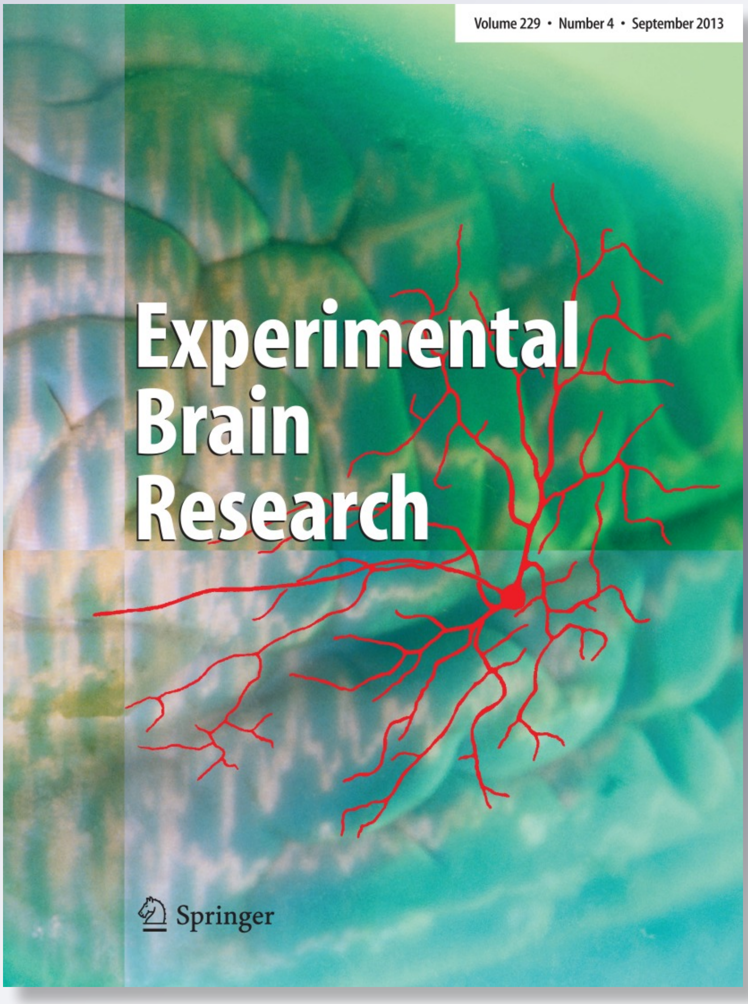
Rhythmic affects on stroke-induced joint synergies across a range of speeds

Matt Simkins, Anne Burleigh Jacobs & Jacob Rosen

Experimental Brain Research

ISSN 0014-4819
Volume 229
Number 4

Exp Brain Res (2013) 229:517-524
DOI 10.1007/s00221-013-3613-2



Exp Brain Res (2013) 229:517–524
DOI 10.1007/s00221-013-3613-2

RESEARCH ARTICLE

Rhythmic affects on stroke-induced joint synergies across a range of speeds

Matt Simkins · Anne Burleigh Jacobs · Jacob Rosen

Received: 1 April 2013 / Accepted: 9 June 2013 / Published online: 23 June 2013

Abstract Joint synergies are one among several diminished motor capabilities that are associated with stroke. These synergies are characterized by a stereotypical combination of involuntary joint coactivations. This research measured the synergistic rotations of the shoulder in response to voluntary rhythmic motion of the elbow across a range of speeds. The experimental protocol included a total of 22 subjects divided into two groups: (1) stroke survivors and (2) neurologically intact controls. Rhythmic motion in stroke survivors resulted in comparable synergies to discrete movement. It was found that hemiparetic subjects had greater synergy than neurologically intact individuals for all speeds. Synergy was quantified using a synergy ratio. This ratio uses elbow rotation as an input in the denominator and shoulder rotation as an output in the numerator. The amount of shoulder synergy varied with the subject's level of impairment as measured by a modified Fugl-Meyer assessment. As rhythmic speeds increased, the synergy ratios became higher for stroke subjects. This effect was especially pronounced for subjects with higher impairment. The relationships between synergies that arise from rhythmic and discrete movements are also discussed. The results of this study may have implications for therapeutic interventions, robotic rehabilitation approaches, and for the design of orthotic devices. More generally, these results shed light on the role of central pattern generators in hemiparetic motion.

M. Simkins (✉) · J. Rosen
Department of Computer Engineering, University of California, Santa Cruz, Santa Cruz, CA, USA
e-mail: msimkins@ucsc.edu

A. Burleigh Jacobs
SensoMotor Neurological Rehabilitation, LLC, Sunnyvale, CA, USA

Keywords Stroke · Synergy · Rhythmic · Speed · Central pattern generator · Rehabilitation

Introduction

Cerebral vascular accidents (CVA), i.e., strokes, can result in a multitude of impairments. Among these are diminished motor coordination and movement speeds. These effects are evident contralateral to the side of the damaged brain hemisphere (Halaney and Carey 1989). One particular aspect of uncoordinated movement is the emergence of joint synergies. Joint synergies resulting from a CVA are characterized by the involuntary rotation of joints in response to voluntary motion of other joints. These synergies have stereotypical features that are classically divided into flexor and extensor synergies (Brunnstrom 1970). Flexor synergy includes the mutual coupling between elbow flexion, shoulder abduction, shoulder extension, and shoulder external rotation (Duncan et al. 1983; Crocher et al. 2012).

