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ABSTRACT.

Temporal structure reveals the potential adaptive strategies employed during upper extremity movements. The authors compared the temporal structure of upper extremity joints under 3 different reaching conditions: preferred speed, fast speed, and reaching with rhythmic auditory cues in 10 individuals poststroke. They also investigated the temporal structure of these 3 reaching conditions in 8 healthy controls to aid in the interpretation of the observed patterns in the poststroke cohort. Approximate entropy (ApEn) was used to measure the temporal structure of the upper extremity joints. ApEn was similar between conditions in controls. After stroke, ApEn was significantly higher for shoulder, elbow, and wrist both at fast speed and with rhythmic cues compared with preferred speed. ApEn at index finger was significantly higher only with rhythmic cues compared with preferred speed. The authors propose that practice reaching at faster speed and with rhythmic cues as a component of rehabilitation interventions may enhance adaptability after stroke.

Keywords: nonlinear, rhythm, stroke, temporal structure, upper extremity

Stroke is the leading cause of adult-onset disability in the United States (American Heart Association, 2016). Up to 85% of individuals with stroke exhibit upper extremity (UE) paresis immediately poststroke (Kwakkel & Kollen, 2007; Kwakkel, Kollen, van der Grond, & Prevo, 2003; Olsen, 1990). Recent advances in our understanding of the principles of neuroplasticity provide motivation for development of interventions enhancing UE function in stroke survivors (Jang et al., 2003; Liepert, Graef, Uhde, Leidner, & Weiller, 2000; Sawaki et al., 2008; Wolf et al., 2006). Regrettably, despite advanced rehabilitation approaches, residual UE impairments still

persist. Current clinical assessment tools do not sensitively quantify the underlying motor impairments, limiting the ability to detect clinically important change in response to rehabilitation interventions. Insights provided by contemporary analyses are needed to better detect motor impairments and develop potentially more targeted and effective rehabilitation interventions after stroke.

While clinical and biomechanical (kinematic) assessments of UE movements are typically used to characterize motor impairments, measures that quantify movement variability offer a unique advantage to assess motor adaptability during performance of everyday tasks (Stergiou & Decker, 2011). However, these approaches have received less attention in clinical motor assessment and rehabilitation studies. Variability in reaching movements, which is characterized by the ability to use the UE joints in multiple ways to complete everyday tasks, is evaluated sparingly on the items of standard clinical tests, such as the Fugl-Meyer (Fugl-Meyer, Jaasko, Leyman, Olsson, & Steglind, 1975) and Wolf motor function tests (Wolf, Catlin, Ellis, Archer, Morgan, & Piacentino, 2001). Biomechanical (kinematic) analysis typically focuses on linear measures (range of motion), which describe the magnitude of the variations around a central point (i.e., the mean; Newell, 1976). However, linear measures ignore the evolution of movement over time thus failing to characterize the moment-to-moment variations, which occur during a goal directed reaching task. Individuals with stroke may gain increased ability to perform daily tasks through compensatory strategies without primary motor deficits being addressed (Corti, 2012; Levin et al., 2009). This is problematic because compensatory strategies are less efficient and may cause pain and long-term orthopedic complications after stroke (Kitago et al., 2013).

The moment-to-moment variations in movement, are referred, to as the temporal structure of variability (Harbourne & Stergiou, 2009; Newell, & Slikfin, 1998), which provides a window to understand the multiple adaptive strategies employed to successfully perform everyday tasks. Practically, the temporal structure of variability is associated with redundancy of the motor system (Stergiou & Decker, 2011). Motor redundancy refers to having more degrees of freedom than necessary to solve a task, which yields multiple strategies to perform a given motor task (Bernstein, 1967). For instance, when reaching to objects at a given distance but varying heights, we employ multiple strategies constantly adjusting (adapting) our reaching patterns to find the optimal solutions. The kinematics of these reaches may appear similar, but the temporal structure of variability in UE movements reveals the moment-to-moment variations, which help us to characterize the redundancy of the motor system and understand the multiple strategies employed to successfully perform these tasks.

