force direction, is unspecified. Examining the resulting orientation of force in a task with unspecified force direction provides an opportunity to probe how the motor system chooses to solve a problem with multiple solutions, a situation of motor redundancy first noted as a key issue in motor control by Bernstein [3]. Investigating how the system chooses to direct force when the particular direction is unspecified may also provide insight into how tasks with strict force direction requirements are performed.

Locomotion studies previously conducted in our lab using a cycling paradigm demonstrated that healthy humans control force direction in a stereotyped way when asked to push in 'the most comfortable manner.' That control was observed as directional consistency in the component of the force of the foot on the pedal (foot force) produced solely

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by muscular effort [1,2]. The directional character of the muscular component of foot force (F_m) was reflected in the shape of the force paths, defined as the set of points traced by the head of the foot force vector during pushing efforts. A linear force path resulted from constant F_m direction; each additional force vector produced by muscle action was in the same direction as the force magnitude increased. The directional character of F_m was of particular importance because it gave insight into how sets of muscles were coordinated during the effort [4,5]. A linear force path for a particular limb posture indicated coordinated ratios of increasing muscle activation around the involved joints. Such coordination would be highly advantageous to the motor system in that coupling the activity around multiple joints would diminish the computational burden associated with controlling the action of a multisegmented limb [3]. Investigation into whether such a control strategy is retained following stroke would increase understanding of both the specific effect of stroke and the organization of control in intact humans.

Elucidating the control organization of such coordinated actions in non-disabled humans can be difficult given the multiple levels of control functioning simultaneously. However, the findings in neurologically disabled humans are potentially easier to understand [6]. Many studies have shown that a key, repeatable, pattern evidenced in post-stroke hemiparesis is disturbed generation of lower limb force [7–10]. Work output capabilities and individual muscle activation patterns in the paretic limbs of this population have been examined [11–16]. However critical investigation into the control of lower limb force magnitude and direction has not been reported. Additionally, studies that have examined post-stroke motor deficit during cycling have based their analyses on muscle activity and force output averaged over multiple crank cycles, and the values were generally grouped either by quadrant of motion or whole cycle [11,13]. Although there has been evidence of systematic increases in muscle activity with workload in a general sense over a range of motion [12], the question remains as to how force output at a particular crank angle is controlled during both static and dynamic tasks. By focusing analysis on one crank angle and investigating the force magnitude and direction of push efforts during both static and dynamic tasks, this study aimed to provide insight into the control of individual lower limb actions under different movement conditions.

The goal of this study was to determine if the strategy of directional invariance used by healthy humans to increase foot force was preserved in patients with post-stroke hemiparesis. Based on the ability of persons to systematically increase muscle activity post-stroke [12] we hypothesized that the linear nature of the force paths would be similar for both legs of these individuals. However, previously observed patterns of leg muscle weakness were likely to alter the ratios of joint torques produced during pushing. Thus, we also hypothesized that the orientation of the force path for the paretic leg would differ from the non-paretic leg. Linearity and direction of force was examined during pushing efforts against pedals attached to both fixed and moving cranks performed by each leg of individuals post stroke. The kinematically constrained task of cycling allowed data analysis that focused on a single crank angle.

2. Methods

Ten participants (two female, eight male) between the ages of 36 and 63 (mean ± S.D. = 50.1 ± 8.3 years) with post-stroke hemiplegia pedaled a stationary recumbent bicycle. Participants had sustained a unilateral cerebrovascular accident (time post-stroke ranged from 0.4 to 10 years; mean = 5.9 ± 3.4 years) with residual lower limb plegia; had no severe perceptual, cognitive, or sensory deficits, no significant lower limb contractures, and no significant cardiovascular impairments contraindicative to pedaling. Table 1 summarizes the participants' clinical data.

Both the University of Wisconsin–Madison and Northwestern University Institutional Review Boards approved the experimental protocol for participation of humans in research. Informed consent was obtained from each participant before the experiment began.

2.1. Instrumentation

A custom-made bicycle ergometer with instrumented pedals, a seat with a backrest, and a motor driven crank was used (Fig. 1). Lower limb motion in the sagittal-plane during seated cycling was constrained by the seat, backboard, and pedal. Because cycling is an activity that naturally occurs almost exclusively in the sagittal plane, no additional means of restraining motion in the frontal plane was present in the ergometer design. However, there is a tendency in the post-stroke hemiparetic population for external rotation of

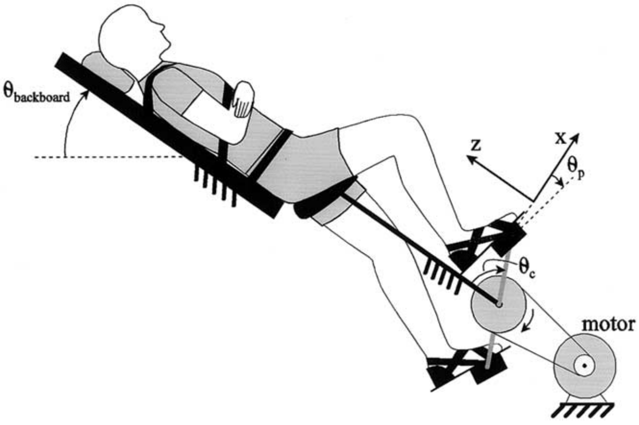

Fig. 1. Experimental set-up. The participants rested against a fixed seat and backboard ($\theta_{backboard} = 35°$). An electric motor held crank position constant during the fixed-crank pushes and maintained constant velocity during the moving-crank condition. The origin of the coordinate system (x, y, z) was at the intersection of the crank pivot and the mid-sagittal plane. The coordinate system shown here is offset from the origin for clarity. The y-axis was positive in the lateral direction for each leg.