Though stroke-induced joint synergies have been extensively researched, many of these evaluations involve discrete arm movements (Schaal et al. 2004). However, many activities of daily living, such as walking, washing, or exercising, involve rhythmic motion. To that end, researchers in the rehabilitation community have proposed rhythmic activities as a form of therapy (Pelton et al. 2010; Malcolm et al. 2009). The extent to which discrete movements generalizes to rhythmic motions is not well known. Indeed, rhythmic arm movements have been shown to suppress certain reflexes (Barzi and Zehr 2008). Furthermore, brain imaging using fMRI has shown that that rhythmic motions require substantially less brain activity than discrete motions (Schaal et al. 2004).

Central pattern generating networks (CPG) have been widely studied. CPG networks are believed to be spinal. They are responsible for a variety of rhythmic muscle contractions including locomotion, breathing, swimming, and swallowing to name a few. It is believed that they consist of two primary networks. The first is a rhythm-generating (RG) network. The second is a pattern formation (PF) network (Rybak et al. 2006). The RG network generates periodic bursts of neural activity for various time periods. In this sense, the RG is analogous to a clock that can produce a variable range of frequencies, as needed. The transition from slow to fast walking is one example of the need for frequency modulation. These periodic bursts are passed to the PF network. The PF then process the periodic burst activity for use with synergist and/or synergistic motor pools. Note, in much of the literature that discusses CPGs, "synergy" is often described as relating to coordinated, normal motion (Tresch and Jarc 2009). The term "synergy" should not be confused with the pathological synergies being discussed heretofore in this work. One explanation for the reduced brain activity associated with rhythmic motion, as compared to discrete motion, is that rhythmic motion requires a reduced set of high-level commands (Pearson 2000). In other words, the descending commands simply establish some level of activity in the CPG. The CPG then regulates rhythmic motion while making needed adjustments based on direct afferent feedback (Grillner 1981).

Beyond CPG considerations, there is an association between synergy and hypertonia, or spasticity (Welmer et al. 2006). Hypertonia is an increased excitation of the lower motor neurons resulting in resistance to muscle stretch. Spasticity is a velocity-dependent phenomenon. Spasticity in stroke survivors is believed to relate to a hyperactive stretch reflex. Previous research on synergy found velocity-dependent shoulder reflexes in response to passive joint rotations (Sangani et al. 2009). Some have proposed that synergies arise from reflexive interactions, including the stretch reflex in particular (McPherson et al. 2011). However, other work that is related to this research has shown that synergies are likely not caused by reflexive chains (Simkins et al. 2013b). Therefore, this analysis will assume that rhythmic synergies originate from CPGs and not exclusively from reflexes.

Synergies are evident for isometric muscle contractions (Dewald and Beer 2001; Dewald et al. 1995, 2001). However, synergies measured in this experiment are based on kinematic data of unconstrained arm movement. Volitional multi-joint arm movements can have complicated, time-varying angular joint velocities. For healthy individuals, it is difficult to precisely define a specific speed for discrete motions because they typically involve bell-shaped velocity profiles (Uno et al. 1989). These ambiguities are exacerbated for hemiparetic limbs which might move in a jerky, highly variable fashion (Kamper et al. 2002). One advantage of sinusoidal motion is that it overcomes the velocity ambiguities associated with discrete motion. Accordingly, synergy is calculated here using volitional, sinusoidal elbow rotation. Sinusoidal elbow rotation has the advantage of providing a smooth, continuous, and well-defined motion that is natural for subjects to maintain. Sinusoidal movement also allows for a simple and direct way to quantify peak angular velocities. Subjects were also asked to perform discrete arm motions. However, this was done for means of comparison, and discrete movements through a range of speeds are not considered.

Methods

Apparatus

A Vicon motion capture system (Vicon Motion Systems, Oxford, UK) recorded marker position. Subjects were seated on a metal chair. A strap was used to fixate the subject's torso to the chair (Cirstea and Levin 2000). Ten ceiling-mounted Vicon MX cameras were pointed at a target volume centered on the subject's arm. The camera sampling rates were 100 Hz. Marker position data had sub-millimeter resolution. Inverse kinematic calculations were accomplished using Vicon Bodybuilder™. Joint angle information was then processed using various Matlab™ scripts (Mathworks Inc., Natick, MA, USA). Fourteen motion capture markers were taped to the thorax and arm of subjects. Marker locations are depicted in Fig. 1.

Subjects

The experimental protocol included two groups. The first group (controls) included 11 neurologically intact males raging in age from 19 to 74. Group 2 included 11 hemiparetic subjects consisting of 6 females and 5 males, ranging in age from 54 to 82. All hemiparetic subjects experienced their CVA more than 6 months prior to participating in the experiment. A modified Fugl-Meyer (mFM) assessment for the upper limbs (Duncan et al. 1983) was conducted to determine the level of disability. No markers were attached to the fingers, and no data were collected that relates to the hand. The Fugl-Meyer was "modified" in that measures relating to the hand were excluded. The hemiparetic subject–screening process attempted to recruit subjects with as wide a range of impairments as possible. Scores from the mFM ranged from 0 (poor) to 14 (normal). The mFM helped to ensure that hemiparetic subject impairment was mild enough that the subjects could accomplish the protocol requirements yet severe enough that their synergies were detectable. Controls for the cause

