Independence and interaction between the control of moving and holding still in post-stroke arm paresis

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Abstract

Moving and holding still have been posited to be under separate control regimes for both eye and arm movements. The paretic arm after stroke is notable for abnormalities both at rest and during movement, thus it provides an opportunity to address the relationships between control of reaching, stopping and stabilizing. Here, we asked whether independence of these behaviors is preserved in arm paresis. To address this question, we quantified resting postural abnormalities in stroke patients by measuring their biases in force production as they held their hand still in various locations in a planar workspace, and then assessed the influence of these resting force biases on active reaching in the same workspace. We found that patients had marked resting postural force biases at each location. However, these biases did not manifest during any phase of planar reaching movements in the setting of weight support: not during initial acceleration, not to mid-trajectory perturbations, and not during deceleration to a stop. Resting force biases only appeared to switch on after a movement had fully stopped. These findings in stroke suggest that moving and holding still are functionally separable modes of control. At the same time, we found that patients' resting postural force biases mirrored characteristics of abnormal synergies active during movement: they markedly decreased when arm support was provided; they were higher in more distal positions which require breaking out of flexion; and they scaled with the Fugl-Meyer score for the upper extremity (a measure of intrusion of abnormal synergies during active

movement). These three shared features suggest a common mechanism for resting postural biases and abnormal synergies, which appears to be a contradiction given the functional separation of moving and holding still observed in the same patients. To resolve this apparent paradox, we propose a model that predicts a breakdown in the functional separation between reaching and holding still when patients move in the absence of weight support. Thus, the model posits that synergies are the behavioral manifestation of a spillover of posture into movement. Mapping these functional systems onto anatomical and physiological details of lesioned substrate after stroke may provide implementation-level insight into how normal arm motor control is assembled.

Introduction

A longstanding idea in motor control is that moving and holding still rely upon separate control regimes. This separation was demonstrated in the context of eye movement control by the classic work of Robinson (Robinson 1970). Other work suggests this separation may extend to other effectors (Shadmehr 2017). In particular, substantial behavioral (Ghez, Scheidt, and Heijink 2007; 2007) and physiological (Kurtzer, Herter, and Scott 2005; Shalit et al. 2012) evidence supports the idea that a similar dissociation governs the control of reaching and holding still for the arm (Jayasinghe, Scheidt, and Sainburg 2022).

Patients with hemiparesis after stroke typically exhibit deficits in the control of both reaching and holding still (Zackowski et al. 2004; Trombly 1992; Levin 1996; Garland, Gray, and Knorr 2009), making hemiparesis a potent model for understanding the interaction, or dissociation, between these two modes of motor control. There is evidence, for example, that reaching and holding might be differentially affected after stroke depending on lesion side (Mani et al. 2013; Schaefer, Haaland, and Sainburg 2009).

Here, we focused on one particular aspect of holding still: resting at a position. One of the most common and readily recognizable motor symptoms after stroke is abnormal resting posture (Twitchell 1951). For example, the typical hemiparetic arm posture consists of flexion at the fingers, wrist, and elbow (Carr and Kenney 1992) (Figure 1A). Prominent approaches for the treatment of hemiparesis have been based on the idea that abnormal resting posture has a direct deleterious effect upon movement control; they advocated for adjusting overall posture in ways that may minimize such effects (Carr and Kenney 1994; 1992; Bobath 1982).

Our main aim was to determine whether resting abnormalities bleed over into active movements in the post-stroke arm. This is of great interest because the ability to separate moving from holding may be precisely what is lost after stroke, and hence postural abnormalities could contaminate voluntary movement. If this is indeed the case, it might lend credence to the idea in rehabilitation that treating resting abnormalities can benefit movements.

In Experiment 1, we assessed resting postural abnormalities by measuring the abnormal resting postural force biases of cerebral stroke patients in a planar workspace. We investigated how these force biases varied with arm position, presence of arm support, and overall motor impairment. In Experiment 2, we proceeded to assess patients' motor control of reaching and holding in the same workspace. We separately investigated effects upon the initial reach and bringing the reach to a stop, as these two might be separately controlled (Ghez, Scheidt, and

Heijink 2007; Scheidt and Ghez 2007). We also investigated active holding control *after* the movement was over, by examining responses to perturbations that attempted to move the arm off the target, in order to confirm that the same controller is engaged for both passive (as in Experiment 1) and active holding at the same position.

Results

Patients showed resting flexor biases across the workspace

We first assessed post-stroke resting postural abnormalities across a 2D workspace. In Experiment 1, participants grasped the handle of a robotic arm, which passively moved their hand to a series of positions that sampled the workspace in front of them (the setup and example workspace are shown in Figure 1C,D). Participants were instructed to maintain grasp, but otherwise relax their arm and not resist the actions of the robotic arm. Once the participant's arm was passively moved to a given position, the robotic arm would hold still for 5 seconds, enabling us to measure the forces involuntarily exerted by the participant as their hand was held at that location. Each experiment block consisted of three visits to each of about 15-25 positions sampled for each participant, in random order. Participants completed four different experiment blocks: two with each arm, with or without arm weight support (provided by an air sled, Figure 1C).

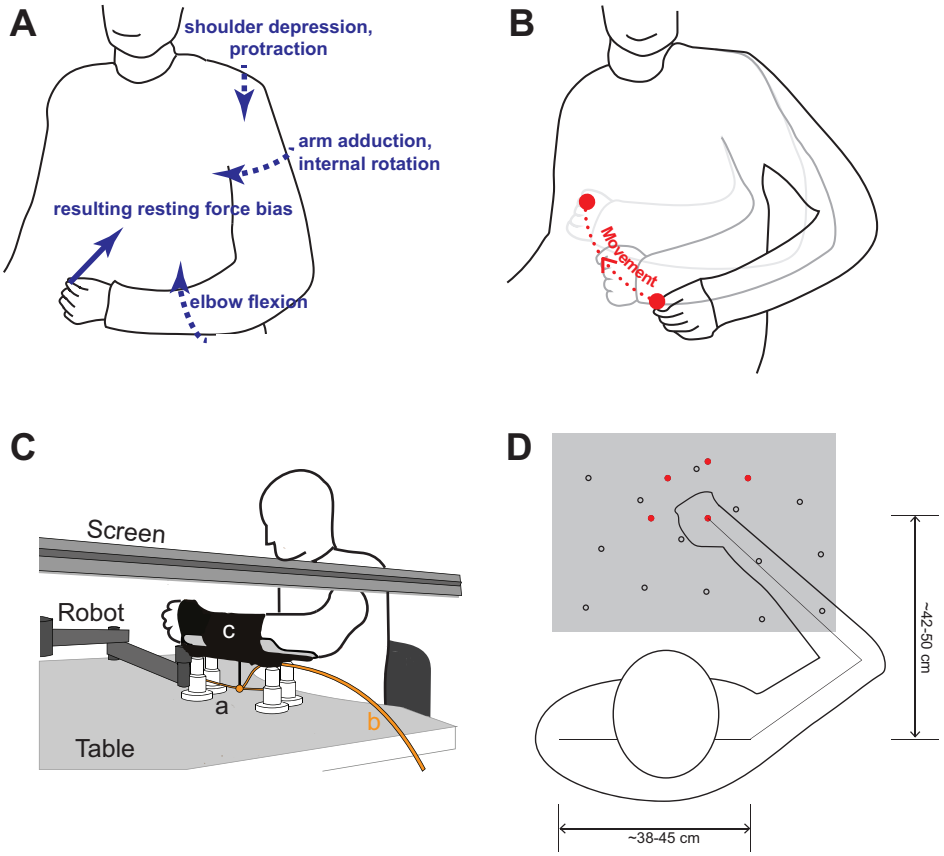

Figure 1: *Conceptual framework and experiment setup.* **A:** A patient exhibiting a typical flexor posture at rest. Dashed arrows indicate elements of the posture: shoulder depression, arm adduction/internal rotation, elbow flexion. The torques involved in each component of the abnormal resting posture translate to a force on the hand (blue arrow); we thus designed an experiment to measure the resting force bias on the hand,

as a marker of the overall postural abnormality. The goal was to compare resting postural force biases to active movement control in the same area (**B**). **C:** Experiment setup. The participant holds the handle of the robotic arm; reach targets and cursor position are projected on a screen on top; for arm support, the participant's arm is strapped on an armrest (**c**) connected to an air sled (**a**) which rests on the table. Air is provided through tube labeled (**b**). **D:** Top-down view of setup, illustrating the different hand positions where resting postural forces were measured in Experiment 1 (open circles). Also shown are the five target positions used in the reaching and holding task for Experiment 2 (filled red circles). The gray box indicates the workspace depicted in Figure 2.

Patients displayed abnormal postural force biases as the hand was held still at various locations across the workspace. Figure 2 shows examples of this for three patients and a typical healthy