The functional role of the temporal structure of variability is to impart adaptability (Kamm, Thelen, & Jensen, 1990; Scholz, 1990). Dynamic systems theory views movement variability as a necessary component of a biological system, which enables us to select the most optimal movement pattern from the interactions among the systems' biomechanical, morphological, cognitive, affective, and environmental (or task)

components (Davids, Button, & Benett, 2008; Kelso, 1984; Scholz, 1990). Motivated by the dynamic systems theory, Stergiou et al. (2006) proposed an optimal variability model suggesting that a healthy motor system is associated with an optimal state of variability and this variability enables individuals to adapt their movement patterns. From their perspective, the principle of optimal variability is explained by an inverted U-shaped relationship between complexity and predictability (Stergiou et al., 2006). The healthy state is represented by an optimal amount of movement variability, which is characterized by high complexity in the temporal structure of variability. Complexity signifies the presence of optimal moment-to-moment variations in the movement produced by a healthy biological system and represents the underlying physiologic capability to adapt to constantly changing demands of everyday tasks (Harbourne & Stergiou, 2009; Stergiou et al., 2006; Vaillancourt & Newell, 2002). The healthy state lies in an intermediate region between maximum predictability and no predictability. A pathological system, which either exhibits maximum predictability (fixed or robotic type movement pattern) or no predictability (random or irregular movement pattern), lies outside the optimal state of variability.

Individuals with moderate stroke sequelae often exhibit atypical uncoordinated and rigid reaching patterns, which limit ability to perform isolated joint movements (Bobath, 1990; Cirstea & Levin, 2000). For instance, shoulder flexion is often accompanied by shoulder abduction, and elbow and wrist flexion (Bobath, 1990). These atypical reaching patterns reveal significantly reduced temporal structure of variability across UE joints, suggesting limited adaptability of reaching patterns, and thus are termed rigid (Lodha, Naik, Coombes, & Cauraugh, 2010; Sethi, Patterson, et al., 2013). The optimal variability model could thus explain reduced adaptability observed in these atypical UE movements in individuals with moderate motor impairments poststroke. Nonlinear measures such as approximate entropy (ApEn) can capture changes in the temporal structure of variability (Harbourne & Stergiou, 2009; Morrison & Newell, 2012; Newell & Corcos, 2003). In contrast to linear measures, such as the range of motion, that are estimated by using only the initial and end values, ApEn uses all the values throughout the range of movement to reveal changes in its temporal structure. Based on the optimal variability model (Stergiou et al., 2006), one of the consequential goals of neurological rehabilitation should be, then, to reduce the movement error (or improve movement accuracy) as measured by linear measures of variability and to improve the temporal structure of movement as measured by nonlinear measures of variability. These changes would suggest that individuals with stroke have found optimal adaptive and effective strategies to perform daily tasks.

In the quest to develop more effective intervention, it is essential to identify movement variables that can augment the temporal structure of variability and adaptability of UE movements. Based on studies using linear measures, speed and rhythm of movement are two simple movement variables that may improve the temporal structure of variability and adaptability of UE movements (DeJong et al., 2012; Malcolm, Massie, & Thaut, 2009; Thaut, Kenyon, Hurt, McIntosh, & Hoemberg, 2002). For

example, reaching at faster than preferred speed has been shown to significantly improve linear measures of UE movements including, reach duration, peak grasp aperture, aperturepath ratio, and reach trajectories (DeJong et al., 2012). Moreover, when moving fast, the performance of individuals poststroke became more similar to healthy controls moving at their preferred speed (DeJong et al., 2012). Apart from movement speed, movements produced with rhythmic auditory cuing in individuals with poststroke have been found to increase UE performance on linear measures including shoulder and elbow joint excursions, reduced compensatory trunk displacement, and straighter reaching trajectories compared to movements produced without rhythmic cues in individuals poststroke (Malcolm, Massie, & Thaut, 2009; Thaut, Kenyon, Hurt, McIntosh, & Hoemberg, 2002).

However, the combined effect of reaching at faster speed and with rhythmic auditory cues on temporal structure of variability of reaching movements has not been explored in either healthy controls or individuals with stroke. Therefore, the purpose of this pilot study was to investigate whether movement variables including speed and rhythm have potent effects that alter the temporal structure of variability of reaching movements in individuals with stroke within a single session. We hypothesized that individuals with stroke would exhibit significantly different ApEn values and range of motion for flexion–extension of the shoulder, elbow, wrist and index finger proximal interphalangeal (PIP) joints while reaching at a faster speed or with rhythmic auditory cues as compared with reaching at preferred movement speed. We also computed the variable error between the index finger and the target to assess a linear measure of reaching accuracy across conditions. We hypothesized that individuals with chronic stroke would not reveal significant differences in the variable error of the index finger location while reaching at a faster speed or with rhythmic auditory cues as compared with reaching at preferred speed. We also analyzed the effects of speed and rhythm on the temporal structure across UE joints and variable error of the index finger marker in healthy controls. These data were used to compare and interpret the findings of individuals with stroke relative to healthy controls.