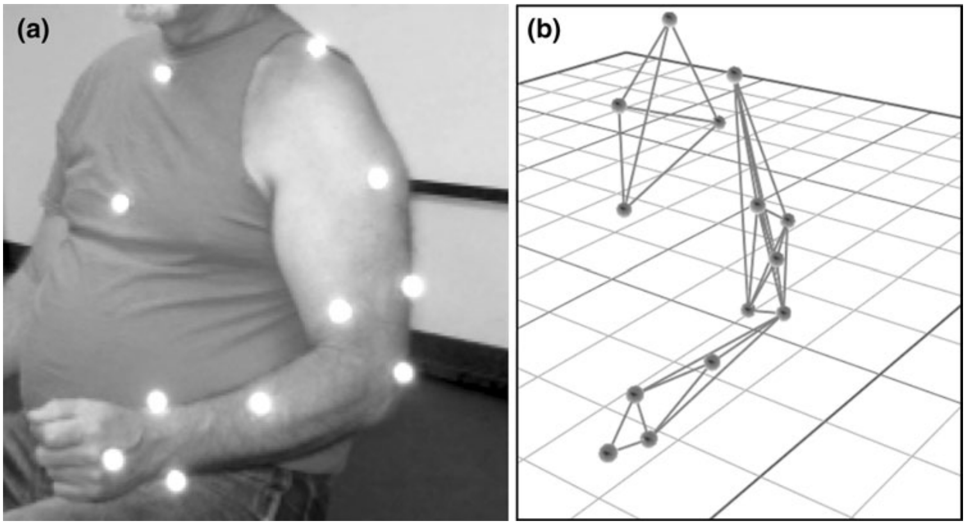


Fig. 1 Experimental setup. **a** Subject. Makers appear as *bright points*. **b** Subject-specific arm model. *Dots* represent marker locations. In most cases, thorax markers were attached to clothing. Shirt fabric was held firm to the skin with double-sided tape

of CVA (hemorrhagic or ischemic) were deemed unnecessary (Archambault et al. 1999). Likewise, there were no controls for gender. The research protocol was approved by the Internal Review Board of University of California, Santa Cruz. All subjects provided written consent prior to participation.

Protocol

A multitude of joints are often affected by synergy. These include the upper and lower limbs, as well as the thorax. This research focuses on shoulder synergy as a response to volitional elbow rotation. This synergy combination was selected because the shoulder tends to respond strongly to elbow rotations (Trumbower et al. 2010; Sangani et al. 2009). Scapular motion in the shoulder is not considered and the shoulder is modeled as a 3 degree-of-freedom joint with perpendicular, coincident axes that are centered on the glenohumeral joint. The axes being considered are parallel to shoulder abduction, flexion, and outer rotation. An axes passing through the medial and lateral epicondyle was used to model the elbow. Thus, the human arm is modeled as a 4 degree-of-freedom serial manipulator.

Table 1 Joint movements

Frequency	Amplitude	Repetitions
$f_1 = 0.33$ Hz	Start position to full flexion	3
$f_2 = 0.83$ Hz	Start position to full flexion	3
$f_3 = \max$	Start position to full flexion	3
$f_4 = \max$	Start position to half flexion	3
$f_5 = \max$	Start to minimum flexion, voluntary tremor	3
Discrete	Start to full flexion, one cycle	3
Discrete	Start to half flexion, one cycle	1

For control subjects, the "affected side" denotes the subject's dominant limb. Frequency f_3 and f_4 are defined as being just below and above the subject's bandwidth, and f_5 is defined as the subjects maximum possible frequency

All arm motions started and ended from approximately the same position as depicted in Fig. 1. All subjects were asked to flex their elbow from the start position to a flexed position. They then extended their elbow back to the start position, thus completing one cycle. For periodic motion, subjects were instructed to cycle their elbow in a smooth, continuous fashion such that the elbow rotation was approximately sinusoidal. All subjects were allowed to oscillate their arms at a given frequency for some time prior to recording the movement. Therefore, the resulting motion is regarded as being steady state. Data recordings included a minimum of 5 cycles. For discrete motion, subjects were asked to move a single joint "slowly and deliberately" from the start position to a flexed position and then back to the start position. To the best of their ability, all subjects were asked to avoid moving joints other than the one being intentionally moved. Amplitudes are summarized in Table 1.

The amplitudes of rotation were divided into "full," "half," and "voluntary tremor." Full elbow flexion amplitude was from the start position to a fully flexed position. The elbow was then extended back to the start position. It is in this sense that the term "full" rotation is used throughout this paper. For half flexion, subjects were asked to rotate their elbow through half the angle that was achieved for full flexion. Finally, for "voluntary tremor," subjects cycled their joint from the start position to the minimum achievable flexion angle and then back to the start position. Note, the term "tremor" used for this study should not be confused with the numerous types of known involuntary tremors.

Frequencies are summarized in Table 1. Frequencies, f_1 and f_2, were regulated by having the subjects sync their elbow cycles to the beat of a metronome (Thaut et al. 2002). Subjects were not allowed to see the metronome so that only audio stimulation was permitted. Metronome frequencies were determined through preliminary testing with a single stroke survivor. Frequency f_1 was selected because it was just fast enough that the resulting arm motions were adequately sinusoidal (smooth). Frequency f_2 was selected because it was just below the maximum frequency that the hemiparetic test subject could achieve for full elbow amplitudes (Molinari et al. 2005). For f_3–f_5, subjects were asked to oscillate their arm as fast as they could. Because such frequencies were not synced to a metronome, they tended to vary from subject to subject. For this reason f_3–f_5 are best thought of as self-selected frequencies/speeds. Frequency f_3 required subjects to oscillate their elbow as fast as possible through full amplitude. Frequency f_4 required subjects to flex their elbow through half the angle that was achieved for full flexion, or ≈50 %. Because the amplitude of rotation was reduced, subjects were able to achieve higher frequencies for f_4 than for f_3. Finally, f_5 required that subjects oscillate their elbow as fast as possible through the smallest possible angle. Because f_5 involved the smallest amplitude, f_5 frequencies were the highest for each subject. Taken together, the elbow "sweeps" through a range of frequencies from lowest to highest.