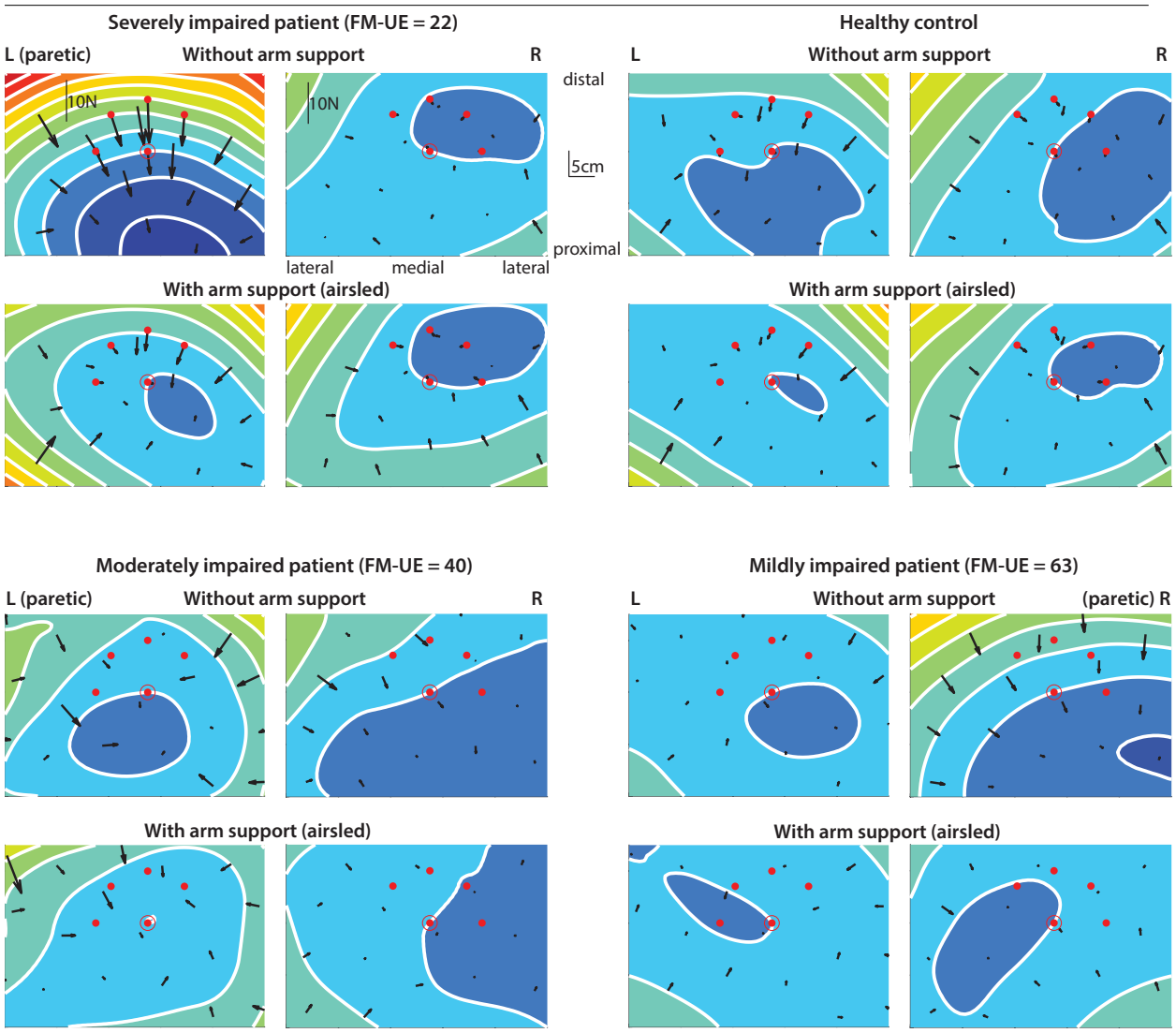

Figure 2: *Examples of resting postural force biases.* Shown are three stroke patients and one healthy

control. Arrows indicate magnitude and direction of abnormal resting postural forces as measured at the hand at each location. Isoclines indicate gradations in force magnitude; different colors indicate different force levels based on these gradations. The red dots are the reach targets, with the center location circled (used in Experiment 2). Note how abnormalities in the paretic side are considerably stronger when arm support is removed. FM-UE: Fugl-Meyer score for the Upper Extremity (0-66).

control participant. The non-paretic arm produced little to no postural force biases, whereas the paretic arm produced substantial postural force biases, particularly when the hand was held in a more distal position. The postural force biases were strongest when participants had to support the weight of the arm against gravity. Moreover, the patient with the highest degree of impairment (top left subplot of Fig. 2, as assessed using the Fugl-Meyer score for the Upper Extremity, FM-UE (Fugl-Meyer et al. 1975)) exhibited the strongest resting postural abnormalities.

To aggregate results across all participants, we focused on five specific hand locations, illustrated by the red dots in Figure 2, and evaluated the corresponding abnormal resting postural force biases. These locations were chosen as they were within the sampled workspace of all participants (as this workspace could differ from one participant to the next), contained both distal and proximal targets, and, importantly, they were also the movement targets used in Experiment 2. Figure 3A shows subject-averaged resting postural forces at each of these five positions and

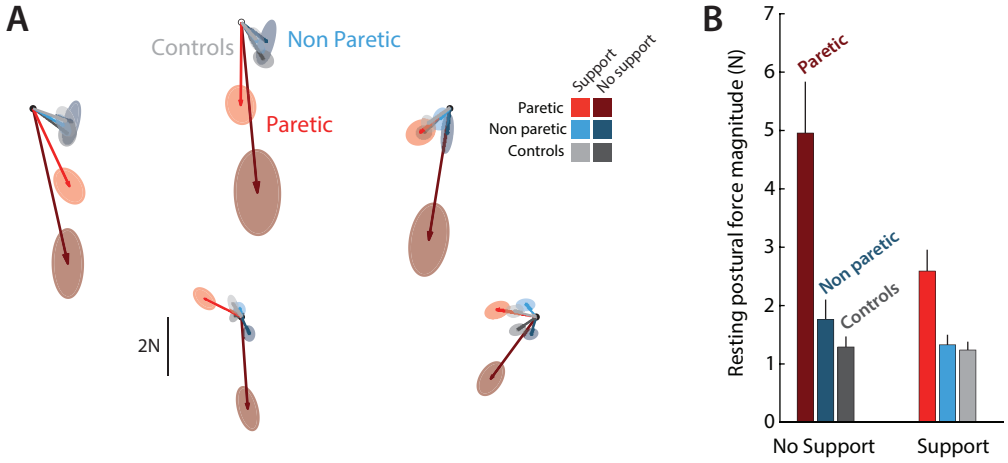

Figure 3: *Average resting postural force biases.* **A:** Average resting postural forces for the paretic (red) and non-paretic (cyan) arms of patients, as well as control participants (gray), illustrating how abnormal forces in the paretic arm are stronger in more distal targets and attenuated when arm support is provided (lighter shades). To average across left- and right-hemiparetic patients, left-arm forces were flipped left to right. **B:** corresponding average resting postural force magnitudes.

Figure 3B shows the corresponding force magnitudes averaged across subjects and the five positions. We made two key observations: first, postural force biases were stronger in locations further away from the body and tended to point towards it; second, postural force biases were roughly halved in magnitude when arm support was provided.

Examining motor control in the same workspace allows us to differentially probe mechanisms of reaching and holding

In Experiment 2, we sought to investigate whether resting postural force biases influence active reaching and holding still in the same workspace. We measured effects upon the initial reach (Figure 4, blue) and bringing the reach to a stop (Figure 4, red), as it has been suggested that these two are separately controlled (Scheidt and Ghez 2007; Ghez, Scheidt, and Heijink 2007); see also (Karst and Hasan 1991; Lestienne 1979). We also asked whether the same controller is engaged while being passively held at a position (as in Experiment 1) vs. actively holding at the same position. For this purpose, we examined active holding control *after* the movement was over (Figure 4, black), using perturbations that attempted to push the arm off the target.

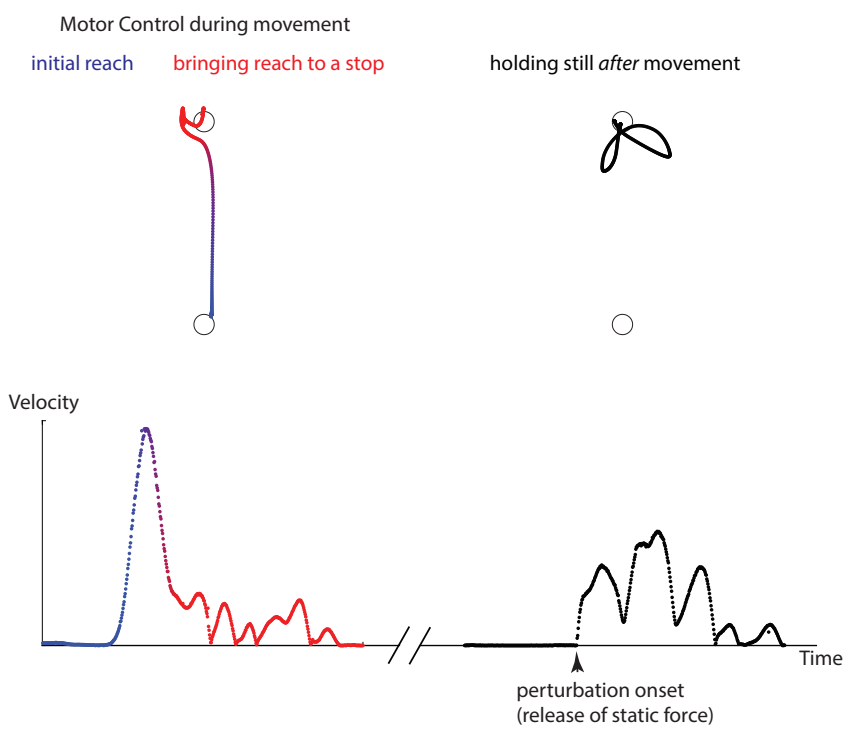

Figure 4: *Three aspects of active motor control that we tested in Experiment 2.* We separately examined the early part of the reaching movement (blue) and the late part, when the arm was coming to a stop (red). This was done by studying both unperturbed movements at different stages and movements that were perturbed with brief force pulses. In addition, we examined active holding control after the movement was

over (black), using perturbations that tried to move the arm away from the held point. Shown is an example of trajectory and (absolute) velocity profiles from the reaching and coming-to-a-stop parts of a trial (left) and active holding against a perturbation after the trial was over (right).

Resting postural force biases did not affect the control of active reaching

In Experiment 2, participants made 10-cm point-to-point reaching movements within an array of five positions (filled circles in Figure 1D), for a total of 8 different movement directions (Figure 5A). Arm support was provided by the air sled (Figure 1C). Patients' movements were generally impaired, taking more time and traveling a longer path to reach the target than controls (Time to target: 1.59±0.12s for patients' paretic side vs. 0.82±0.03s for controls, $p = 0.00006$; Path traveled to target 13.4±0.6cm for patients' paretic side vs. 10.8±0.2cm for controls, $p = 0.0071$, Fig. 5B,C).

We then examined whether patients' movements reflected the resting postural biases measured in Experiment 1 in the same workspace. Would there, for example, be a difference when moving the arm through a high-postural bias area vs. a low-postural bias area? We examined effects of resting postural biases upon the initial reach vs. the approach to a hold position, as separate mechanisms may be involved in the control of each phase of movement (Scheidt and Ghez 2007; Ghez, Scheidt, and Heijink 2007; Sainburg, Ghez, and Kalakanis 1999; Karst and Hasan 1991; Hannaford and Stark 1985); thus, resting postural biases might affect one phase but not the other. Specifically, we examined: (a) whether the direction of resting postural forces at the start position influenced trajectory deviations near the beginning of the movement (Figure 5D) or (b) whether direction of resting postural forces at the end position influenced trajectory deviations near the end of the movement (Figure 5G, after the participant reached within 2cm of the target).