Table 1
Clinical data

Participant	Age (years)	Sex	Months post-stroke	Plegic side	Motor score[a]	Sensation score[b]	Total modified Fugl–Meyer score (%)
A[c,d]	52	M	101	Right	26	11	83
B[d]	51	M	117	Right	26	12	89
C[c,d]	56	M	86	Left	28	12	85
D[c]	63	M	80	Right	28	12	87
E[c,d]	55	M	78	Left	24	10	79
F[c,d]	55	M	73	Right	23	12	81
G[c]	40	F	97	Left	18	10	74
H[c,d]	37	M	38	Left	30	10	88
I[c]	49	M	19	Left	20	12	79
J[c]	62	F	92	Right	20	8	74

[a] Based on the Fugl–Meyer scale (maximum score = 34).
[b] Based on the Fugl–Meyer scale (maximum score = 12).
[c] Completed fixed-crank protocol.
[d] Completed moving-crank protocol.

the affected leg at the hip. Therefore subjects were closely monitored for excessive lateral knee motion and verbal feedback was given to aid subjects in maintaining the limb in the sagittal-plane. The fixed seat, backboard, and torso straps maintained a constant hip position throughout the experiment (Fig. 1). Force transducers in each pedal measured the three-dimensional foot force vector (Delta 660, ATI-IA Inc., Garner, NC) from the stroke affected (paretic) and less affected (non-paretic) limbs. Three optical encoders, one at each pedal spindle and one coupled to the crank, provided measurements of the crank and pedal angles. These encoders measured the angle of the pedal relative to the crank (θ_{p}) and crank relative to the laboratory frame (θ_{c}).

Pedaling velocity was controlled by an electric motor. For the fixed-crank condition the motor limited the crank to less than 2° of motion. For the moving-crank condition the velocity was 40.5 ± 0.8 rpm. By controlling crank velocity, the motor provided mechanical isolation of the two pedals such that the force applied to one pedal did not affect the motion of, or force applied to, the contralateral pedal.

2.2. Experimental

While seated on the ergometer, subjects generated force against pedals under two conditions: (a) fixed crank; and (b) crank moving at 40 rpm. Crank angle was zero at the top-dead center position for the right leg and positive clockwise (Fig. 1). The bike was modified to include a tiltable backboard that was rotated up to 35° from horizontal for all sessions.

Force signals, the crank angle, and the pedal angles were digitized (12-bit) for 5 s at 1000 Hz and recorded on magnetic media. We filtered each digitized signal with a three-point median filter followed by a low-pass filter (fourth-order Butterworth, zero-lag, cutoff frequency of 20 Hz). The sprung weight of the pedal was accounted for in the calculation of the laboratory-frame components of the foot force (F_x, F_y, F_z) (Fig. 1).

With the crank angle fixed at 90°(mid-power), the participants performed push efforts to force magnitude targets. The purpose of the force magnitude target was to ensure that participants generated multiple force magnitudes and not to obtain specific levels of effort. The participants first performed a maximum effort, and then force targets ($|\mathbf{F}|_{\mathrm{target}}$) were set at 50 and 70% of that maximum effort for each leg. The pushing efforts were performed with one leg at a time while the other foot rested on its pedal. The force targets were presented in a blocked random order, with each target presented three times. Each force target was presented to the participant as a horizontal line on a real-time bar graph of force magnitude. The participants were instructed to push 'in the most comfortable manner' to achieve a momentary peak force magnitude near the target level and then to relax within the five seconds of data acquisition. Pushing in the most comfortable manner was emphasized while accuracy of matching force target value was not. The participants practiced as much as they desired (typically three to five pushes) with each leg before data was collected.

During the moving-crank session, the participants attempted to match their peak force magnitude in each crank cycle to the $|\mathbf{F}|_{\mathrm{target}}$ while pedaling with both legs. The participants were instructed to pedal as though riding a standard bicycle and adjust peak pushing effort during each crank revolution so that the force feedback line was within the target range. $|\mathbf{F}|_{\mathrm{target}}$ of 40, 60 and 70% of maximum effort were presented in random order, with two trials at each $|\mathbf{F}|_{\mathrm{target}}$. These target levels differed from the fixed-crank pushes because our analysis technique required a spread of data at each crank angle for the moving-crank condition. After a few pedaling cycles to attain steady state, and at least two cycles of peak force magnitude within ±10% of $|\mathbf{F}|_{\mathrm{target}}$, 5 s of data were recorded. This process was repeated for each $|\mathbf{F}|_{\mathrm{target}}$, and for both legs. The participants pedaled with both legs though feedback from only one foot was presented at a time. Analysis focused on the forces generated by the leg for which feedback was provided.

2.3. Data analysis

We analyzed foot forces generated (1) at a specific crank angle position to control for the force of gravity and (2) with a specific limb motion to control for the foot force due to limb inertia. Push efforts that did not achieve a 100 N change in $|\mathbf{F}|$ were discarded. Efforts were analyzed even if the subject failed to achieve, or exceeded, the force target.

We chose the middle of the power stroke as the crank angle for analysis in both the static and dynamic tasks. This is the portion of the pedaling cycle where healthy subjects tend to generate the largest forces and therefore various $|\mathbf{F}|_{\text{targets}}$ were expected to elicit a range of force magnitudes at this crank angle. This is also the phase most critical to power production and therefore locomotive success in pedaling. Thus, this crank angle is appropriate to accurately characterize the geometric properties of the muscular component of foot force.

Due to the kinematic constraints of these tasks, changes in foot force were produced by muscle effort alone [1,2]. Each time the crank angle was at the mid-power position for the leg focused on the task, either fixed there or moving through that angle, the hip and pedal pivot were in the same relative position. The limb had one remaining degree-of-freedom that can be quantified with pedal inclination angle. For the present we assume that the pedal inclination angle was the same every time the crank was at the specified angle. The effect of this assumption will be addressed later. Therefore the limb geometry was always the same at mid-power, yielding a constant component of foot force due to gravity. Additionally, if the crank angular velocity was constant then the limb's acceleration was also the same each time the crank was at the specified angle. This results in a constant inertial contribution to foot force at that crank angle. With both the gravity and inertia forces constant all changes in foot force measured at that crank angle were due solely to changes in muscle activation and thereby reflected neural control.