Data analysis

This study is intended to determine the frequency response of synergistic joints. This experiment used an approach that is analogous to frequency response methods used in engineering (Nise 2004). However, many of the data analysis techniques that are typically employed in engineering applications (such as those involving transfer functions) are regarded as poor models for the nervous system (Stark et al. 1961; Bennett et al. 1992). Therefore, the similarities between this approach and conventional engineering approaches relate more to the experimental design and not to the analysis.

Synergy is calculated using a linear model. The linear model treats the joint being intentionally rotated as the independent variable (the elbow). The involuntary synergistic response is treated as the dependent variable (the shoulder). The linear model is obtained using a least squares linear fit and is given by

$$\Theta_{\text{Shoulder}} = m * \Theta_{\text{Elbow Flexion}} + b \tag{1}$$

where Θ_{Shoulder} might represent shoulder flexion, abduction, or rotation. The constant b is in degrees and it equals the angle of the start position. The constant m is a unitless constant and it relates to the strength of the synergistic interaction. For example, at $m = 1.5$, for every degree the elbow flexes the shoulder will rotate by 1.5°. An alternative interpretation for m is as a ratio of angles. The numerator of m is the involuntary angular change of the shoulder. The denominator is the voluntary angular change of the elbow. Therefore, coefficient m is referred to as the synergy ratio.

The constant b in (1) relates to the start/end position. Because all subjects utilized the same start/end position, this number was approximately the same for all subjects. Therefore, b contains no meaningful information relating to joint synergy and is not discussed in detail.

Damping properties (passive stiffness) from connective tissues in the upper arm are expected to have an insignificant effect and are assumed to be velocity insensitive (Given et al. 1995). Because sinusoidal rotation has a non-constant velocity, velocity data are presented in terms of peak velocity. For sinusoidal rotation, the peak angular velocity is given by

$$\omega_{\max} = a \cdot f \tag{2}$$

where a is the amplitude in degrees and f is the frequency in Hz. Elbow and shoulder frequencies were determined by identifying dominant modes using discrete Fourier transforms. Hemiparetic subjects exhibited only one dominant mode. Therefore, non-synergistic muscular innervations, such as clonus, are believed not to have played a role (Agarwal and Gottlieb 1977). Amplitudes were calculated using the local maximum and minimum values of the respective periodic functions in the time domain. Average amplitudes were calculated using a minimum of 3 cycles for any given data set.

Results

Frequency response

Overall, hemiparetic subjects had smaller amplitudes than controls. The amplitudes of elbow rotation are plotted with 95 % confidence intervals (CI) against the 5 frequencies in Fig. 2. According to the protocol, the amplitudes for f_1, f_2,

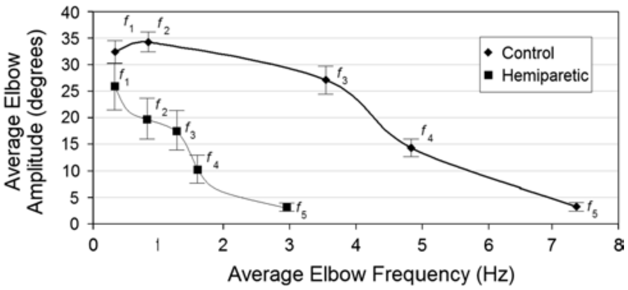


Fig. 2 Summary of frequency response for elbow for sinusoidal flexion extension. Depicted are 95 % CI. *Squares* and *diamonds* indicate mean values, $n = 33$

and f_3 should have been the same. However, subjects seem to have had changing amounts of overshoot/undershoot for the different frequencies.

As is evident from Fig. 2, hemiparetic subjects achieved significantly lower frequencies than controls when asked to oscillate their arms as fast as they could. These frequencies pertain to the self-selected speeds, f_3–f_5. Because the amplitude for f_4 was half, as much as the amplitudes for f_1–f_3, f_4 is analogous to the bandwidth frequency. The average f_4 frequency for healthy subjects (controls) was 3 times larger than for hemiparetic subjects, or 4.84 versus 1.62 Hz.

Though dynamic effects likely play a significant role for healthy subjects (control), they are not believed to play a significant role for hemiparetic subjects. Dynamic effects do not begin to dominate for cyclic elbow flexion until frequencies exceed around 5 Hz (Bennett et al. 1992). The fastest hemiparetic elbow frequencies achieved were around 3 Hz, see Fig. 2. Therefore, active muscle contractions are likely to dominate the observed kinematics for hemiparetic subjects at all frequencies.