To investigate within-subject effects, we selected, for each patient, the two movement directions for which the corresponding postural forces had strongest opposing effects – i.e. the most rightwards (CW) vs. most leftwards (CCW) with respect to movement direction. The idea behind picking the most extreme values for each individual was to maximize our sensitivity in detecting potential effects of resting postural forces upon active movement. Our analysis found no significant differences in directional biases between these two conditions: while the selected start-point force biases differed considerably between the most CCW and CW cases (1.8±0.4N vs. -1.8±0.3N, correspondingly [negative signs indicating CW forces], $t_{15}=6.24$, $p=0.00002$), the corresponding initial angular deviations did not (0.4±2.1˚ for the most CCW vs. -0.6±2.3˚ for the most CW postural force, $t_{15}=0.27$, $p=0.79$, Figure 5I-K). Similarly, while the selected end-point force biases also differed considerably (1.8±0.3N vs. -1.8±0.4N, for the most CCW vs. CW cases,

t_{15}=6.24, p=0.00002), the corresponding endpoint angular deviations did not (4.4±2.4˚ for the most CCW vs. 5.2±2.6˚ for the most CW postural force, t_{15}=0.30, p=0.77, Figure 5L-N). In summary, our data thus provide no compelling support for the idea that resting postural abnormalities do not directly affect active movement.

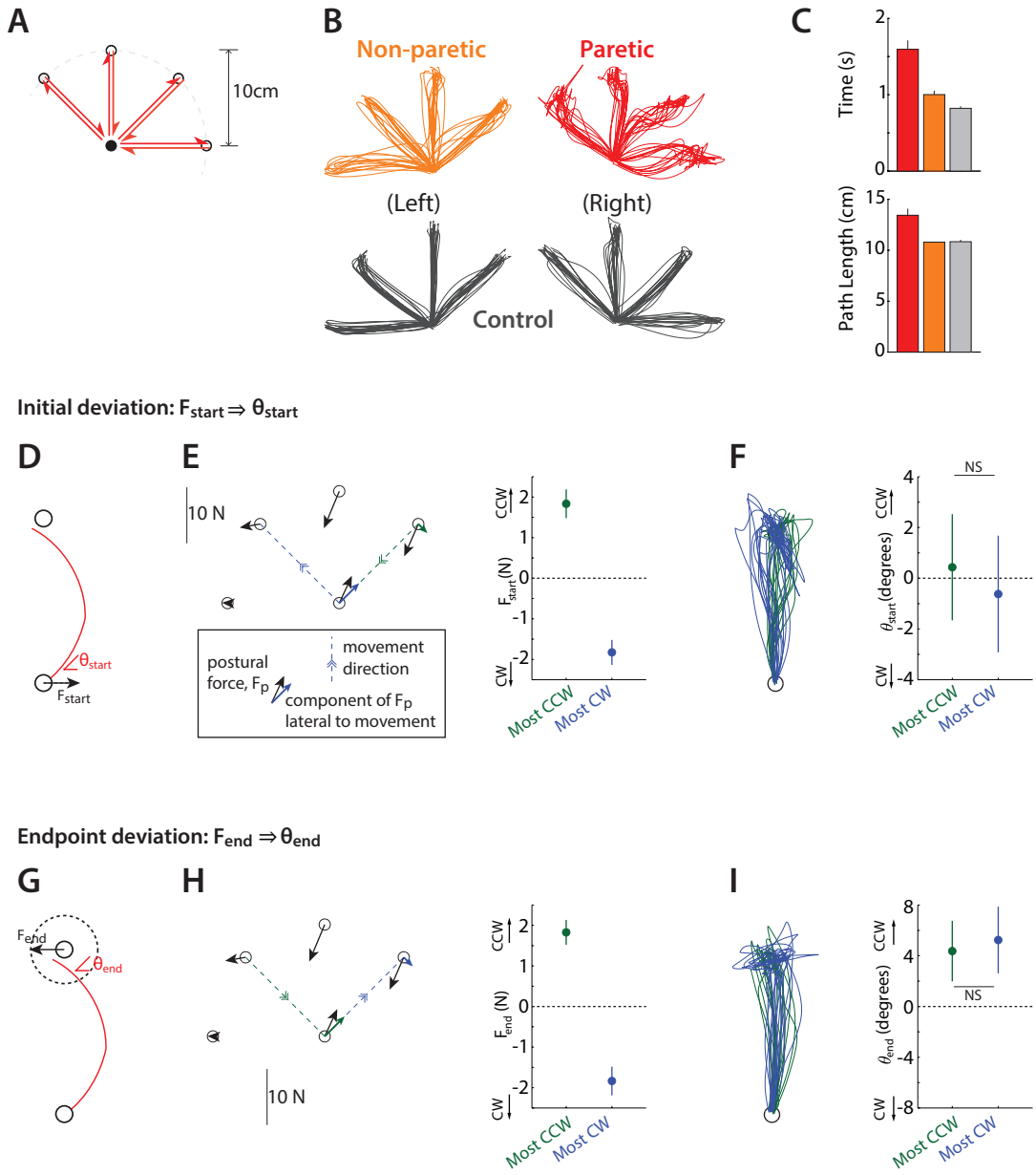

Figure 5: *Abnormal resting postural force biases do not interact with active reaching.* **A:** Target array for Experiment 2 (movement task), illustrating the 5 start/end points of reaches and the 8 movement directions. **B:** Example outwards trajectories (unperturbed trials) for a patient (orange: non-paretic side; red: paretic side) and a healthy control (gray). **C:** Subject-averaged reach performance based on either time (top) or path length to target (bottom) indicates impaired reaching control in patients' paretic side. **D-F:** Within-

subject analysis of whether resting postural forces at movement start bias early movement towards their direction. **E:** For each individual, we selected the direction where F_{start} was the strongest counterclockwise (CCW, green) or clockwise (CW, blue). The left panel shows this selection for an example participant: postural forces at start position were projected lateral to the movement direction, allowing us to select movement directions for which this component was directed the strongest CCW or CW. The right panel shows the magnitude of these selected components across all patients. **F**: Left: Corresponding movement trajectories (rotated so start position is at the bottom and end position at the top) for the directions selected for the same example participant. Right: Average initial angular deviations, θ_{start}, for the selected directions for each participant, revealing no difference and thus no effect of F_{start} upon the movement. **G-I:** same as D-F but for endpoint resting postural forces, F_{end} and endpoint deviations, θ_{end}. Errorbars indicate SEM.

Resting postural force biases did not affect responses to perturbations during movement

A potential limitation in our data and analysis above is that unperturbed reaching movements may provide limited sensitivity in detecting effects of resting postural force biases, as any such effects may be largely compensated by a predetermined motor plan. In that case, it may be better to directly compare these force biases to responses to less predictable, externally applied force perturbations. We thus further assessed how resting postural forces may affect responses to mid-movement perturbations. In 1/3 of randomly selected reaching movements in Experiment 2, we imposed a 70ms duration, 12-N force pulse lateral to the movement (Smith and Shadmehr 2005) 2cm into the reach. Half of these pulses were clockwise (CW, blue in Figure 6A) and the other half were counter-clockwise (CCW, red in Figure 6A). We first verified that these pulses had a clear effect upon movement: perturbed movements took longer to complete in both patients (paretic movement time: 1.72±0.13s vs. 1.59±0.12s, t_{15} =5.31, p = 0.00009) and controls (movement time: 0.88±0.03s vs. 0.82±0.03s, t_8 =5.56, p = 0.0005). Patients generally had impaired response to these pulses compared to their non-paretic side and healthy controls, deviating further (maximum lateral deviation – paretic: 2.5±0.1cm vs. non-paretic: 2.1±0.1cm [p= 0.0008] and controls: 2.1±0.1cm [p = 0.0357] - average of CW and CCW pulses) and taking a longer time to stabilize in the pulse direction (i.e. minimize lateral velocity; paretic: 1.15±0.06s vs. non-paretic: 0.75±0.03s [p=0.00002] and controls: 0.74±0.02s [p = 0.00003]). This is illustrated in Figure 6B,C.

We then investigated whether resting postural forces played any role in patients' response to the pulse perturbation. We hypothesized that, should resting postural forces play a role, they would tend to reduce the effect of the pulse if they were in the opposite direction, and exaggerate it they

were in the same direction. We thus compared the lateral component of resting postural forces in the start position against (a) the maximum lateral deviation in the direction of each pulse and (b) the time taken to stabilize in the pulse direction (settling time). When we compared force magnitude vs. maximum lateral deviation in pulse direction we found no relationship for either CCW pulses (R^2=0.02, p = 0.57) or CW pulses (R^2=0.03, p = 0.55) as shown in Figure 6D, right. Similarly, we found no relationship when comparing force magnitude vs. settling time (CCW pulses: R^2=0.002, p = 0.86, CW pulses: R^2=0.05, p = 0.41, Figure 6D, left). In summary, our analysis did not show any specific, consistent effects of resting posture upon the responses to the pulse perturbations.