TABLE 1. Participant demographics: Individuals with stroke

Participant	Gender	Age (years)	Affected side	UE_FM (max score = 66)	Lesion location	Months after CVA
1	Male	76	Left	41	Right middle cerebral artery	102
2	Male	62	Left	27	Right middle cerebral artery	118
3	Female	62	Left	27	Posterior periventricular white matter	19
4	Male	72	Right	38	Left lacunar infarct	24
5	Male	76	Right	53	Left middle cerebral artery	174
6	Male	77	Right	38	Left pontine infarct	34
7	Male	72	Left	30	Right middle cerebral artery	48
8	Male	66	Left	43	Right striatocapsular infarct	105
9	Male	70	Left	45	Right cerebellar stroke	14
10	Male	68	Right	31	Left middle cerebral artery	16
$M \pm SD$		67 ± 8.9		36.9 ± 7.9		53.3 ± 50.9

Note. UE_FM — Upper Extremity Fugl Meyer; CVA — Cerebrovascular accident.

Methods

Participants

Ten individuals with chronic stroke with a mean age of 67 § 8.9 years old were recruited from a larger randomized controlled clinical trial. Participants were included if they (a) were between the ages of 18 and 90 years old; (b) had experienced a single ischemic stroke at least 6 months prior, which was confirmed with MRI scans; (c) were able to extend two fingers and the thumb at least 10; (d) were able to elevate the UE in scapular plane (combination of flexion and abduction) at least 30 using at least 45 active elbow extension available; (d) were able to follow two-step commands; or (e) had no history of more than minor head trauma, subarachnoid hemorrhage, dementia or other neurologic disorder or dysfunction, drug or alcohol abuse, schizophrenia, serious medical illness, or refractory depression. Demographic characteristics and the motor severity scores (Fugl-Meyer UE subscale) of the participants poststroke are reported in Table 1. in addition, we studied a convenience sample of healthy, right-handed, controls (eight women and one man; M age D 57 § 6.49 years).

Procedures

Eligible participants provided written informed consent approved by the University of Florida Health Sciences Center Institutional Review Board and North Florida/South Georgia Veterans Health System Research and Development Committee. The participants were part of a larger clinical trial of UE motor rehabilitation. The data presented are from the preintervention measurement session. All participants performed reach-to-point movements, which were captured by a 3D motion capture system (Vicon 612/T40; Oxford Metrics, Oxford, England). Participants poststroke reached to a target with the more impaired UE. Healthy controls reached with their nondominant hand. Sixty-seven reflective markers were secured to various landmarks of the upper body. Marker placements were determined using the Plug-in-UE marker set defined by our laboratory (Patterson, Bishop, McGuirk, Sethi, & Richards, 2011). A consistent starting position was used for all trials (see Figure 1a). Participants were seated on an adjustable, backless bench without trunk restraint with the knees flexed 90 and the feet flat on the floor. The hands were placed palm down on a table in front of them and supported in 90 of elbow flexion by arm rests level with the table.

Once instructed participants reached for a mark, drawn at 80% arm's length, defined as the distance from the acromion process to the tip of the middle finger, at midline on a table directly in front of them (see Figure 1b). This distance has been referred to as the critical boundary for reaching (Mark et al., 1997); healthy individuals use only UE joints to reach for objects within this workspace whereas reaching for objects beyond this boundary requires anterior flexion of the trunk. All participants were tested under three conditions: (a) self-paced reaching at preferred speed (PREFERRED), (b) reaching as fast as possible (FAST), and (c) reaching with rhythmic auditory cues (RHYTHM) generated by a metronome matched to each participant's

preferred speed. All participants performed four trials in each condition, with the first serving as a practice trial. For the PREFERRED condition, participants were instructed to reach for the mark and return to the starting position at their comfortable or preferred speed. The start of each trial was cued with a "go" command. The trial ended upon return to the start position. For the FAST condition, participants were cued with a "go" command to reach for the target and return to the starting position as fast as possible. For the RHYTHM condition, participants synchronized their reaching movements to the metronome cues. To determine the metronome pace, each participant was instructed to practice reaching to the mark 5–7 times, during which the pace of the metronome was matched to his or her preferred speed. Thereafter, participants were instructed to start the reaching movement at one metronome beat, touch the mark at the next metronome beat, and return to the starting position at the third metronome. Participants performed this sequence continuously three times in a row, without waiting for a "go" command from the examiner on the second and third trials. Because reaching faster or with rhythmic auditory cues might have biased later reaches made at preferred speed, we did not control for the randomization of the order of the three reaching conditions. As we worried that the collection of only three trials of reaching might not be enough to extract variability data, we compared ApEn in shoulder, elbow, and wrist joints computed from 30 trials of reaching with ApEn of those same joints during the first three trials of those reaches in a small sample (n D 3) in both groups. The ApEn values across all joints were similar between the three and the 30 trials (Table 2). Therefore, to minimize participant burden all subsequent participants performed only four trials in each condition, with the first serving as a practice trial. Additionally, we have validated this approach with similar number of trials in previous studies (Sethi, Davis, McGuirk, Patterson, & Richards, 2013; Sethi, Patterson, et al., 2013).