Confounding effects of auditory stimulation

As was described previously, f_2 (0.83 Hz) was selected because it was near the anticipated upper limit for hemiparetic subjects. Frequency f_2 was actually too fast for Subject 5, 6, and 7. In the case of Subject 5, the subject chose to oscillate her arm as fast as she could even though she could not actually sync with the metronome. Interestingly, Subjects 6 and 7 were able to oscillate in sync with auditory cues from the metronome. However, when Subjects 6 and 7 were asked to oscillate their arms at maximum speed, the frequency was actually *less* than their metronome frequencies. Subject 6 and 7 achieved 0.83 Hz with the metronome but were only able to achieve 0.73 and 0.80 Hz, respectively, without the metronome. These effects have been observed elsewhere (Chen et al. 2006; Thaut et al. 2002), and this result may imply that the metronome had a confounding effect, albeit slight.

Synergistic response and involuntary joint rotation

The synergistic shoulder responses were entrained with the elbow regardless of frequency. Some have referred to relationships such as this as having a 1:1 frequency ratio (Swinnen et al. 1997). Therefore, it appears that shoulder rotations in hemiparetic subjects were only dominated by the synergistic response to elbow rotation.

One goal of this research was to investigate synergistic speed dependencies. For the self-selected speeds, f_3–f_5, the frequencies were increased by successively reducing amplitudes. Therefore, the synergistic response of the shoulder could depend on frequency *or* amplitude. In accordance with Table 1, full amplitude and half amplitude synergies were measured for discrete motions using approximately the same speeds. Student's 2-sample *t* tests were used to compare half amplitude and full amplitude synergies for all 3 axes of the shoulder. This resulted in *p* values ranging from 0.497 to 0.998. Therefore, there were no statistically significant differences in shoulder synergy between full and half-angle elbow rotations for hemiparetic subjects. This comparison suggests that the synergistic differences found for f_4 are comparable to f_1–f_3 in that the effects of amplitude are not significant. Note, such a comparison was not done for f_5. Notwithstanding, f_5 synergies exhibited somewhat unusual features that are discussed further on.

Not only did hemiparetic subjects with lower mFM scores (higher impairment) have more synergy, they also tended to have lower peak velocities. Equation (3) is the regression equation for peak velocities for the self-selected speeds, f_3–f_5, versus the hemiparetic mFM scores. The regression equation was,

$$\text{Velocity} = 3.77 + 1.50 * \text{mFM} \tag{3}$$

with indicator variable 1.50 having a *p* value <0.001. Therefore, for self-selected speeds, there was a significant, positive correlation between velocity and mFM scores. In other words, hemiparetic subjects with lower impairment (as measured by the mFM) were able to achieve higher velocities. Intuitively, it would seem that higher frequencies would correspond to higher elbow velocities. However, this was not the case. Recall from (2) that peak oscillatory angular velocity is equal to the product of amplitude and frequency. Because amplitude was reduced for the two highest frequencies, the peak speeds for f_4 and f_5 were actually less than the speeds for f_2 and f_3.

Synergy ratios are depicted in Fig. 3a–c for the 3 degrees of freedom of the shoulder in response to rhythmic elbow flexion and extension. Notice that the CI's are not ordered in terms of increasing frequency. Instead, they are ordered according to increasing speeds as calculated by (2). Not surprisingly, the control subjects in Fig. 3a had very low synergy. For this reason, the CI's depicted in Fig. 3a are somewhat trivial and are presented for purposes of comparison only. As was shown in (3), hemiparetic subjects with various levels of impairment achieved different self-selected speeds. Therefore, despite the fact that the CI's for f_3–f_5 fall into distinct categories in Fig. 3, these self-selected speeds actually relate to a range of frequencies. In order to distinguish lower and higher impairment groups, Fig. 3b depicts synergy for subjects with lower impairments and Fig. 3c depicts subjects with higher impairments. Only a statistical summary of hemiparetic subjects is provided in Fig. 3b, c. However, subjects tended to exhibit synergistic responses that spanned a continuum. In an effort to describe these differences schematically, the cartoon in Fig. 3d depicts this