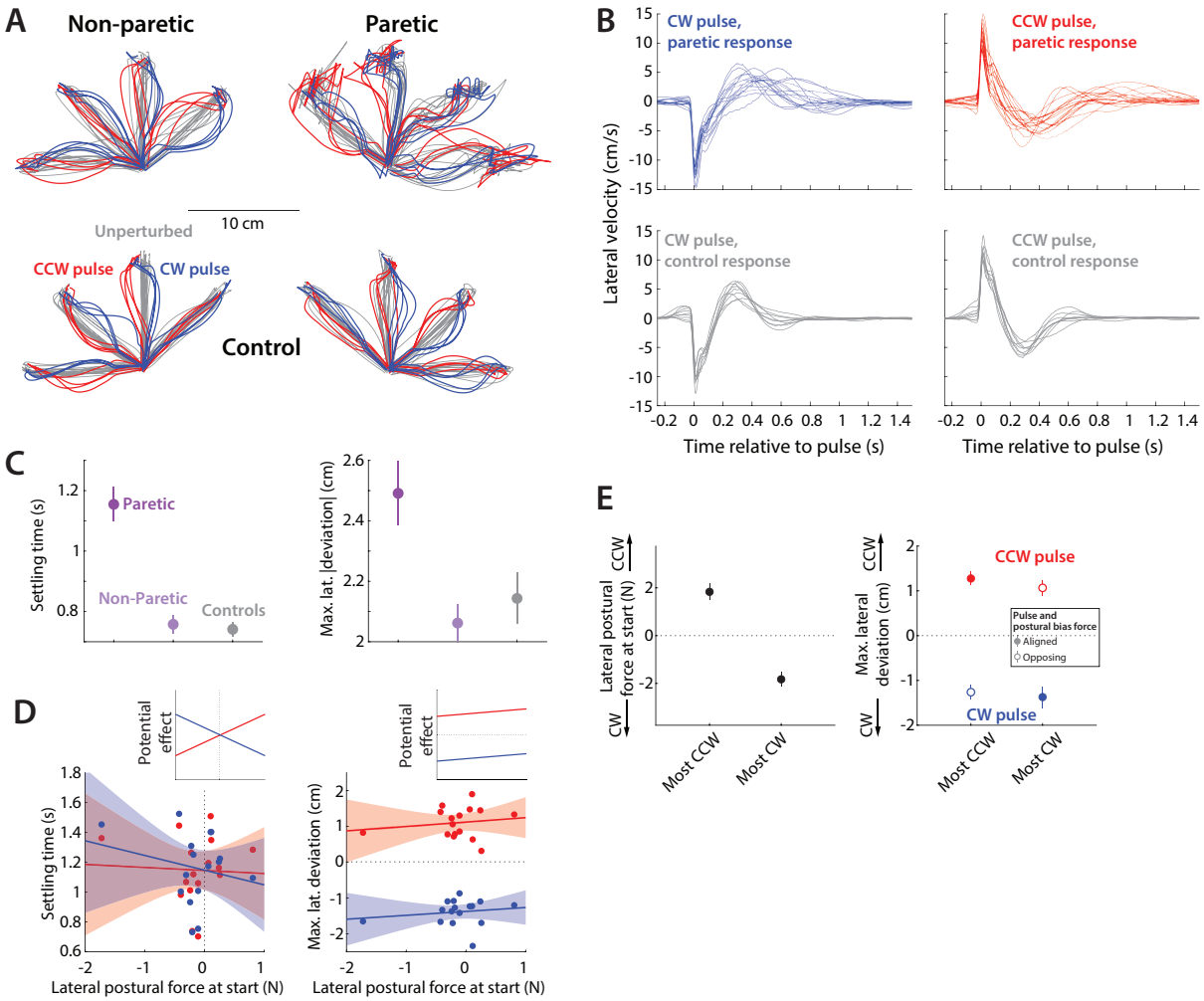

Figure 6: *Responses to movement perturbations and relationship to resting postural forces.* **A:** Examples of perturbed (red: perturbed with CCW pulse; blue: perturbed with CW pulse) and unperturbed (gray) outward trajectories - same individuals as in Figure 4B. **B:** Lateral velocity (positive: CCW to movement) before and after pulse onset, and corresponding responses from controls (gray), illustrating how patients,

in response to the pulse, take longer time to settle and tend to experience larger lateral deviations compared to controls. **C:** Summary performance measures for patients and controls, indicating impaired performance with the paretic side: settling time (left) and maximum lateral deviation on pulse direction (right). **D:** Left: Across-patient comparison between settling time and lateral postural bias force on movement start. Inset indicates expected relationships if resting postural biases were affecting the response against the pulse. Paretic data shown. Red: CCW pulse; Blue: CW pulse. Right: similar to left, but for (signed) maximum lateral deviation for the two types of pulses. **E:** Within-individual analysis: here, for each individual, we selected the movements for which the starting-position resting postural force would be either the strongest CCW or CW (left); we then examined the corresponding maximum lateral deviations (right). Mirroring the analysis shown in D, any potential effects of the most CCW vs. most CW resting postural forces are inconsistent: in one case there is a tendency for increased deviation when the resting postural force is aligned with (filled circles) instead of opposing (open circles) the pulse, and in the other case it is the other way round.

However, this across-subject analysis may mask within-subject differences. Thus, in a manner similar to the previous section, we selected, for each patient, the directions where the starting lateral postural force was most CCW vs. the most CW, and compared the corresponding deviations. We found no clear effect: when resting postural forces were the most opposed to the pulse (vs. most aligned with it) there were no clear differences in deviation along the pulse direction, for neither the CCW pulses (1.28±0.15 vs. 1.07±0.17cm, t_{15} = 0.93, p = 0.36), the CW pulses (1.37±0.23 vs.1.26±0.16cm, t_{15} =0.40, p = 0.69), or with both types of pulses pooled together (from 1.16±0.12 vs. 1.32±0.12cm, t_{15} =0.78, p = 0.81). In short, we found no evidence for an interaction between resting postural force biases and the ability to respond to perturbations that were applied during movement.

Resting postural force biases emerged during active holding at the end of movement

Finally, in Experiment 2, we also investigated the relationship between resting postural force biases and active holding. In 20 out of 96 movements in each block, participants had to hold steady on the target for an additional 5 to 7 seconds (Figure 7A). During this time, the robot imposed a 6N force in one of four directions (45˚, 135˚, 225˚, 315˚, as shown in Figure 7B). This force was gradually applied over two seconds, held at a 6N level for 3-5 seconds, and then abruptly released, acting to displace the arm in the direction opposite to the original force, as illustrated in the examples in Figure 7B.

Patients showed impaired capacity to resist and recover from this perturbation (the abrupt release of the imposed force), taking a longer time (0.96±0.05s for the paretic side vs. 0.79±0.03s for the

non-paretic side [p = 0.014] and 0.78±0.06s for controls [p = 0.04], Figure 7E, left) and traveling a longer path (7.7±0.7cm for the paretic side vs. 5.9±0.3cm for the non-paretic side [p = 0.011] and 5.4±0.5cm for controls [p = 0.024], Figure 7E, middle) to stabilization; moreover, they deviated more in the direction of the perturbation (3.2±0.2cm for the paretic side vs. 2.6±0.1cm for the non-paretic side [p = 0.014] and 2.3±0.2 for controls [p = 0.0087], Figure 7E, right).

To investigate whether resting postural force biases affected the control of actively holding still at across workspace locations, we performed a within-individual analysis analogous to the one we used for unperturbed reaches and moving perturbations. We first projected resting postural forces upon the directions of the static perturbation to assess the component of the resting postural force that opposed or aligned with the static perturbation. We then selected, for each patient, the two position/perturbation direction combinations for which these forces were either the most opposed (Figure 7F, green) to the perturbation or the most aligned (Figure 7F, blue) with it. For these selected position/perturbation direction combinations, we compared capacity to resist and recover from the perturbation, and found that this capacity was indeed better when the resting postural force was in a direction that opposed the perturbation (path traveled to stabilization: 4.4±1.2cm vs. 6.9±1.0cm, p=0.0198; time to stabilization: 0.7±0.1s vs. 1.0±0.1s, p = 0.0023; maximum deviation: 1.9±0.4cm vs. 3.2±0.4cm, p = 0.0003, corresponding to cases with the most opposed vs. the most aligned resting postural force). In short, our perturbations revealed that resting flexor biases switched on after movement was over, providing evidence for separate control between moving and holding still.

Resting postural force biases and abnormal synergies

The observation, from Experiment 1, that resting postural force biases are reduced by external arm support bears parallels to the same effect of arm support upon abnormal synergies active during movement (Sukal, Ellis, and Dewald 2007). Yet, Experiment 2 found no relationship between resting postural force biases and active movement control. To further investigate this apparent paradox, we examined the relationship between resting postural force biases and abnormal synergies in further detail. To assess the level of synergy abnormalities for each patient, we measured their Fugl-Meyer scores for the upper extremity (FM-UE), a scale which was designed to capture abnormal muscle synergy after stroke (Fugl-Meyer et al. 1975; Brunnstrom 1966) and closely corresponds to EMG-based synergy measures (Bourbonnais et al. 1989).

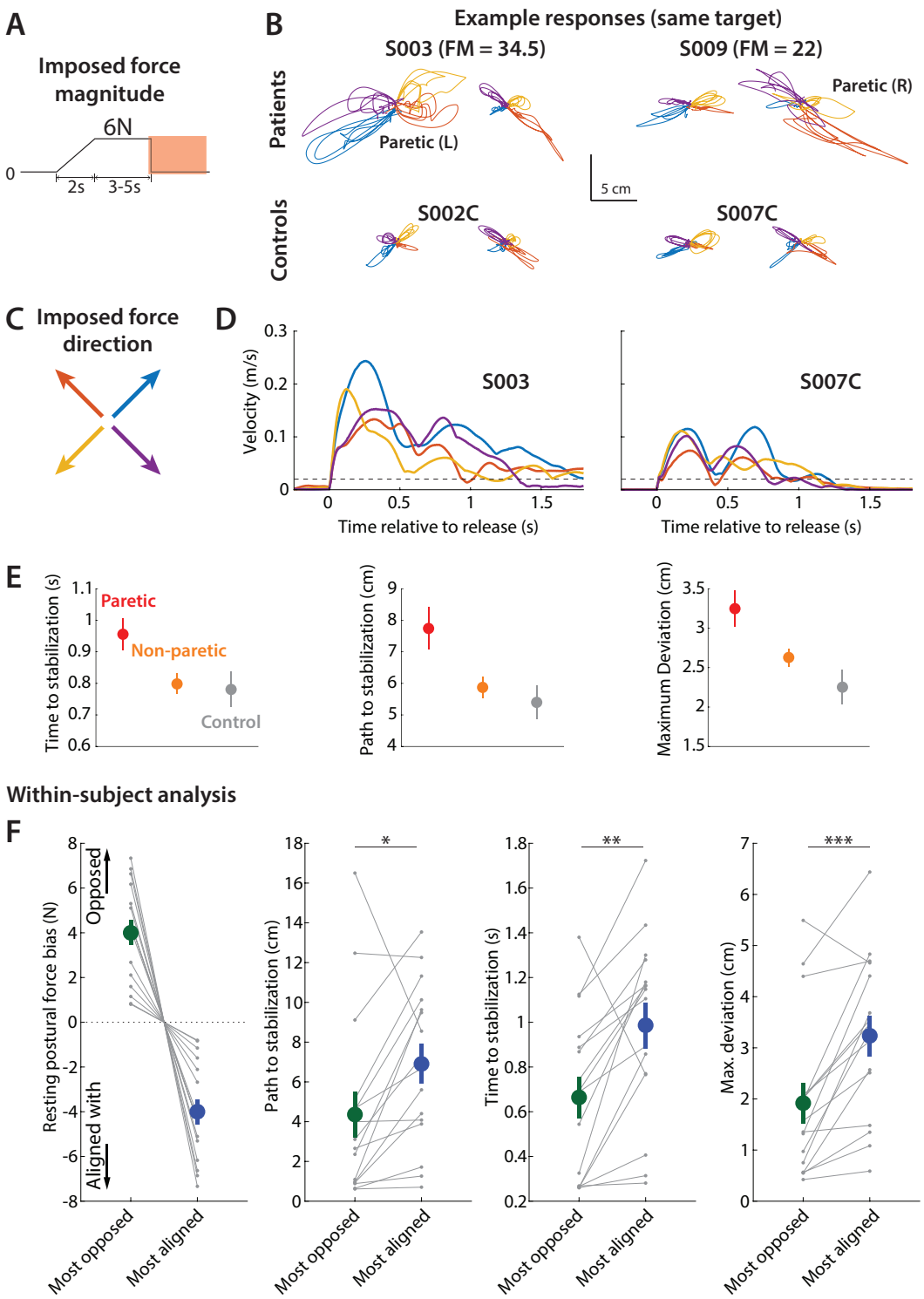

Figure 7: *Responses to static perturbations and their relationship to resting postural forces*. **A:** Time course of the perturbation. **B:** Example responses (all for the same position in the workspace) from two patients (top row) and two controls (bottom row). **C:** Corresponding imposed force directions, the abrupt removal of

which perturbs the movement in the opposite direction (compare with B). **D:** Examples of tangential velocity profiles after the sudden release to the imposed hand force, averaged for all trials at the same position for each participant. Dashed line indicates the 2cm/s threshold used to assess time to stabilize. Left; example patient (paretic side); Right; example control. Colors correspond to different directions of the imposed hand force. **E:** Summary of performance metrics after the perturbation for the paretic and non-paretic side of patients and healthy controls. **F:** Within-subject analysis of the relationship between resting postural forces in the direction of the perturbation vs. performance against the perturbation. For each individual, we selected the two position/perturbation direction combinations for which resting postural forces were either the most opposed (green) to the perturbation or the most aligned (blue) with it. From left to right: forces in selected position/perturbation direction combinations; corresponding path traveled to stabilization; corresponding time to stabilization; corresponding maximum deviation. This analysis suggests that restoring hand position after the perturbation is indeed easier when resting postural forces opposed, rather than were aligned with, the perturbation. Gray dots indicate individual data; colored dots and errorbars indicate mean±SEM.