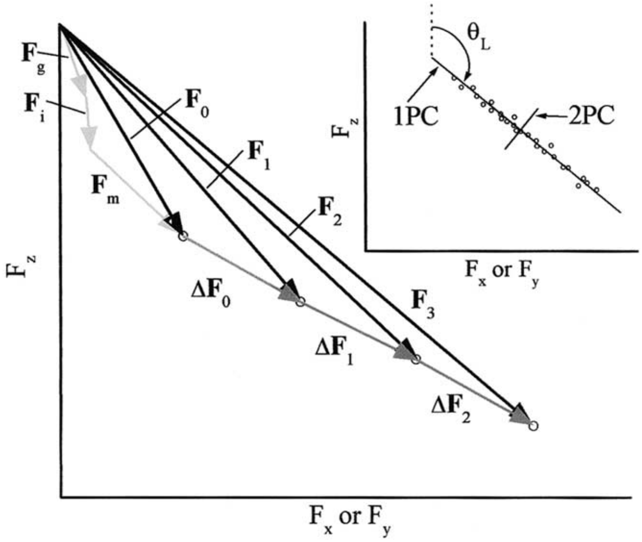

Fig. 2. Force path represents the muscular component of foot force. Foot forces measured for varying levels of pushing effort comprise a force path. The foot force was generated by muscle action, the force of gravity, and inertial effects and thus may be decomposed into vectors due to each of these: muscle action (F_{m}), gravity (F_{g}), inertia (F_{i}). For a given limb posture F_{g} is constant. The pushes conducted in this study were performed with constant crank velocity (0 in the fixed-crank condition, 40 rpm in the moving-crank condition) and thus the inertia contribution to the foot force was also constant. As the participant increased pushing effort we observed that the total foot force vector ($\mathbf{F}_0$, $\mathbf{F}_1$, $\mathbf{F}_2$, $\mathbf{F}_3$) changed magnitude and direction through time in a coordinated manner such that the path of the vector tip was well described by a straight line. Thus, the added force vectors ($\Delta\mathbf{F}_0$, $\Delta\mathbf{F}_1$, $\Delta\mathbf{F}_2$) have nearly constant direction. These added force vectors are due to changes in muscle forces and the resultant joint torques and thus the directional invariance of the added force vectors is a property of motor control. Inset: principal component analysis quantifies force path direction and linearity. The first principal component (1PC) identified the orientation of the force path (θ_{L}) relative to the *z*-axis for both sagittal and frontal-planes. The second principal component (2PC) was perpendicular to 1PC.

2.4. Muscular component of foot force analysis

2.4.1. Generating the force paths

A force path was the set of forces measured at a specific crank angle for a constant crank velocity (either 0 or 40 rpm). This set of forces all had the same components due to gravity and inertia so any differences among the forces were due to changes in muscle activation alone [1,2]. Thus, the force path characterizes the component of foot force generated by muscular effort for pushing efforts against both fixed and moving pedals (Fig. 2). The geometric character of the force path was assessed to provide insight into the organization of lower limb motor control.

To assess the coordination of muscle activation while force was increasing, as opposed to the relaxation phase, the force paths for the fixed-crank condition consisted of a subset of the force data from the 5-s record for each push. We selected the data between the start of data acquisition and the time of peak force magnitude (Fig. 3). Plotting the sagittal- and frontal-plane force pairs through a single push phase yielded a force path (Fig. 3B). The relationship between any two successive points on the force path indicated the direction of the foot force component generated by increases in muscular force. A linear force path resulted from muscle-generated force vectors with constant direction throughout the push effort. After determining that force path geometric character was independent of the time rate of force change, we interpolated points along this force path at 2 N intervals. This achieved a uniform density of points along the path, which aided in quantifying the geometric character of the force path independent of the rate at which individual pushes were performed. The force paths for multiple pushes formed one grouped force path for each subject. All further analysis and discussion refers to the grouped force paths.

The analysis of the moving-crank condition involved constructing the force path as the set of points in force space measured on sequential passes of the crank through the mid-power crank angle (Fig. 4A) [2]. The participants

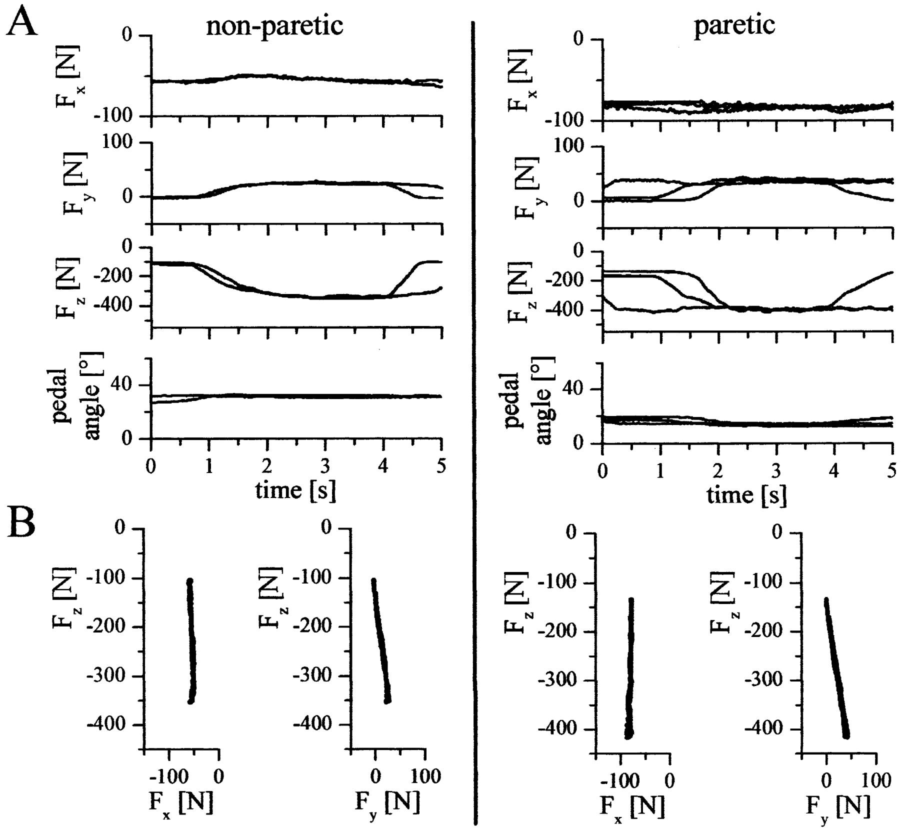

Fig. 3. Sample data: fixed-crank. For the fixed-crank condition force paths were constructed for each push effort and were well described by a straight line, as represented by the data of a single subject. The non-paretic limb data is shown on the left, and the paretic limb data is shown on the right. Panel A: for each limb, the upper four plots show the force components (x, y, z) and pedal angle plotted vs. time. Repeated pushes for one $|\mathbf{F}|_{\text{target}}$ are shown. Only two pushes are shown for the non-paretic limb because the participant failed to push for one of the trials. Also note that the participant began to push before the data acquisition started for one of the paretic limb pushes. The nearly constant pedal angle illustrated in the fourth plot for each limb supports our assumption of constant limb geometry. Panel B: the force paths generated from the fixed-crank pushes in panel A. Frontal-plane paths are on the left, sagittal-plane on the right.