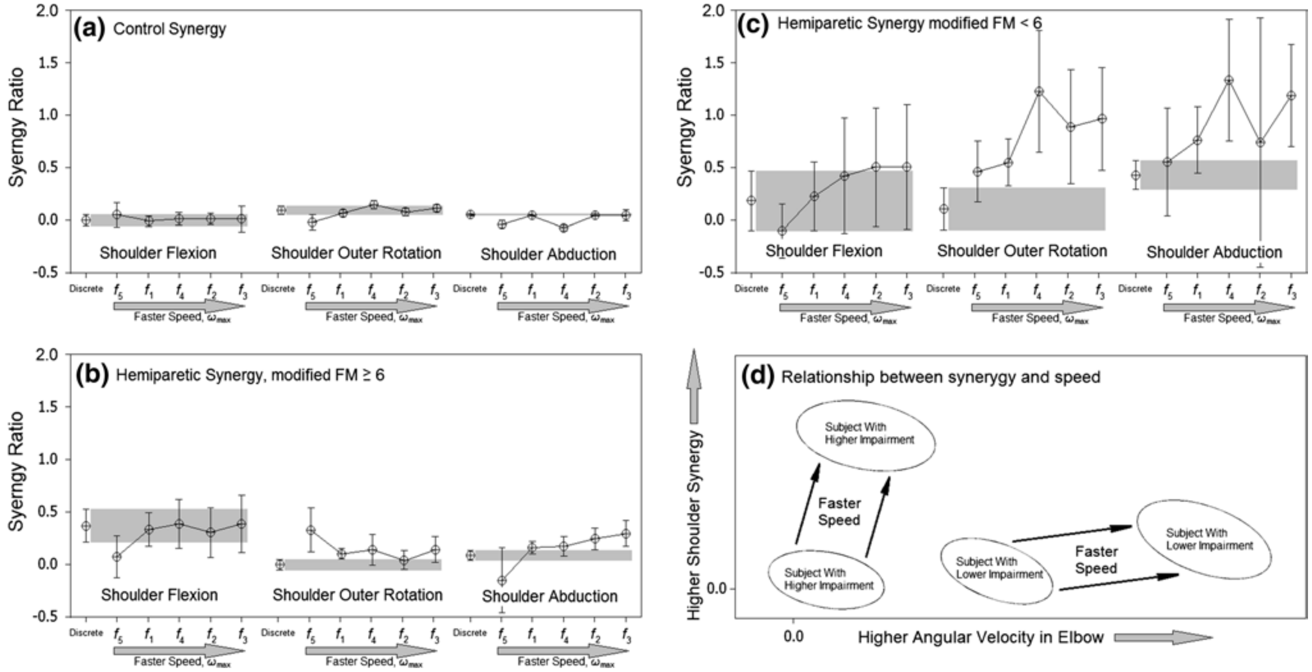


Fig. 3 Shoulder synergies ratios as a response to rhythmic elbow motion for all subjects. Data are summarized in **a**–**c** as 95 % CIs with mean connecting lines. Frequencies are ordered by increasing peak angular velocity. Depicted in **d** is a representational cartoon for 2 subjects

trend for 2 hypothetical subjects. The higher impairment subject in Fig. 3d exhibits relatively low synergy at the slower speeds. As subjects with higher impairment attempt to move their arm more quickly, their arm synergy increases rapidly. In contrast, the lower impairment subject in Fig. 3d achieves higher speeds with relatively low synergy. As subjects with low impairment move their arm still faster, the synergy increases only slightly.

Discrete synergy is plotted in Fig. 3a–c. For discrete motion, all subjects were simply told to move their arm “slowly and deliberately.” Based on visual inspection, discrete movements for hemiparetic subjects roughly matched the lowest metronome speed, f_1. In order to compare discrete and rhythmic motion, the mean synergy for discrete motion is plotted in Fig. 3 with a gray line that ranges the CI for discrete synergy and that extends to the right. For hemiparetic subjects with lower impairment in Fig. 3b, the f_1 synergy CI is within or near this gray band. Therefore, in these cases, it might be said that rhythmic synergies roughly match discrete synergies for cases of lower impairment. However, for subjects with higher levels of impairment in Fig. 3c, shoulder abduction and outer rotation synergies appear elevated for f_1 as compared to discrete synergy. Overall, the mean rhythmic synergies in Fig. 3c were greater than discrete synergy for every frequency except f_5, shoulder flexion.

In most cases, the synergies for rhythmic motion roughly match the synergies for discrete motion in that the synergies are in the same direction. However, some intervals had mean values that were in the opposite direction of discrete synergies. In particular, these include f_1 frequencies that are below the zero synergy line (negative synergy). Physically, this translates to rotations that are in the reverse direction. These include shoulder abduction in Fig. 3b and shoulder flexion in Fig. 3c. These data suggest that for some subjects, the shoulder flexion and abduction synergies switched direction for f_5. Given that f_5 translated to the lowest angular velocities, such a finding would suggest that for sufficiently low speeds, the direction of synergistic rotation might reverse. As was discussed previously, f_5 utilized tremor amplitudes. Earlier, it was shown that discrete full-angle synergy was comparable to half-angle synergy. However, such a comparison was not done for small amplitude synergies associated with f_5. Therefore, for tremors, minimal amplitude might have played a role and the speed dependency assumption for f_5 synergy is less certain.

It is worth noting that elbow flexion, a flexor synergy, resulted in shoulder flexion for both discrete and rhythmic motion. However, shoulder extension is classically described as belonging to the flexor synergies. Therefore, these results suggest that isolated elbow flexion contradict the stereotypical description of synergy. One explanation for this contradiction is that flexor synergies involve a multiplicity of joints, whereas this research only involves volitional movement of isolated joints. Confounding effects are inherent in multi-joint synergistic movements (Simkins et al. 2013a). Therefore, isolated joint interactions that violate the stereotypical description were expected.

Discussion

Arm speeds of hemiparetic subjects not regulated by a metronome were much lower than the arm speeds of neurologically intact subjects. This result is consistent with the known speed limitations of paretic arm movements. Hemiparetic subjects also had smaller rotational amplitudes. This was likely the result of range of motion limitations that are typical in the paretic arm. Not surprisingly, hemiparetic subjects had larger synergy ratios than neurologically intact individuals for all speeds. This was the case even for the least affected hemiparetic subject. These results compare well with the literature (Reisman and Sholz 2003).

Spasticity could explain the relationship between lower speeds and higher synergy. As was described previously, higher synergy is often concomitant with more severe spasticity. Some researches have suggested that spasticity is stronger for unloaded movements (Dietz et al. 1991). Indeed, hemiparetic arm speeds in this study were low enough that dynamic loading should not have been a significant factor.