We entered patients' resting postural bias magnitudes into a mixed-effects ANOVA with FM-UE (continuous), Proximity (distal: the three locations furthest from the body, proximal: the two locations closest to the body, Figure 8, top left) and Support (with air sled, without air sled) as factors, evaluating for main effects and interactions. All factors showed significant main effects, with resting postural force magnitudes decreasing with FM-UE (R^2 = 0.30, $p < 10^{-8}$), target proximity (η^2 = 0.12, p = 0.00003), and weight support (η^2 = 0.12, p = 0.00003). Significant interactions were observed between FM-UE and both Support (p=0.0055) and Proximity (p=0.0037), with Support and Proximity becoming more important for lower FM-UE scores (i.e. higher overall motor impairment / higher synergy intrusion) as illustrated in Figure 8.

In summary, we made three key observations on abnormal resting postural force biases. First, like abnormal synergies, they were exaggerated when active arm support was required. Second, they were higher in more distal positions where the elbow would tend to be extended; if patients were to actively reach to the same locations, they would have faced increased intrusion of flexor synergy. Third, they scaled with the synergy-based FM-UE. These observations thus suggest a common mechanism behind resting postural force biases and abnormal synergies.

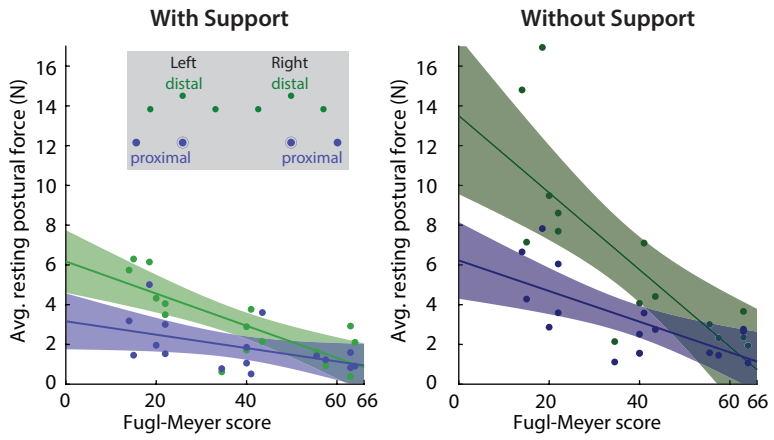

Figure 8. Across-patient relationships of FM-UE (/66, higher scores indicating lower impairment) and resting postural force magnitudes, for distal (green) and proximal (blue) target positions, with (left) and without support (right). Note the strong effects of arm support, proximity, and FM-UE. Lines indicate linear fits; shading indicates 95% confidence interval for each fit.

Discussion

We assessed abnormal resting posture in stroke patients by measuring the resting force biases they involuntarily exerted while their arm was held at different points within a planar workspace (Experiment 1). We found that these resting postural force biases were strongest in more distal positions of the arm, generally pulling the arm toward a flexed position, and were significantly reduced when the arm muscles were relieved of the need to support the weight of the arm. We then proceeded to assess reaching and holding control in the same workspace (Experiment 2) and examined whether resting postural forces could partially account for deficits in the motor control of reaching and holding still in the same patients. Remarkably, these resting postural force biases did not seem to have a detectable effect upon any component of active reaching but only emerged during the control of holding still after the movement ended. These results suggest a clear dissociation between the control of movement and posture. At the same time, assessing patients' impairment using the FM-UE, a metric designed to measure for abnormal synergies during 3D arm movements, revealed a strong association between resting force biases and abnormal movement synergies; which raises the possibility that the observed dissociation of movement and posture control for planar weight-supported movements may break down for unsupported 3D arm movements. This dissociation raises interesting questions about both the neural architecture supporting it and approaches to rehabilitation of the post-stroke arm.

Dissociation between reaching and holding still

Previous research provides evidence for a separate control of reaching and holding in the healthy arm – for a review, see (Shadmehr 2017) and (Jayasinghe, Scheidt, and Sainburg 2022). For example, following visuomotor rotation training specific to the move phase using an out-and-back movement, participants did not transfer this rotation to holding still after a point-to-point reach in the same direction: after the movement was over, and visual feedback was removed, participants' held positions drifted from the rotated movement endpoint towards the baseline hold position (Scheidt and Ghez 2007). There is also neurophysiological evidence for separate control of moving and holding still. A previous study found populations of neurons in the macaque M1 that represent mechanical loads during posture or movement but not both (Kurtzer, Herter, and Scott 2005) whereas another recorded both cortical neurons and spinal interneurons and found that they coded parameters related to movement vs. posture maintenance, respectively (Shalit et al. 2012).

The dissociation reported here between the control of reaching and holding still in the post-stroke arm is consistent with a recently proposed hybrid model, which posits distinct controllers for

363 reaching and for bringing the arm to a stop (Jayasinghe, Scheidt, and Sainburg 2022). Our finding
364 that resting posture control does not interact either with the initial reach or bringing the arm to a
365 stop extends this idea, suggesting *three* distinct controllers: one for the initial reach; another for
366 bringing the arm to a stop; and another for control of holding still at the endpoint after movement
367 is over. It should be noted, however, that having distinct neural circuits for reaching and holding
368 does not rule out interactions between the two. For example, in parallel to the control of saccadic
369 eye movements, the holding circuit may integrate information from the moving circuit; we recently
370 found evidence for this in both healthy participants and patients with hemiparesis (Albert et al.
371 2020).

372 *Potential causes of abnormal resting flexor biases*

373 How are resting postural biases generated? One possibility would be abnormally low thresholds
374 of the stretch reflex (Levin and Feldman 1994). Passively extending the elbow, even at low speeds
375 – something Experiment 1 did – can lead to increased muscle activity which may persist long after
376 the end of movement (Levin et al. 2000; Kanade-Mehta et al. 2022). Alternatively, Experiment 1
377 revealed a number of parallels between resting postural force biases and abnormal synergies: a
378 propensity for flexion, mitigation by arm weight support and, a significant correlation between
379 patients' resting postural force biases and their FM-UE scores – a measure designed to assess
380 the post-stroke abnormal synergies that manifest during active movement. These similarities raise
381 the possibility that post-stroke resting postural biases and movement synergies share a
382 generative mechanism. Consistent with this possibility is the finding that externally-imposed elbow
383 flexion led to (involuntary) shoulder flexion and external adduction in stroke patients but not
384 healthy controls, suggesting abnormal synergy patterns do not require active voluntary movement
385 to be expressed (Sangani et al. 2007)– also (Schmit and Rymer 2001).

386 It has been proposed that abnormal resting posture after brain injury can be ascribed to an
387 extrapyramidal system (i.e., other than the corticospinal tract, CST) (Denny-Brown 1964).
388 Consistent with this suggestion, multiple lines of evidence from animal models point towards
389 reticulospinal tract (RST) involvement in postural control (Takakusaki 2017; Deliagina et al. 2007;
390 Lacquaniti et al. 1997). Lesions of the monkey ponto-medullary reticular formation resulted in
391 abnormal postures; notably, these were characterized by trunk and limb flexion, elevated
392 shoulders, and arms held close to the body (Lawrence and Kuypers 1968) mirroring the flexor
393 posture pattern seen after stroke. Several studies in cats also suggest a postural role for the
394 reticulospinal tract. Reticular formation neurons respond to vestibular inputs such as head tilts or
395 whole-body tilts, in line with a role in adjusting posture against gravity (Bolton et al. 1992;

Pompeiano et al. 1984; Matsuyama and Drew 2000). Electrical and chemical stimulation of the reticular formation leads to modulations in muscle tone (Takakusaki et al. 2016), and reticulospinal neurons display tonic activity patterns related to postural adjustments that precede movement, rather than the movement itself (Schepens and Drew 2004).

There is also recent evidence that the RST is the descending system responsible for the generation of large forces during voluntary movement (Tapia et al. 2022; Glover and Baker 2022). Thus, the RST may control posture and force production in the upper limb. Upregulation of the RST has also been implicated in the generation of abnormal movement synergies after stroke. For example, the ipsilateral RST can facilitate flexors but suppress extensors, a pattern mirroring the flexor synergy (Davidson and Buford 2004; Davidson, Schieber, and Buford 2007).

There appears to be a contradiction, however, between the idea that abnormal postures share a common mechanism with abnormal movement synergies, and our finding that there was no evidence for intrusion of resting biases into any aspect of active reaching. A potential explanation for this apparent contradiction is that our experiments were conducted on a 2D surface with weight support. In contrast, the FM-UE scale, a synergy-based measure, is performed by patients in 3D without weight support. It has been shown that for 3D reaching, patients with chronic stroke express intrusive flexor synergies (Zackowski et al. 2004). Weight support – as was used in Experiment 2 – also reduces intrusion of flexor synergies for planar movements (Sukal, Ellis, and Dewald 2007) and improves planar kinematics (Beer et al. 2004; 2007). Earlier work in a weight-supported planar task found similar degrees of reaching abnormality whether movements were made in or out of synergy (Levin 1996); more recently, we showed that reaching dexterity can be dissociated from synergy intrusion when arm support is provided (Hadjiosif et al. 2022). Thus, while resting flexor biases and movement-related flexor synergies are both reduced by full weight support, synergies seem to be more so. This is not inconsistent with positing a shared substrate for the abnormalities at rest and during movement, but this substrate might play more of a role in holding still than in moving, and so the consequences of when it is damaged may be more apparent in a task that stresses holding still over movement. We suggest a putative framework below.