generated a variety of foot force magnitudes guided by the three force magnitude targets (each target was presented twice). Five seconds of data at approximately 40 rpm yielded three to four passes through mid-power, and therefore three to four force measurements per target. This resulted in a total of 18–24 forces for each force path (Fig. 4B) (3 targets×2 repetitions×3–4 cycles). A force path was constructed for each subject at the mid-power crank angle for each leg. Linear interpolation was used to estimate the foot force at this crank angle because the measured data occurred at evenly spaced points in time and not exactly at the mid-power position of the crank.

2.4.2. Principal component analysis

Once the fixed and moving force paths were generated, a straight line was fit to each force path in both the sagittal and frontal planes. The technique of principal components (PC) was chosen over the more common least-squares regression because PC provided a means of quantifying the primary orientation of that data with accuracy that was independent of the orientation itself [17]. The first PC (1PC) aligned with the direction of maximal variation in the data, the second PC (2PC) was perpendicular to 1PC (Fig. 2 inset). The foot forces were then expressed relative to the PC as $(F_{1\text{PC}}, F_{2\text{PC}})$.

2.4.3. Force path geometric character

The geometric character of each force path in each plane was evaluated with the linear merit (LM) [18].

$$\text{LM} = 1-(\sigma_{2\text{PC}}/\sigma_{1\text{PC}}) \qquad (1)$$

where the $\sigma_{1\text{PC}}$ and $\sigma_{2\text{PC}}$ are the standard deviations of the $F_{1\text{PC}}$ and $F_{2\text{PC}}$, respectively. The LM measured the proportion of the variation in the data that was accounted for by a straight line. For data that lay on a line, $\text{LM} = 1$, and for data uniformly distributed in force space, $\text{LM} = 0$.

Three outcome measures were considered: shape of the force paths (LM), orientation of the 1PC of the force paths in the frontal- and sagittal-planes (θ_L relative to the z-axis, Fig. 2 inset), and pedal angle. The independent variables across which those measures were compared were: leg (non-paretic vs. paretic) and motion condition (fixed- vs. moving-crank). Student's t-test was used for all comparisons and a paired t-test was used when appropriate. We considered $P<0.05$ to be significant.

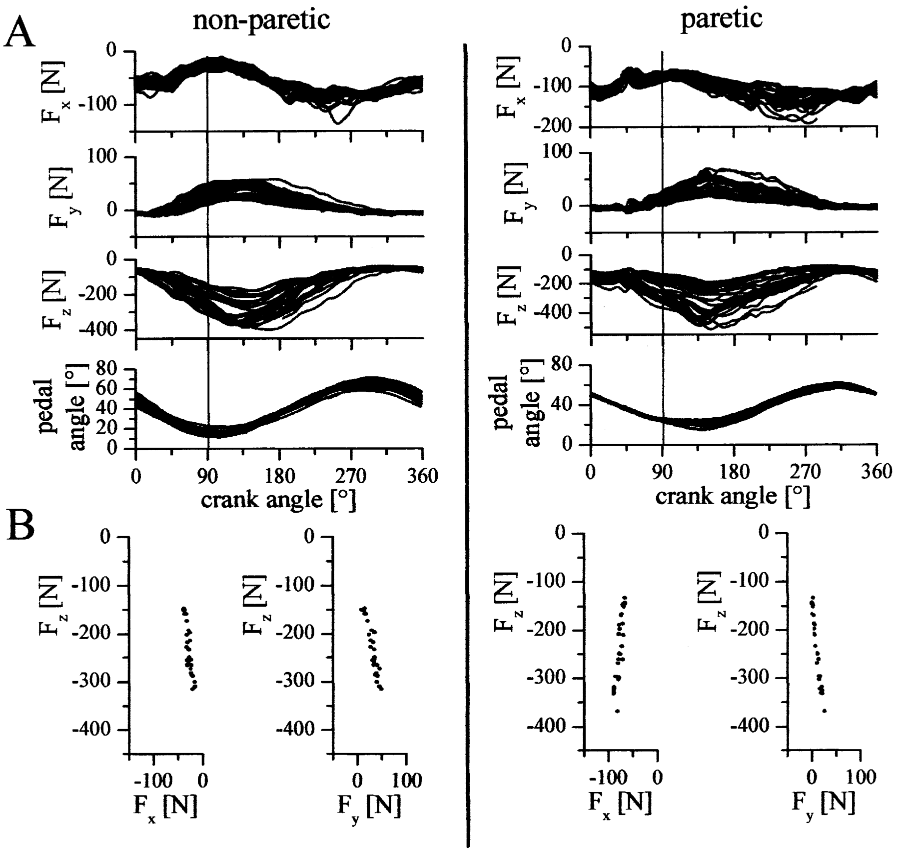

Fig. 4. Sample data: moving-crank. For the moving-crank condition force paths were constructed from a series of push efforts and were well described by a straight line, as represented by the data of a single subject. The non-paretic limb data is shown on the left, and the paretic limb data is shown on the right. Panel A: for each limb, the upper four plots show the force components (x, y, z) and pedal angle plotted vs. crank angle. Repeated pushes for all $|\mathbf{F}|_{\mathrm{targets}}$ are shown. The assumption of nearly constant pedal angle during pushes at a particular point in the crank cycle is supported by the fourth plot which shows a repeatable pattern of changing pedal angle with crank angle. This indicates that the participant was at a similar geometry for each pass through a particular crank angle. Panel B: The forces measured as the crank passes through the mid-power angle (90°) comprise the force path. Frontal-plane paths are on the left, sagittal-plane on the right.