Rhythmic synergy mostly matches the directions of discrete synergy in that elbow flexion was coupled with shoulder flexion, outer rotation, and abduction. The mFM uses discrete movements. Subjects who exhibited higher impairment on the mFM likewise exhibited elevated synergy for both discrete and rhythmic motion. Therefore, hemiparetic individuals with elevated synergy for discrete movements are likely to exhibit elevated synergy for rhythmic movements. Though rhythmic and discrete synergies showed some common features, there were differences too. At comparable speeds, rhythmic synergies were elevated for hemiparetic subjects. This was especially the case for subjects with higher impairment (mFM < 6). Fictive walking in mesencephalic cats is a classic example of CPGs. Indeed, in experiments involving mesencephalic cats, or cats under curarization, it was more easy to induced walking when treadmill stimulation was used to prime the CPGs (Grillner 1981). A possible explanation for the slightly higher rhythmic synergies in this work is that the CPGs were primed by the repeated oscillations associated with rhythmic motion.

The similarities between discrete and rhythmic motion in hemiparetic arms suggest that measures and therapies that target rhythmic synergy *might* generalize to discrete movement. As was described previously, discrete arm speeds were not tightly controlled. Therefore, based on these results, the extent to which synergies will change as arm speed changes for discrete movement is unclear. A more thorough comparison between rhythmic and discrete motion would likely require additional research that includes discrete movements across a range of speeds.

Othotic devices might benefit from these findings. There has been an ongoing effort to develop mobile, active orthotics for the upper and lower limbs (Dollar and Herr 2007). At the time of this writing, commercial efforts to produce such devices include the Tibion Bionic Leg (Tibion Corp., Sunnyvale, CA, USA) for the lower limbs and the mPower 1000 (Myomo Inc., Cambridge, MA, USA) for the upper limbs. If such devices are used to improve coordinated movement through a range of speeds, this research suggests that accommodations might be required for angular joint velocity. Likewise, rehabilitation robotics could benefit from these findings (Perry et al. 2007). Adaptive techniques that can optimize rehabilitation strategies over the course of a patient's recovery might result in improved outcomes (Squeri et al. 2011). For a robotic system to monitor and adjust to changes in synergy, (Dipietro et al. 2007) these results show that the type of movement, discrete or rhythmic, and the speed of that movement might affect the ways in which a robot measures patient progress.

These findings have neurological implications. As was described in the Introduction, CPGs likely consist of two stages. The RG generates a periodic burst, and the PF processes those bursts for use with a given motor pool. Given that CPGs are believed to generate rhythmic motion, it follows that rhythmic synergy is also generated by CPGs. Subjects were able to achieve a variety of frequencies, so in that respect, the hemiparetic RG appears functional. However, the maximum achievable hemiparetic frequencies for f_5 were less than half of what neurologically intact subjects could achieve. While it is tempting to attribute these lower achievable frequencies to reduced RG bandwidth, it is unclear if this effect is instead related to other factors such as spasticity.

Finally, the PF regulates motor pools. Therefore, *pattern formation networks within central pattern generators are potentially the source of hemiparetic synergies* for rhythmic motion. To the extent that an analogous process is at work for discrete motion, the PF is likewise implicated. Importantly, the PF is spinal, while a CVA, by definition, involves supraspinal damage. Therefore, the incorrect handling of motor pools by the PF is likely caused by faulty or incomplete descending commands (Cheung et al. 2009) from the brain.

Acknowledgments Foremost, we would like to thank all of the stroke survivors for their altruistic participation. Also appreciated is Professor Gabriel Elkaim of UCSC for his suggestions on the experimental design and to professor emeritus Nancy Byl at UCSF for her advise and insights on hemiplegia.

References

Agarwal G, Gottlieb G (1977) Oscillation of the human ankle joint in response to applied sinusoidal torque on the foot. J Physiol 268:151–176

Archambault P, Pigeon P, Feldman A, Levin M (1999) Recruitment and sequencing of different degrees of freedom during pointing movements involving the trunk in healthy and hemiparetic subjects. Exp Brain Res 126(1):55–67

Barzi Y, Zehr EP (2008) Rhythmic arm cycling suppresses hyperactive soleus h-reflex amplitude after stroke. Clin Neurophysiol 119:1443–1452

Bennett DJ, Hollerbach JM, Xu Y, Hunter IW (1992) Time-varying stiffness of the human elbow joint during cyclic voluntary movement. Exp Brain Res 88:433–442

Brunnstrom S (1970) Movement therapy in hemiplegia: a neurophysiological approach. Harper and Row, New York, pp 7–14

Chen J, Zatorre R, Penhune V (2006) Interactions between auditory and dorsal premotor cortex during synchronization to musical rhythms. Neuroimage 32:1771–1781

Cheung VCK, Piron L, Agostini M, Silvoni S, Turolla A, Bizzi E (2009) Stability of muscle synergy for voluntary actions after cortical stroke in humans. Proc Natl Acad Sci 106(46):19563–19568

Cirstea MC, Levin MF (2000) Compensatory strategies for reaching in stroke. Brain 123(5):940–953

Crocher V, Sahbani A, Robertson J, Roby-Brami A, Morel G (2012) Constraining upper limb synergies of hemiparetic patients using a robotic exoskeleton in the perspective of neuro-rehabilitation. IEEE Trans Neural Syst Rehabil Eng 20(3):247–257

Dewald J, Beer R (2001) Abnormal joint torque patterns in the paretic upper limb of subjects with hemiparesis. Muscle Nerve 24(2):273–283

Dewald J, Pope P, Given J, Buchanan T, Zev Rymer W (1995) Abnormal muscle co-activation patterns during isometric torque generation at the elbow and shoulder in hemiparetic subjects. Brain 118(2):495–510

Dewald J, Sheshadri V, Dawson M, Beer R (2001) Upper-limb discoordination in hemiparetic stroke: implications for neurorehabilitation. Top Stroke Rehabil 8(1):1–12