We posit that the motor system has separable functional modes for moving vs. holding still and that this is accomplished by differentially weighting the contributions of descending systems that are operative in both modes (Figure 9). The CST is weighted more towards fast and fractionated control during movement, the RST, in contrast, is weighted more towards slower postural control

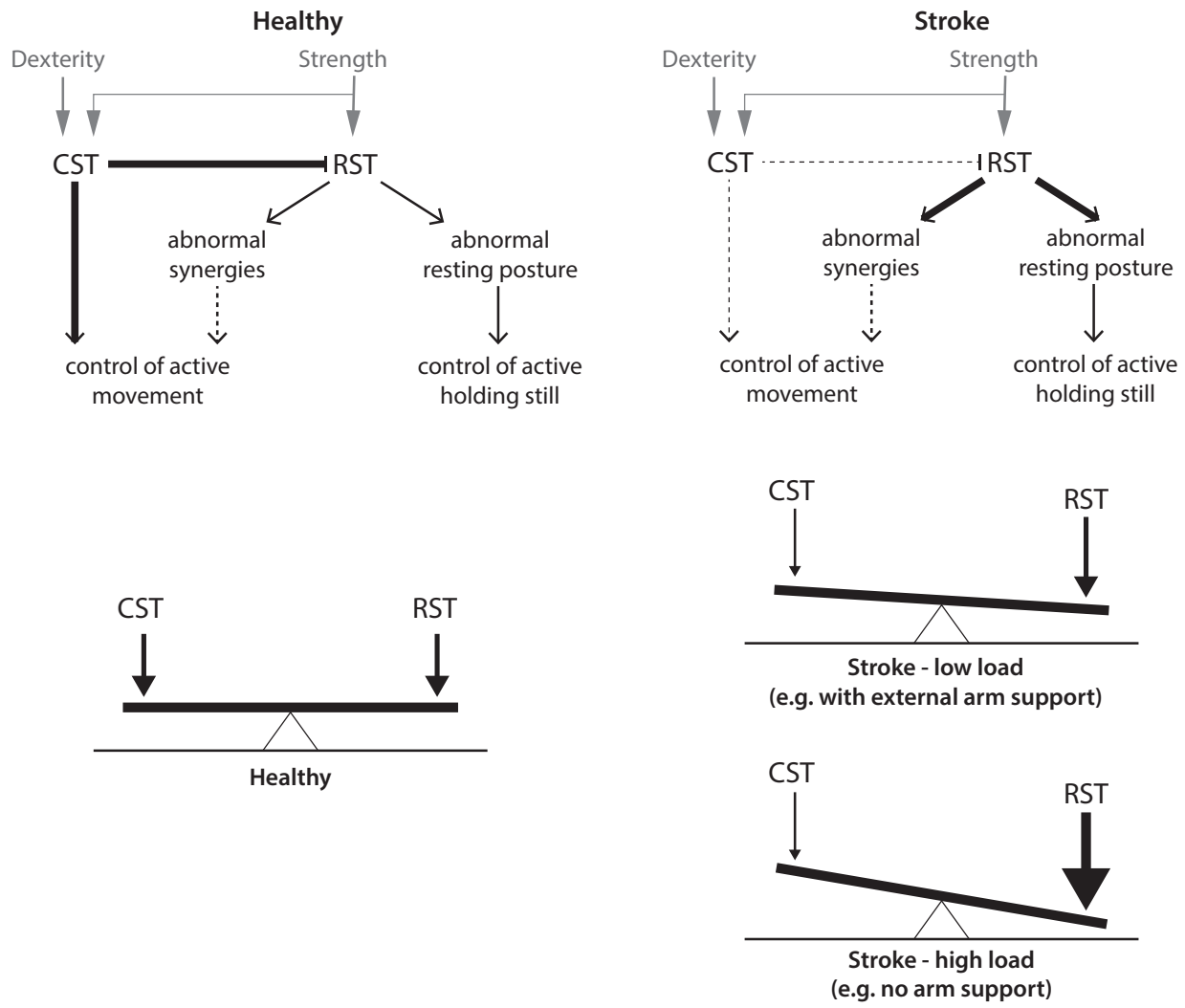

Figure 9: An architecture for the separable control of reaching and holding and spillover effects in stroke.

and generation of large isometric forces. In healthy individuals, these two modes are kept in balance by the CST, which has a moderating influence on the RST (Schepens and Drew 2006) (Figure 9, left). In this framework, the CST is the controller during movement and the modulator during holding still.

Damage to the CST after stroke reduces its moderating influence upon the RST (Figure 9, right); influence that is likely further compromised by upregulation of the RST through plasticity mechanisms (Zaaimi et al. 2012; García-Alías et al. 2015). This model can explain our results as follows: As the CST is the dominant system during movement it can still modulate the RST in this mode, especially with weight support, as this reduces RST drive, with the consequence that

resting biases do not markedly contaminate active movement. Conversely, the RST is the dominant system for postural control, and it can overcome CST modulation, which leads to the resting biases we observed. Thus, weight support allows a weakened CST to keep moving protected from abnormal holding, but it cannot prevent abnormal holding itself. The interesting implication of this model is that synergies are in fact spillover into movement of postural abnormalities when the CST can no longer modulate increased RST activation, for example in the case when weight support is removed.

A side question is how the control of decelerating to a stop – which we show here is distinct from the control of holding still after the movement is over – fits within such a scheme. A recent saccade study found that deceleration may be controlled through the cerebellum (Sedaghat-Nejad et al. 2022). In reaching tasks in the mouse, activity in the interpositus nucleus scaled with limb deceleration (Becker and Person 2019) whereas disruption of the pontine nuclei did not impair movement initiation as much as components related to bringing the movement to a stop such as success, endpoint position, and endpoint variance (Guo et al. 2021).

Increased recruitment of the RST has also been implicated in power grip (Baker and Perez 2017; Tazoe and Perez 2017). This may be relevant to our study, as our task had participants actively grasp the handle of the robot for both Experiments 1 and 2. If increased RST recruitment indeed explains abnormal resting postural biases as we discussed in the previous section, a stronger grasp would in turn increase the strength of these biases even further. Moreover, the intermixing of free-reaching and perturbation trials in Experiment 2 could have led to increased uncertainty in environmental dynamics; uncertainty can lead to adjustments such as even stronger grip (Hadjiosif and Smith 2015). Yet, despite this potential additional RST recruitment, we found that resting biases did not affect active reaching and only switched on after the reach was over; detected when we a applied a perturbing hand force. This result lends further support for separate controllers for reaching vs. holding still.

Conclusions

Our examination of the interplay between abnormalities in moving and holding still in patients making planar reaching movements suggests the existence of two functional modes of control, likely constructed out of differing configurations of the CST and the RST. The components of the paretic syndrome – loss of dexterity, weakness, abnormal resting posture, and intrusive synergies provide insight into how a normal movement is assembled by descending systems. To the degree that hemiparesis is a consequence of the CST losing, to varying degrees, both its direct control

472 over motor neurons and its modulatory influence on the RST, then physiological and behavioral
473 interventions may need to target these components separately.

Materials and Methods

Participants and Ethics Statement

A convenience sample of 16 stroke patients (age: 58.5±17.8 [average±standard deviation], 9 female) and 9 age-range matched healthy control participants (age: 62.6±15.2, 6 female) were recruited for this study. Table 1 shows details for each patient whereas Table 2 shows summary demographics and assessment metrics for patients and controls. Procedures were approved by the Johns Hopkins Institutional Review Board, and participants provided written informed consent.

Eligibility criteria

We recruited patients with hemiparesis due to stroke. To be eligible for the study, participants had to be adult, exhibit some movement with the affected arm, and be able to provide informed consent and understand the tasks involved. Exclusion criteria were marked cognitive impairment (assessed based on the Montreal Cognitive Assessment, MoCA, cutoff of 20); severe aphasia or ideomotor apraxia, neglect or hemianopia; and orthopedic or pain issues.

Task Details

Participants were seated on a chair and grasped the handle of a robotic arm; the handle moved either passively (by itself, Experiment 1) or actively (by the participant, Experiment 2) on the horizontal plane. In Experiment 2 and in two out of four conditions of Experiment 1, participants' lower arm was supported using a custom-made air-sled (Figure 1C). Above the plane was a screen which blocked direct vision of the arm; on this opaque screen we continuously projected a cursor indicating hand position (diameter: 3 mm), as well as the currently active target (diameter: 10 mm). Handle position was recorded at 200Hz, whereas subject-produced forces on the handle were recorder using a 6-axis force transducer. Experiments typically began with the paretic arm (see specific details below); for healthy controls, an arm was randomly assigned as primary, with its schedule matching that of the paretic arm in stroke patients.

Experiment 1: Measuring resting postural abnormalities

Following a previous paradigm (Simo et al. 2013; Laczko et al. 2017), Experiment 1 assessed resting postural forces by passively moving participants to different positions in the 2D workspace and holding them still in each position while it measured the forces they inadvertently exerted. Participants were instructed to maintain grasp on the robotic handle but otherwise rest and not resist the robot's motion as it slowly (5s movement time) moved from one position to the next and held them still (5s more). The array of positions (see Figure 1) could vary from one participant to the next and consisted of 18 - 22 positions (see exception below), whereas each position was

visited three times. During the passive moving and holding, a 3 mm white cursor indicating handle position and a 10 mm yellow disk indicating the destination of the passive movement were displayed. Though not essential from the participant's point of view, this allowed the experimenter to monitor the status of the experiment.

Each participant completed 4 blocks, two with each arm and in each arm support condition (air sled, no air sled); the typical order was {paretic, no support} -> {non-paretic, no support} -> {paretic, air sled support} -> {non-paretic, air sled support}; however in four individuals we completed the two paretic blocks first as their hand had to be secured to the handle (with self-adherent wrap) for a stable grasp.

In one patient, due to high resistance to passive moving and holding when air sled support was not provided, we were only able to measure resting postural forces in only 5 points under the no-air sled condition (the 5 points used to summarize the data).

Each block typically took less than 10 minutes to complete, with Experiment 1 lasting about 40 minutes including breaks.

Experiment 2: Assessing reaching control

In the same workspace as Experiment 1, Experiment 2 assessed motor control in a reaching task. Participants made 10-cm point-to-point reaches within an array of 5 targets (diameter: 10mm) within the workspace (Figure 4A), sampling 8 different movement directions. A white cursor (diameter: 3mm), indicating hand position, was visible throughout the experiment. Participants were instructed to try and stop at each target within a 600-800ms window after movement onset. At the end of movement, feedback was provided to indicate whether they were too fast (time <600ms, target turning red), too slow (time>800ms, target turning blue), or within the right time range (target "exploding" with a chirping sound).

The experiment was divided into blocks of 96 movements each (12 in each of the 8 movement directions). It began with three blocks with the paretic arm, followed by three more with the non-paretic arm, two more with the paretic arm, and ending with two blocks with the non-paretic arm. The first block with each arm was a familiarization block. Arm support (air sled) was provided throughout the experiment, and breaks were given between blocks.