3. Results

Due to either fatigue or technical difficulties, not all subjects were able to complete the entire protocol. To ensure effortful push efforts were being analyzed, force paths of at least 100 N were required for inclusion in the analysis. Together these factors resulted in nine subjects with fixed-crank push efforts, and six subjects with moving-crank efforts. Five subjects completed both conditions. Table 1 indicates which subjects produced forces used for analysis under particular conditions.

3.1. Geometric character of the force paths

Under all conditions the force paths for both the paretic and non-paretic limbs were well described by straight lines (e.g. Figs. 3B, 4B and 5) as indicated by linear merit scores near one. The fixed-crank condition in both the sagittal and frontal planes yielded mean linear merit scores of 0.97 ± 0.01 for the non-paretic limb and 0.96 ± 0.02 for the paretic limb. No significant difference was observed between the limbs in either plane ($P = 0.133$ sagittal and $P = 0.184$ frontal). The non-paretic limb in the moving-crank condition in the sagittal plane had a linear merit score of 0.90 ± 0.08 while the paretic limb was 0.92 ± 0.02 ($P = 0.46$, no significant difference between the limbs). In the frontal plane the non-paretic limb linear merit score was 0.93 ± 0.02 while the paretic limb scored 0.95 ± 0.04 ($P = 0.29$, no significant difference between the limbs). Comparisons between the fixed- and moving-crank conditions for a particular limb revealed no difference in LM scores for the non-paretic limb ($P = 0.10$), while moving-crank efforts with the paretic limb showed significantly lower LM scores than those under the fixed-crank condition ($P = 0.007$). Overall, the geometric character of the paths was well described by a straight line and was comparable between the two limbs.

3.2. Force path orientation

The force paths were oriented differently for paretic compared with non-paretic limbs (Figs. 6 and 7) in both the fixed- and moving-crank conditions. For fixed-crank

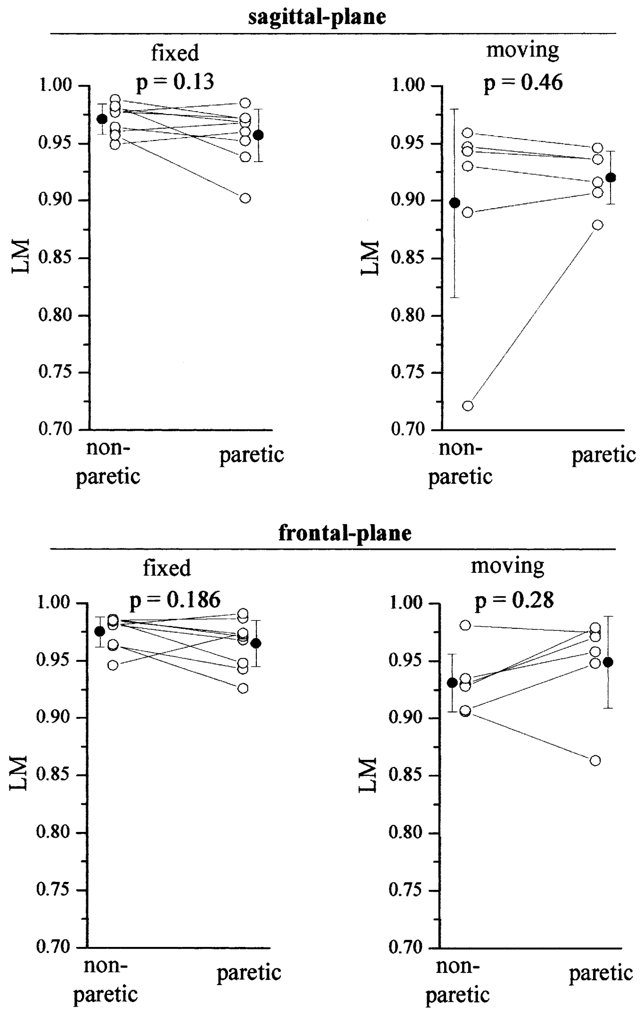

Fig. 5. Force path linearity. Force paths generated by both limbs were well described by straight lines in both the fixed- and moving-crank conditions. Linearity was assessed using the linear merit (LM, see Section 2), where a perfectly linear force path would score 1.0. Individual participant force path LM scores are plotted for each limb in each condition, and in each plane (○). The mean (± S.D.) force path LM score (●) is also shown. No significant difference was seen between limbs ($P>0.05$) in any plane or condition.