Dietz V, Trippel M, Berger W (1991) Reflex activity and muscle tone during elbow movements in patients with spastic paresis. Am Neurol Assoc 30:767–779

Dipietro L, Krebs H, Fasoli S, Volpe B, Stein J, Bever C, Hogan N (2007) Changing motor synergies in chronic stroke. J Neurophysiol 98(2):757–768

Dollar AM, Herr H (2007) Active orthoses for the lower-limbs: challenges and state of the art. International Conference on Rehabilitation Robotics, Noorwijk, Netherlands

Duncan PW, Propst M, Nelson SG (1983) Reliability of the Fugl-Meyer assessment of sensorimotor recovery following cerebrovascular accident. Phys Ther 63(10):1606–1610

Given JD, Dewald JP, Rymer WZ (1995) Joint dependent passive stiffness in paretic and contralateral limbs of spastic patients with hemiparetic stroke. J Neurol Neurosurg Psychiatr 59:271–279

Grillner S (1981) Control of locomotion in bipeds, tetrapods, and fish. In: Brooks VB (ed) Handbook of physiology, sec 1, vol 2., Wiley, New York, pp 1179–1236

Halaney M, Carey J (1989) Tracking ability of hemiparetic and healthy subjects. Phys Ther 69(5):342–348

Kamper D, McKenna-Cole A, Kahn L, Reinkensmeyer D (2002) Alterations in reaching after stroke and their relation to movement direction and impairment severity. Arch Phys Med Rehabil 83:702–707

Malcolm MP, Massie C, Thaut M (2009) Rhythmic auditory—motor entrainment improves hemiparetic arm kinematics during reaching movements: a pilot study. Top Stroke Rehabil 16:69–79

McPherson JG, Stienen AHA, Drogos JM, Dewald J (2011) The relationship between the flexion synergy and stretch reflexes in individuals with chronic hemiparetic stroke. IEEE International Conference on Rehabilitation Robotics Rehab Week Zurich, ETH Zurich Science City, Switzerland

Molinari M, Leggio M, Filippini V, Gioia M, Cerasa A, Thaut M (2005) Senorimotor transduction of time information is preserved in subjects with cerebellar damage. Brain Res Bull 67(6):448–458

Nise NS (2004) Frequency response techniques. In: Control systems engineering, 4th edn, ch. 10, sec. 1. Wiley, New York, pp 591–597

Pearson KG (2000) Neural adaptation in the generation of rhythmic behavior. Annu Rev Physiol 62:723–753

Pelton TA, Johannsen L, Chen H, Wing AM (2010) Hemiparetic stepping to the beat: asymmetric response to metronome phase shift during treadmill gait. Neurorehabil Neural Repair 24(5):428–434

Perry J, Rosen J, Burns S (2007) Upper-limb powered exoskeleton design. IEEE/ASME Trans Mechatron 12(4):408–417

Reisman D, Sholz J (2003) Aspects of joint coordination are preserved during pointing in persons with post-stroke hemiparesis. Brain 126(11):2510–2527

Rybak IA, Shevtsova NA, Lafreniere-Roula M, McCrea DA (2006) Modelling spinal circuitry involved in locomotor pattern generation: insights from deletions during fictive locomotion. J Physiol 577(2):617–639

Sangani S, Starsky A, McGuire J, Schmit B (2009) Multijoint reflex response to constant-velocity movements of the stroke elbow. J Neurophysiol 102(3):1398–1410

Schaal S, Sternad D, Osu R, Kawato M (2004) Rhythmic arm movement is not discrete. Nat Neurosci 7(10):1136–1143

Simkins M, Byl N, Burleigh-Jacobs A, Rosen J (2013a) Stroke induced synergistic phase shifting: a case for parametric modeling of human movement (unpublished)

Simkins M, Refai A, Rosen J (2013b) Upper limb joint space modeling of stroke induced synergies using isolated and voluntary arm perturbations (unpublished)

Squeri V, Basteris A, Sanguineti V (2011) Adaptive regulation of assistance 'as needed' in robot-assisted motor skill learning and neuro-rehabilitation. IEEE International Conference on Rehabilitation Robotics Rehab Week Zurich, ETH Zurich Science City, Switzerland

Stark L, Iida M, Willis P (1961) Dynamic characteristics of the motor coordination system in man. Biophys J 1(4):279–300

Swinnen SP, Dounskaia N, Walter CB, Serrien DJ (1997) Preferred and induced coordination modes during the acquisition of bimanual movements with a 2:1 frequency ratio. J Exp Psychol Hum Percept Perform 23:1087–1110

Thaut MH, Kenyon GP, Hurt CP, McIntosh GC, Hoemberg V (2002) Kinetmatic optimization of spatiotemporal patterns in paretic arm training with stroke patients. Neuropsychologia 40:1073–1081

Tresch M, Jarc A (2009) The case for and against muscle synergies. Curr Opin Neurobiol 19(6):601–607

Trumbower RD, Ravichandran VJ, Krutky MA, Perreault EJ (2010) Contributions of altered stretch reflex coordination to arm impairments following stroke. J Neurophysiol 104(6):3612–3624

Uno Y, Kawato M, Suzuki R (1989) Formation and control of optimal trajectory in human multijoint arm movement. Biol Cybern 61:89–101

Welmer AK, Holmqvist LW, Sommerfeld D (2006) Hemiplegic limb synergies in stroke patients. Am J Phys Med Rehabil 85(2):112–119