Except for two participants, who performed Experiment 1 and Experiment 2 on different days due to limitations in their schedule, the entirety of each session – consisting of Experiment 1, Experiment 2 and standard assessments - took place on the same day and typically lasted about 3.5 hours with breaks given between the blocks as necessary.

Most trials (two-thirds) consisted of unperturbed movements to assess reaching control. In the remaining third of reaches, a 12N, 70 ms bell-shaped force pulse was applied by the robot lateral to the movement direction (Smith and Shadmehr 2005) after participants reached 2cm away from the starting position. On half of these trials the perturbation was oriented leftwards with respect to the movement (counterclockwise pulse) and the other half rightwards (clockwise pulse).

A fraction of trials in each block (20/96) imposed a static perturbation after movement in order to assess active holding control. For these trials, the holding time at the target was extended by 5 to 7 seconds, during which participants were instructed to hold still on the target (to remind them, the word “HOLD” were shown close to the target). During this extended hold period, a 6N force was gradually imposed over 2s in one of four different directions (45˚, 135˚, 225˚, 315˚), held constant for a pseudorandom time interval uniformly ranging from 3 to 5 seconds, and then abruptly released. Each block presented each position/static perturbation direction combination exactly once.

Data Analysis

Analysis was performed using MATLAB (Mathworks, Natick MA). For Experiment 1, we averaged resting postural forces within the latter 4s of the 5s passive holding period for each trial. To ensure forces were relatively stable and any movement minimal during the averaging window, the exact window could be further adjusted based on visual inspection of force and position traces. To obtain a measure of the average resting postural force at each position for each individual, we further averaged forces across the three visits to the same position. For comparisons, we focused on resting postural forces on the 5 positions shown in Figure 2; these forces were obtained directly (when the exact positions were sampled for the individual) or through interpolation (7/16 patients and 3/9 controls).

For Experiment 2, movement onset was defined as the moment in which participants’ velocity away from the starting position exceeded 3.5 cm/s, whereas movement end was defined as the moment the participant was within the target and moving at a speed of less than 3.5 cm/s. Initial reaching angles were calculated between movement onset and 150ms later; endpoint reaching angles were calculated between the moment the participant first reached within 2cm of the target and 150ms later. In pulse trials, settling time was defined as the time taken from pulse onset to the first moment absolute lateral velocity dipped below 2cm/s and remained below that amount for at least 100ms (or the movement ended). In static perturbation trials, settling time was defined as the time taken from perturbation onset (release of holding force) to the moment when velocity

570 dipped below 2cm/s (and remained below that amount for at least 100ms) and the distance from
571 the target was less than 2cm.

572 *Data exclusion criteria*

573 In Experiment 1, some trials (1.49%) were flagged as erroneous after visual inspection of force
574 profiles, in cases where forces appeared unstable and/or substantial movement was detected
575 during the period in which the robot was to hold the handle still at each measurement position.

576 In Experiment 2, we excluded as outliers movements in which initial movement direction (150ms
577 after movement onset) was ≥90° away from target direction. This excluded 0.95% of patients'
578 movements and 0.33% of controls' movements.

579 Moreover, in some hold perturbation trials (Figure 7), patients took a long time to reach the
580 stabilization criterion described in the previous section; mistakenly, our setup limited its recording
581 time to only the first 2s after force release. The exact time to stabilization thus could not be
582 measured for these particular trials, so they had to be excluded from analysis. Though only
583 13.2±3.3% (mean±SEM) of paretic stabilization trials were thus excluded in the patient population
584 (1.4±0.4% in their non-paretic side, 0.4±0.4% [2 trials] in controls), there were three patients for
585 which excluded trials were 25% or more of all paretic trials. To ensure there are no systematic
586 effects of this issue, we repeated the analysis of Figure 7G (a) by excluding these three patients
587 altogether or (b) by assigning a value of 2.0 seconds to the affected trials. In both cases, we found
588 results similar to our main analysis (Figure S1).

589 *Statistical comparisons*

590 In Experiment 1, we used an ANOVA to investigate any effect of conditions {Position
591 (distal/proximal), Support (with/without air sled), FM-UE (continuous)} and their interactions. We
592 used paired t-tests for the within-subject comparisons in Experiment 2.

593 *Fugl-Meyer assessments*

594 Assessments were separately scored by AMH and KK with scores subsequently averaged (hence
595 some scores having decimal values). For cases of substantial score differences (3 points or more)
596 scores were again reviewed by both raters together.

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599 Sheikh Khalifa Stroke Institute.

ID	Age (5y range)	Sex	Time since stroke	Handed-ness	Paretic arm	FM-UE (/66)	ARAT (/57)
S001	76-80	M	2y	Right	Left	57.5	57
S002	51-55	M	6y	Right	Left	40	47.5
S003	66-70	F	7y	Right	Right	34.5	19
S004	26-30	F	5y	Right	Left	55.5	43.5
S005	76-80	M	13mo	Right	Right	43.5	34
S007	51-55	F	2mo	Left	Right	63	57
S008	51-55	F	14mo	Right	Left	41	25
S009	56-60	F	5y	Right	Left	22	3
S010	66-70	M	5y	Right	Left	20	12
S011	41-45	F	20mo	Right	Right	64	57
S012	46-50	M	6y	Right	Left	18.5	6.5
S013	66-70	M	9y	Right	Left	14	8
S014	41-45	F	16mo	Right	Left	40	39.5
S015	61-65	F	10y	Right	Left	22	4.5
S016	36-40	F	21mo	Amb.	Right	62.5	57
S017	46-50	M	3mo	Right	Left	15	3

Table 1. *Patient characteristics*. FM-UE: Fugl-Meyer Assessment for the Upper Extremity; ARAT: Action Research Arm Test.

	Stroke patients	**Controls**
N	16	9
Age	58.5±17.8	62.6±15.2
Gender	7M/9F	3M/6F
Paretic side	11L/5R	n/a
FM-UE	38.3±18.2	66.0±0.0
ARAT	29.6±21.8	57.0±0.0
MoCA	24.9±3.1	28.1±1.6
Time since stroke	[2mo,10y]	n/a

Table 2. *Summary of patient and control characteristics*. FM-UE: Fugl-Meyer Assessment for the Upper Extremity (/66); ARAT: Action Research Arm Test (/57). MoCA: Montreal Cognitive Assessment (/30).). Here, ± indicates standard deviation.

Bibliography

Albert, Scott T, Alkis M Hadjiosif, Jihoon Jang, Andrew J Zimnik, Demetris S Soteropoulos, Stuart N Baker, Mark M Churchland, John W Krakauer, and Reza Shadmehr. 2020. "Postural Control of Arm and Fingers through Integration of Movement Commands." *Elife* 9.

Baker, Stuart N, and Monica A Perez. 2017. "Reticulospinal Contributions to Gross Hand Function after Human Spinal Cord Injury." *Journal of Neuroscience* 37 (40): 9778–84.

Becker, Matthew I, and Abigail L Person. 2019. "Cerebellar Control of Reach Kinematics for Endpoint Precision." *Neuron* 103 (2): 335–48.

Beer, Randall F, Julius PA Dewald, Michelle L Dawson, and W Zev Rymer. 2004. "Target-Dependent Differences between Free and Constrained Arm Movements in Chronic Hemiparesis." *Experimental Brain Research* 156 (4): 458–70.

Beer, Randall F, Michael D Ellis, Bradley G Holubar, and Julius PA Dewald. 2007. "Impact of Gravity Loading on Post-stroke Reaching and Its Relationship to Weakness." *Muscle & Nerve: Official Journal of the American Association of Electrodiagnostic Medicine* 36 (2): 242–50.

Bobath, B. 1982. "Adult Hemiplegia: Evaluation and Treatment, 2nd Edn. London: Spottiswoode Ballantyne." *Physiotherapy in Disorders of the Brain*.

Bolton, PS, T Goto, RH Schor, VJ Wilson, Y Yamagata, and BJ Yates. 1992. "Response of Pontomedullary Reticulospinal Neurons to Vestibular Stimuli in Vertical Planes. Role in Vertical Vestibulospinal Reflexes of the Decerebrate Cat." *Journal of Neurophysiology* 67 (3): 639–47.

Bourbonnais, Daniel, SHARYN VANDEN NOVEN, Kathleen M Carey, and William Z Rymer. 1989. "Abnormal Spatial Patterns of Elbow Muscle Activation in Hemiparetic Human Subjects." *Brain* 112 (1): 85–102.

Brunnstrom, Signe. 1966. "Motor Testing Procedures in Hemiplegia: Based on Sequential Recovery Stages." *Physical Therapy* 46 (4): 357–75.

Carr, EK, and FD Kenney. 1992. "Positioning of the Stroke Patient: A Review of the Literature." *International Journal of Nursing Studies* 29 (4): 355–69.

———. 1994. "Observing Seated Posture after Stroke: A Reliability Study." *Clinical Rehabilitation* 8 (4): 329–33.

Davidson, Adam G, and John A Buford. 2004. "Motor Outputs from the Primate Reticular Formation to Shoulder Muscles as Revealed by Stimulus-Triggered Averaging." *Journal of Neurophysiology* 92 (1): 83–95.

Davidson, Adam G, Marc H Schieber, and John A Buford. 2007. "Bilateral Spike-Triggered Average Effects in Arm and Shoulder Muscles from the Monkey Pontomedullary Reticular Formation." *Journal of Neuroscience* 27 (30): 8053–58.

Deliagina, Tatiana G, Pavel V Zelenin, Irina N Beloozerova, and Grigori N Orlovsky. 2007. "Nervous Mechanisms Controlling Body Posture." *Physiology & Behavior* 92 (1–2): 148–54.

Denny-Brown, D. 1964. "The Extrapyramidal System and Postural Mechanisms." *Clinical Pharmacology & Therapeutics* 5 (6part2): 812–27.

Fugl-Meyer, Axel R, L Jääskö, Ingegerd Leyman, Sigyn Olsson, and Solveig Steglind. 1975. "The Post-Stroke Hemiplegic Patient. 1. a Method for Evaluation of Physical Performance." *Scandinavian Journal of Rehabilitation Medicine* 7 (1): 13–31.

García-Alías, Guillermo, Kevin Truong, Prithvi K. Shah, Roland R. Roy, and V. Reggie Edgerton. 2015. "Plasticity of Subcortical Pathways Promote Recovery of Skilled Hand Function in Rats after Corticospinal and Rubrospinal Tract Injuries." *Experimental Neurology* 266 (April): 112–19. https://doi.org/10.1016/j.expneurol.2015.01.009.

Garland, S Jayne, Vicki L Gray, and Svetlana Knorr. 2009. "Muscle Activation Patterns and Postural Control Following Stroke." *Motor Control* 13 (4): 387–411.