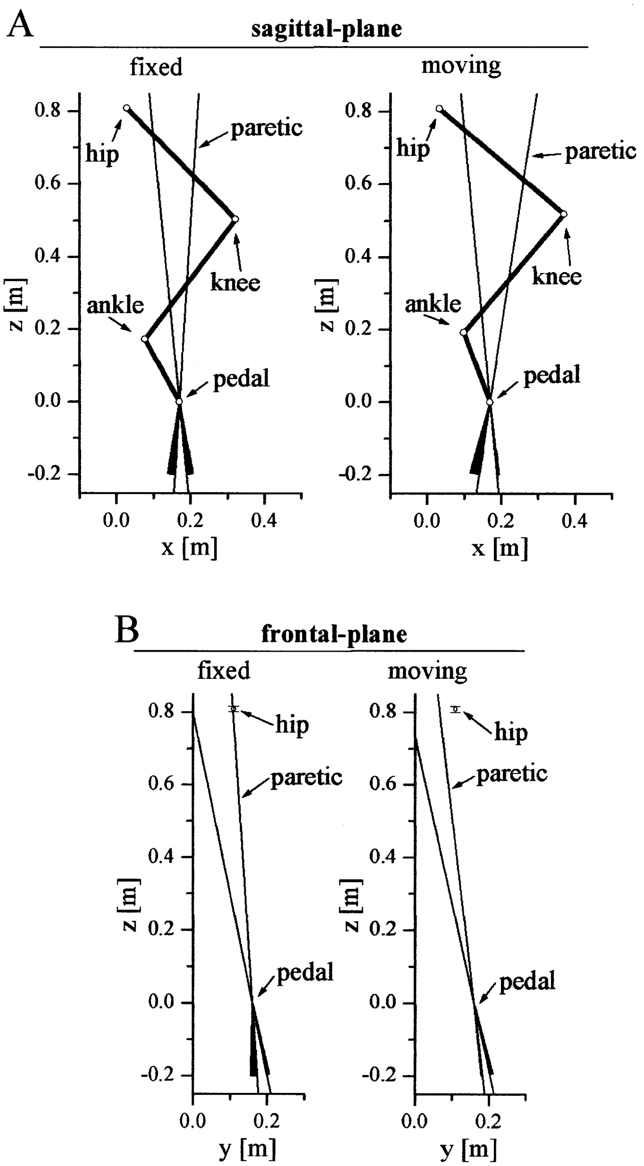

Fig. 6. Force path orientation: graphical. Force path orientation was different for paretic vs. non-paretic limbs in both the sagittal and frontal-planes. Upper panels: stick figures of the mean hip, knee, and ankle positions are displayed with the mean force path directions for each limb and movement condition. The solid wedge below each pedal position represents one standard deviation in force path direction. In both the fixed- and moving-crank conditions the paretic limb force path orientation was rotated anteriorly, away from the hip relative to the non-paretic limb. Lower panels: mean frontal-plane force path directions for both limbs in the fixed-crank (left) and moving-crank (right) conditions. For both the fixed- and moving-crank conditions, the paretic limb force paths were rotated laterally toward the hip of the pushing limb.

pushes in the sagittal-plane the force paths for the paretic limb were rotated anteriorly so that the 1PC line projected further from the hip and closer to the knee relative to the non-paretic limb by an average of 9.0 ± 8.1° ($P = 0.014$; non-paretic = 174.5 ± 4.8°, paretic 183.5 ± 5.6°). In the frontal-plane the paths generated by the paretic limb were rotated laterally by an average of 7.6 ± 6.1° ($P = 0.0079$; non-paretic = 168.7 ± 2.1°, paretic 176.3 ± 4.8°) relative to the non-paretic limb, so as to be closer to parallel to the mid-sagittal plane. Significant differences in the force path orientations were also seen in both planes for the moving-crank condition, and the direction of rotation was the same as in the fixed-crank condition (Figs. 6 and 7). In the sagittal-plane, force paths were rotated toward the knee by an average of 13.8 ± 5.6° ($P = 0.003$; non-paretic = 174.6 ± 1.7°, paretic 188.4 ± 7.1°). In the frontal plane force paths were rotated laterally by an average of 5.7 ± 2.6° ($P = 0.004$; non-paretic = 167.7 ± 2.3°, paretic = 173.4 ± 1.0°). Comparing the force path orientations across conditions within a particular limb revealed no significant differences in either plane. Therefore, the forces generated by the paretic limb in either condition

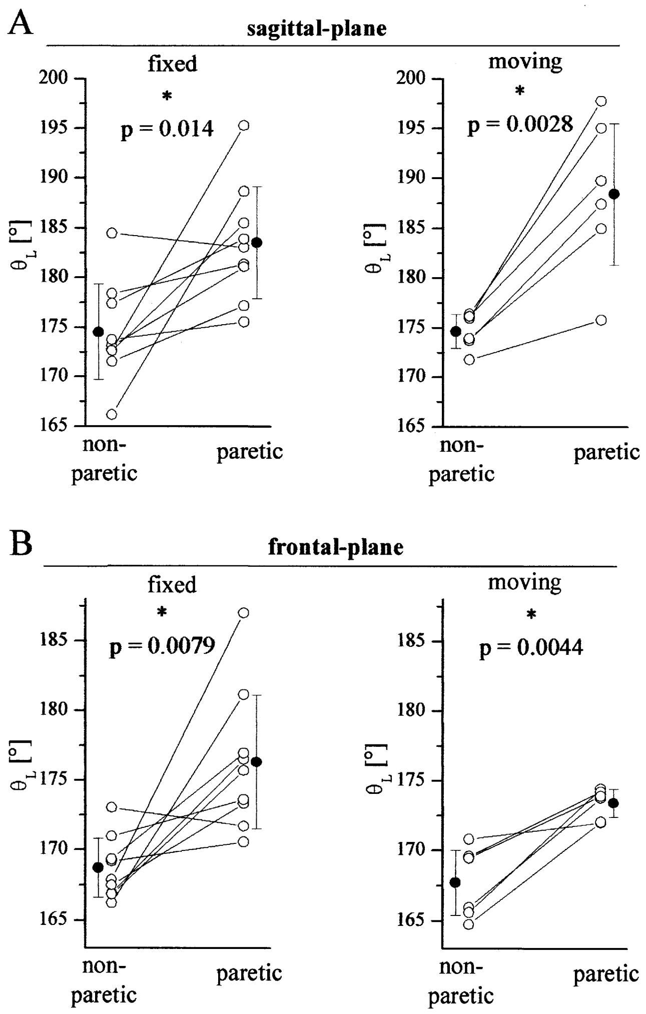

Fig. 7. Force path orientation: numerical. Force path orientations were significantly different between limbs for both the fixed and moving-crank conditions in both planes. Individual participant force path orientations are plotted for each limb in each condition and in each plane (○). The mean (± S.D.) force path orientation (●) is also shown.

were rotated anterior to the hip and lateral relative to the non-paretic limb.

3.3. *Effect of pedal angle changes*

For the fixed-crank efforts participants were found to use similar pedal angles, and therefore similar limb configurations, for pushes with both the paretic and non-paretic limbs ($P = 0.24$). For the moving-crank condition the pedal angles were also similar for the two legs ($P = 0.08$). However, upon closer examination of the moving-crank condition it was observed that subject E performed efforts with large and sudden changes in pedal angle that increased the variance considerably. Removing this subject from the analysis resulted in a highly significant difference ($P = 0.006$) between limbs for the moving-crank condition. The mean difference in pedal angle was $3.2 \pm 1.2^\circ$ with the paretic limb using a heel position dropped lower (greater dorsi-flexion) than the non-paretic limb.