Ghez, Claude, Robert Scheidt, and Hank Heijink. 2007. "Different Learned Coordinate Frames for Planning Trajectories and Final Positions in Reaching." *Journal of Neurophysiology* 98 (6): 3614–26. https://doi.org/10.1152/jn.00652.2007.

Glover, Isabel S., and Stuart N. Baker. 2022. "Both Corticospinal and Reticulospinal Tracts Control Force of Contraction." *The Journal of Neuroscience* 42 (15): 3150. https://doi.org/10.1523/JNEUROSCI.0627-21.2022.

Guo, Jian-Zhong, Britton A Sauerbrei, Jeremy D Cohen, Matteo Mischiati, Austin R Graves, Ferruccio Pisanello, Kristin M Branson, and Adam W Hantman. 2021. “Disrupting Cortico-Cerebellar Communication Impairs Dexterity.” *Elife* 10: e65906.

Hadjiosif, Alkis M, Meret Branscheidt, Manuel A Anaya, Keith D Runnalls, Jennifer Keller, Amy J Bastian, Pablo A Celnik, and John W Krakauer. 2022. “Dissociation between Abnormal Motor Synergies and Impaired Reaching Dexterity after Stroke.” *Journal of Neurophysiology* 127 (4): 856–68.

Hadjiosif, Alkis M., and Maurice A. Smith. 2015. “Flexible Control of Safety Margins for Action Based on Environmental Variability.” *Journal of Neuroscience* 35 (24): 9106–21. https://doi.org/10.1523/JNEUROSCI.1883-14.2015.

Hannaford, Blake, and Lawrence Stark. 1985. “Roles of the Elements of the Triphasic Control Signal.” *Experimental Neurology* 90 (3): 619–34. https://doi.org/10.1016/0014-4886(85)90160-8.

Jayasinghe, Shanie AL, Robert A Scheidt, and Robert L Sainburg. 2022. “Neural Control of Stopping and Stabilizing the Arm.” *Frontiers in Integrative Neuroscience* 16.

Kanade-Mehta, Priyanka, Maria Bengtson, Tina Stoeckmann, John McGuire, Claude Ghez, and Robert A Scheidt. 2022. “Spatial Mapping of Posture-Dependent Resistance to Passive Displacement of the Hypertonic Arm Post-Stroke.” *BioRxiv*, January, 2022.11.13.516311. https://doi.org/10.1101/2022.11.13.516311.

Karst, GM, and Z Hasan. 1991. “Timing and Magnitude of Electromyographic Activity for Two-Joint Arm Movements in Different Directions.” *Journal of Neurophysiology* 66 (5): 1594–1604.

Kurtzer, Isaac, Troy M Herter, and Stephen H Scott. 2005. “Random Change in Cortical Load Representation Suggests Distinct Control of Posture and Movement.” *Nature Neuroscience* 8 (4): 498–504.

Lacquaniti, F, C Maioli, NA Borghese, and L Bianchi. 1997. “Posture and Movement: Coordination and Control.” *Archives Italiennes de Biologie* 135 (4): 353–67.

Laczko, Jozsef, Robert A Scheidt, Lucia S Simo, and Davide Piovesan. 2017. “Inter-Joint Coordination Deficits Revealed in the Decomposition of Endpoint Jerk during Goal-Directed Arm Movement after Stroke.” *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 25 (7): 798–810.

Lawrence, Donald G, and Henricus GJM Kuypers. 1968. “The Functional Organization of the Motor System in the Monkey: II. The Effects of Lesions of the Descending Brain-Stem Pathways.” *Brain* 91 (1): 15–36.

Lestienne, F. 1979. “Effects of Inertial Load and Velocity on the Braking Process of Voluntary Limb Movements.” *Experimental Brain Research* 35 (3): 407–18.

Levin, Mindy F. 1996. “Interjoint Coordination during Pointing Movements Is Disrupted in Spastic Hemiparesis.” *Brain* 119 (1): 281–93.

Levin, Mindy F, and Anatol G Feldman. 1994. “The Role of Stretch Reflex Threshold Regulation in Normal and Impaired Motor Control.” *Brain Research* 657 (1–2): 23–30.

Levin, Mindy F, Ruud W Selles, Martine HG Verheul, and Onno G Meijer. 2000. “Deficits in the Coordination of Agonist and Antagonist Muscles in Stroke Patients: Implications for Normal Motor Control.” *Brain Research* 853 (2): 352–69.

Mani, Saandeep, Pratik K Mutha, Andrzej Przybyla, Kathleen Y Haaland, David C Good, and Robert L Sainburg. 2013. “Contralesional Motor Deficits after Unilateral Stroke Reflect Hemisphere-Specific Control Mechanisms.” *Brain* 136 (4): 1288–1303.

Matsuyama, Kiyoji, and Trevor Drew. 2000. “Vestibulospinal and Reticulospinal Neuronal Activity during Locomotion in the Intact Cat. II. Walking on an Inclined Plane.” *Journal of Neurophysiology* 84 (5): 2257–76.

Pompeiano, O, Diego Manzoni, UC Srivastava, and G Stampacchia. 1984. “Convergence and Interaction of Neck and Macular Vestibular Inputs on Reticulospinal Neurons.” *Neuroscience* 12 (1): 111–28.

Robinson, DA. 1970. “Oculomotor Unit Behavior in the Monkey.” *Journal of Neurophysiology* 33 (3): 393–403.

Sainburg, RL, C Ghez, and D Kalakanis. 1999. “Intersegmental Dynamics Are Controlled by Sequential Anticipatory, Error Correction, and Postural Mechanisms.” *Journal of Neurophysiology* 81 (3): 1045–56.

Sangani, Samir G, Andrew J Starsky, John R Mcguire, and Brian D Schmit. 2007. “Multijoint Reflexes of the Stroke Arm: Neural Coupling of the Elbow and Shoulder.” *Muscle & Nerve* 36 (5): 694–703.

Schaefer, Sydney Y, Kathleen Y Haaland, and Robert L Sainburg. 2009. “Hemispheric Specialization and Functional Impact of Ipsilesional Deficits in Movement Coordination and Accuracy.” *Neuropsychologia* 47 (13): 2953–66.

Scheidt, Robert A, and Claude Ghez. 2007. “Separate Adaptive Mechanisms for Controlling Trajectory and Final Position in Reaching.” *Journal of Neurophysiology* 98 (6): 3600–3613.

Schepens, Bénédicte, and Trevor Drew. 2004. “Independent and Convergent Signals from the Pontomedullary Reticular Formation Contribute to the Control of Posture and Movement during Reaching in the Cat.” *Journal of Neurophysiology* 92 (4): 2217–38.

———. 2006. “Descending Signals from the Pontomedullary Reticular Formation Are Bilateral, Asymmetric, and Gated during Reaching Movements in the Cat.” *Journal of Neurophysiology* 96 (5): 2229–52.

Schmit, Brian D, and W Zev Rymer. 2001. “Identification of Static and Dynamic Components of Reflex Sensitivity in Spastic Elbow Flexors Using a Muscle Activation Model.” *Annals of Biomedical Engineering* 29 (4): 330–39.

Sedaghat-Nejad, Ehsan, Jay S Pi, Paul Hage, Mohammad Amin Fakharian, and Reza Shadmehr. 2022. “Synchronous Spiking of Cerebellar Purkinje Cells during Control of Movements.” *Proceedings of the National Academy of Sciences* 119 (14): e2118954119.

Shadmehr, Reza. 2017. “Distinct Neural Circuits for Control of Movement vs. Holding Still.” *Journal of Neurophysiology* 117 (4): 1431–60.

Shalit, Uri, Nofya Zinger, Mati Joshua, and Yifat Prut. 2012. “Descending Systems Translate Transient Cortical Commands into a Sustained Muscle Activation Signal.” *Cerebral Cortex* 22 (8): 1904–14.

Simo, L. S., Piovesan, D., Botzer, L., M. Bengston, Ghez, C. P., and Scheidt, R.A. 2013. “Arm Kinematics during Blind and Visually Guided Movements in Hemiparetic Stroke Survivors.” In *Neuroscience Meeting Planner*, Program No. 652.06/VV4 2013. San Diego, CA: Society for Neuroscience.

Smith, Maurice A, and Reza Shadmehr. 2005. “Intact Ability to Learn Internal Models of Arm Dynamics in Huntington’s Disease but Not Cerebellar Degeneration.” *Journal of Neurophysiology* 93 (5): 2809–21.

Sukal, Theresa M, Michael D Ellis, and Julius PA Dewald. 2007. “Shoulder Abduction-Induced Reductions in Reaching Work Area Following Hemiparetic Stroke: Neuroscientific Implications.” *Experimental Brain Research* 183 (2): 215–23.

Takakusaki, Kaoru. 2017. “Functional Neuroanatomy for Posture and Gait Control.” *Journal of Movement Disorders* 10 (1): 1.

Takakusaki, Kaoru, Ryosuke Chiba, Tsukasa Nozu, and Toshikatsu Okumura. 2016. “Brainstem Control of Locomotion and Muscle Tone with Special Reference to the Role of the Mesopontine Tegmentum and Medullary Reticulospinal Systems.” *Journal of Neural Transmission* 123 (7): 695–729.

Tapia, Jesus A., Takamichi Tohyama, Annie Poll, and Stuart N. Baker. 2022. “The Existence of the StartReact Effect Implies Reticulospinal, Not Corticospinal, Inputs Dominate Drive to Motoneurons during Voluntary Movement.” *The Journal of Neuroscience* 42 (40): 7634. https://doi.org/10.1523/JNEUROSCI.2473-21.2022.

Tazoe, Toshiki, and Monica A Perez. 2017. “Cortical and Reticular Contributions to Human Precision and Power Grip.” *The Journal of Physiology* 595 (8): 2715–30.

Trombly, Catherine A. 1992. “Deficits of Reaching in Subjects with Left Hemiparesis: A Pilot Study.” *The American Journal of Occupational Therapy* 46 (10): 887–97.

Twitchell, Thomas E. 1951. “The Restoration of Motor Function Following Hemiplegia in Man.” *Brain* 74 (4): 443–80.

Zaaimi, Boubker, Steve A Edgley, Demetris S Soteropoulos, and Stuart N Baker. 2012. “Changes in Descending Motor Pathway Connectivity after Corticospinal Tract Lesion in Macaque Monkey.” *Brain* 135 (7): 2277–89.

Zackowski, Kathleen M, AW Dromerick, SA Sahrmann, WT Thach, and Amy J Bastian. 2004. “How Do Strength, Sensation, Spasticity and Joint Individuation Relate to the Reaching Deficits of People with Chronic Hemiparesis?” *Brain* 127 (5): 1035–46.