Changes in pedal angle for the set of forces that made up each force path were unrelated to force path geometry. Pedal angle was not correlated with deviations of force off the force path, as determined by regression analysis of F_{2PC} versus pedal angle, in either fixed-crank (mean $r^2 = 0.08$) or the moving-crank conditions (mean $r^2 = 0.041$).

4. Discussion

This study demonstrated that during pedaling tasks force paths in both planes (frontal and sagittal) and conditions (fixed- and moving-crank) were well described by straight lines, confirming our first hypothesis. The force paths were directed differently for the paretic versus the non-paretic limb in a manner consistent with our second hypothesis. To our knowledge, this is the first study to characterize the geometry and direction of foot force in humans with hemiparesis.

The high degree of force path linearity seen from both paretic and non-paretic limbs suggests a significant level of intermuscular coordination is retained post-stroke. In order to generate a linear force path, the muscular component of foot force must remain directionally constant with increasing magnitude. This requires the system to increase overall force while coordinating muscle activity across the hip, knee, and ankle joints. The ability of the system to preserve the direction of F_m across multiple pushes during a dynamic task was evident in the high LM scores for both limbs under both conditions (Fig. 5); suggesting that the chosen force direction was controlled throughout pushing efforts equally well with both limbs. This level of control is consistent with previous studies in healthy human subjects performing the fixed-crank task [1], and suggests that the neural mechanism for maintaining a particular coordination of muscle action across increases in effort is preserved post-stroke.

A critical difference between the fixed- and moving-crank efforts examined in this analysis is that moving-crank force paths were generated from a number of independent push trials, while a static force path resulted from a single push effort. Thus each data point in a dynamic force path was independent, while each point in a static force path was not. Static force paths could easily have been generated from a set of push efforts, just as was done for the moving-crank analysis. The fact that the static force paths for each individual participant had a highly repeatable orientation and linearity means that a force path generated from this set of push efforts would have similar properties to that from a single push. Generating fixed-crank force paths from a set of pushes, instead of a single static effort, requires the participant to perform a larger number of push efforts at a greater variety of effort levels. Given the fatigue limitations of the post-stroke hemiparetic population, analysis from single push efforts were performed for the fixed-crank condition.

Although both the paretic and non-paretic limbs were found to generate linear force paths, the orientation of the force paths differed between limbs. The mean force path orientation of the non-paretic limbs was similar ($P = 0.92$) to that reported in a previous study for healthy humans pushing on a stationary pedal [1]. These observations show that while one aspect of the control was preserved following stroke, another was altered. The preserved control was the maintenance of constant F_m direction with increasing force magnitude discussed above. Control was altered in that the observed orientation of the force path for the paretic limb differed from that of the non-paretic limb and the limbs of individuals that had not had a stroke. A shift in force path orientation is only possible through a redistribution of net joint torques among the hip, knee, and ankle. If two linear force paths had different orientations, then different ratios of torque were used to generate the muscular components of force [1,19]. This is due to the fact that as force path orientation shifts, the moment arms of the force path about each joint change (Fig. 6). For example, a shift in orientation that moved the force path away from the hip required a larger moment arm about that joint, and thus a greater hip torque than a force path passing closer to the hip. The present results in the sagittal-plane indicate that paretic limb force paths pass further from the hip and ankle, and closer to the knee as compared to non-paretic limb paths (Fig. 6A). This results in an increased moment arm at the hip and ankle, and a decreased moment arm at the knee. Since extension torques are required about all three joints for pushing tasks at mid-power, this result implies increased dependence on hip extensors and ankle plantar-flexors, and decreased dependence on knee extensors for force production post-stroke. This is consistent with findings of reduced knee extension and leg press capabilities of the paretic compared to non-paretic limb without decreased ability to produce hip extension [7] as well as more broadly reported weakness in paretic limb knee extensors [7,9,10].

The increased moment arm at the ankle means that the ankle torque must be larger (for a constant force magnitude) in the paretic compared to the non-paretic limb. This is inconsistent with the observation of weak plantar-flexor muscles following stroke [8,20,21]. The reduced capacity to produce torque about a joint will cause the foot force to tend to pass closer to that joint but we observed that the foot force passed farther from the ankle joint for the paretic limb. However, the decreased capacity for torque generation at the knee and ankle tend to cause the foot force to deviate in opposite directions. The fact that the knee is farther from the pedal pivot than is the ankle causes the knee torque weakness to dominate over the ankle torque weakness, resulting in a foot force that is biased toward the knee despite concurrent ankle weakness.

Our finding of repeatable, highly linear force paths raises an important question of neural control verses mechanical constraint. There exists the possibility that force path linearity results not from an imposed control strategy, but as the direct result of the system's mechanics. One mechanical consideration concerns force path orientation and the effect of changing pedal angle during both fixed push efforts and moving-crank cycling. The observed difference in mean pedal angle between limbs in the moving-crank condition is not likely to be the cause of the shift in force path orientation. Not only was the difference in pedal angle small compared to the orientation shift (3.2 vs. 13.8°), but the pedal angle change was in the opposite direction as the force path orientation shift. The foot of the paretic limb was rotated counter-clockwise (as viewed in Fig. 1) while the force paths were rotated clockwise. This was true for each participant as well as for the mean values. If one was to produce push efforts with equivalent force relative to the pedal, but under one condition rotate the pedal counter-clockwise, the force path would also rotate counter-clockwise. A counter-clockwise force path rotation moves the path closer to the hip in the sagittal plane. This is opposite the orientation shift observed here between the paretic and non-paretic limbs, and thus this simple mechanical explanation cannot account for the force path orientation shift. Additionally the lack of correlation between pedal angle and force path geometry ($F_{2\mathrm{PC}}$) further suggests that foot force was controlled independent of pedal angle.

Frontal-plane force path orientation shifts were also seen with the paretic relative to the non-paretic limb. In the frontal-plane, only torque at the hip and ankle can be modulated by muscle forces to alter foot force in the medial-lateral direction. Torque at the knee can alter the axial component of force (z-axis in the present study, see Fig. 1) in the frontal-plane, but its contribution to the medial-lateral force component cannot be actively controlled. The contribution of the ankle to medial-lateral force is likely to be due primarily to passive forces, and to be small as a result of the small foot force moment arm at this joint. Therefore, the primary determinant of force path orientation in the frontal-plane is the result of the ratio of hip abductor torque to knee extensor torque. The paretic limb force paths were rotated laterally compared to the non-paretic limb, resulting in an orientation more parallel to the midline (Fig. 6B). Force paths for both limbs projected at or medial to the hip of the pushing limb, however the non-paretic paths are directed even more medially than the paretic resulting in a greater moment arm at the hip for the non-paretic limb. Greater abduction torque was required by the non-paretic limb relative to the paretic limb to generate this force path orientation. The requirement for abduction torque in the paretic limb has been greatly decreased by the shift in orientation of the force path. This altered dependance is consistent with previous findings in the hemiparetic population during a standing weight-transfer task [22,23].

The force path orientation shift of the paretic limb in the sagittal plane is also consistent with the altered pattern of muscle activation observed previously in dynamic cycling activities [12]. Those studies reported phase shifts in the activation of the paretic limb's rectus femoris (RF), biceps femoris (BF), and gastrocnemius (GA) muscles. The effect of this shift was that at the mid-power crank position the paretic limb activated the BF and GA muscles while showing almost no activity in the RF. In contrast, the non-paretic limb at mid-power showed peak activity in the RF, while the BF and GA were inactive. The effect of these differences was a decrease in net knee extension torque from the paretic limb, and such a decrease is consistent with the sagittal-plane force path orientation shift closer to the knee as observed in this study. It is important to note that in those studies, altered activation patterns were considered issues related specifically to the dynamic nature of the cycling task. However, it is difficult to separate the effects of posture and temporal factors related to the movement around the crank path in dynamic cycling since the circular pedal path couples time and posture. In this study, we separated the effects of posture from other factors in the moving-crank task by also examining force output during fixed-crank pushing efforts and found that the force path orientations were the same as during dynamic cycling (Figs. 6 and 7). This result suggests that posture could have been responsible for the observed force output, because temporal or movement related factors were not present in the fixed-crank task. Timing, as it relates to the time-course of production of a single pushing effort, was certainly still a factor in the fixed-crank efforts, however the timing of the limb moving around the crank cycle was removed from the task. The generation of linear force paths suggests that limb posture might trigger a particular coordinated action to generate force output with a specific constant direction.

The fact that muscle weakness following stroke has been documented extensively in the literature, and that this weakness can explain the sign of altered force path orientation seen here, might seem to suggest that muscle force limitations, and not control, may account for these findings. Although this possibility cannot be definitively ruled out, there is evidence that the pattern of muscle weakness following stroke is present immediately following the vascular accident [7,9]. This suggests that disuse atrophy alone cannot explain the weakness, implicating a change in the ability to issue control commands. This change in control could cause a shift in force path orientation. It is important to note that studies of muscle weakness have generally been performed under conditions of voluntary maximal muscle contraction where there are two critical contributors to the force output; the control signal and the muscle. Reduced force or torque output, interpreted as muscle weakness, could result from decreased capacity of the muscle and/or decreased control signal. The chronic stroke population investigated here certainly exhibited some evidence of classic post-stroke disuse atrophy, however the source of this disuse was likely secondary to control changes. A recent study comparing force output between the paretic and non-paretic tibialis anterior muscle of chronic stroke participants found that direct stimulation of the paretic muscle yielded greater force output than similar stimulation of the same muscle in the non-paretic limb, even though the paretic muscle produced less force under voluntary control [24]. Those findings suggest deficits in voluntary force production can be the result of altered descending command and not muscle weakness. The early onset of force output decrease from particular muscle groups and retention of output capacity under direct stimulation, combined with the preservation of coordination that results in invariant F_m direction observed in the present study, suggest that the findings seen here result, at least in part, from alterations in motor control.

Taken together, post-stroke findings of force path linearity with altered orientation of that force path suggest a dissociation of control localization for force path linearity and orientation. Separations of control have been suggested previously, such as that between force direction and force magnitude, by studies in both the frog [25,26] and cat [27] where force magnitude and direction have been shown to be controlled independently. The observation of preserved force path linearity following stroke suggests that a mechanism coordinating relative muscle activities is located at a level in the motor control system less affected by stroke. The observations of preserved force path linearity post stroke and a spinal mechanism that generates repeatable force direction in frogs and rats [25,28] suggest that the mechanism may reside at a subcortical, likely spinal, level. The force path orientation shifts observed suggest that although the ability to coordinate the activity among muscles was preserved, the ratio of relative activity dictating the particular orientation was altered post-stroke. Therefore, an orientation, or coordination, command may be localized at a level affected by the stroke.

The shift in orientation of the paretic limb foot force observed in this study could be a contributing factor to the asymmetrical weight bearing commonly observed following stroke [29,30]. When transitioning from sit to stand, the feet must generate push force to lift the body. If a person were to push with their paretic limb in a direction different than that of a healthy limb, angular momentum would be disturbed and the person would be likely to fall. The paretic limb force passing more anterior would generate a sagittal-plane torque that would rotate the person back into the sitting posture. One means to compensate for such a disturbance would be to decrease the magnitude of the paretic limb force and thus decrease the disturbing torque. The need to support body weight leads to a compensatory increase in the force magnitude of the non-paretic limb resulting in loading asymmetry. While multiple factors may contribute to post-stroke asymmetrical loading, the redirection of paretic limb force observed in this study may be an important contributor. The finding that persons with hemiparesis can regain loading symmetry [29,30] suggests that the ability to redirect the force in an appropriate manner can be regained with training. Future research should be directed at determining the neural factors producing force redirection and whether specifically retraining force direction control could be an effective therapy to develop loading symmetry.

In conclusion, the present study provides evidence for the preservation of the ability to coordinate relative muscle activity post stroke, as evidenced by foot force linearity. Additionally, an alteration in the particular coordinated ratio of muscle action was demonstrated by the shifted orientation of this linear character in both the sagittal and frontal planes for forces generated by the patients' paretic limbs. This suggests that cerebral vascular accidents could impair aspects of motor planning related to the particular coordinated ratio of muscle activities that result in force output orientation, while leaving intact aspects of control that determine the coordinating action producing overall linear character of foot force.

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