FIGURE 1. (a) Starting position. (b) Ending position.
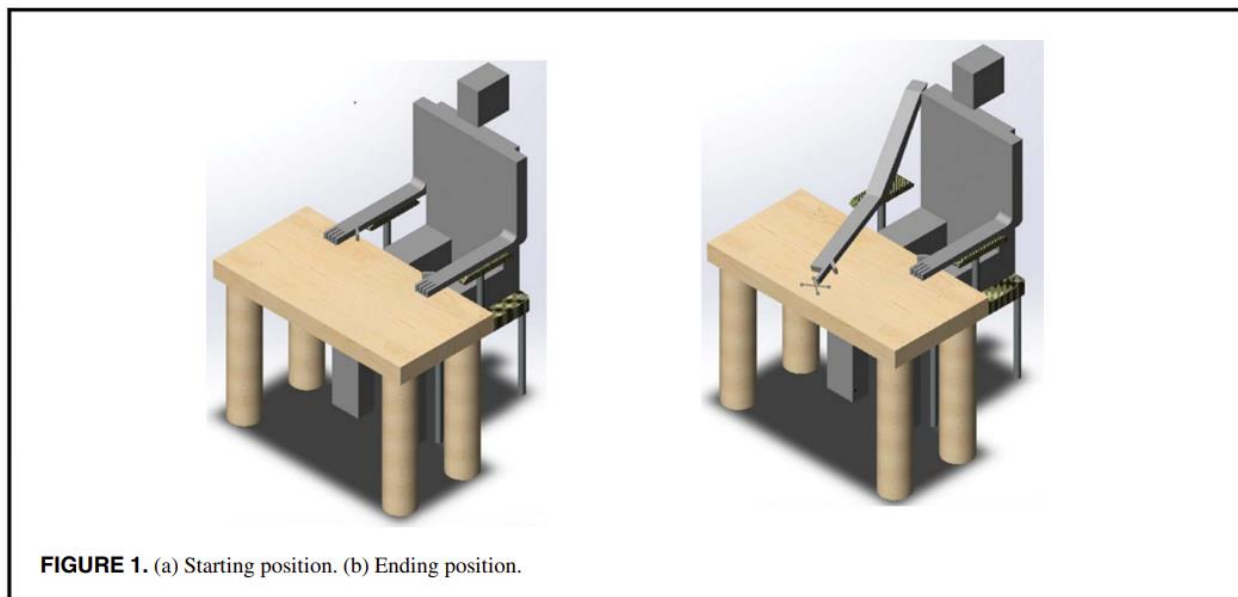
Data Analysis

Reaching kinematics were recorded using a 12-camera VICON motion capture system cameras (Vicon 612/T40; Oxford Metrics, Oxford, England). The data were collected using Vicon Nexus 1.5.2 software and sampled at 100 Hz. Data analyses were performed on the last three trials. Marker data were manually labeled and reconstructed using VICON software and modeled using SIMM 4.2 (MusculoGraphics, Inc, Santa Rosa, CA), to compute shoulder, elbow, wrist, and PIP (index finger) joint angles. The start of reach was identified as the point at which the velocity of the index finger marker exceeded 5% PV and the termination of reach as the point at which velocity of this marker fell below 5% PV. One degree of freedom in the sagittal plane (flexion–extension) was used to determine shoulder, elbow, wrist, and index finger PIP joint angle. To retain the inherent temporal structure of variability in the data, kinematic trajectories were not filtered (Rapp, Albano, Schmah, & Farwell, 1993). The use of unfiltered data when evaluating the temporal structure of variability is a standard approach to evaluate how movement variability changes over time and has been used in multiple studies (Kyvelidou, Harbourne, Shostrom, & Stergiou, 2010; Sethi, Patterson, et al., 2013; Stergiou, Moraiti, Giakas, Ristanis, & Georgoulis, 2004). The kinematic data described here resulted in four time series, one for each joint (shoulder, elbow, wrist, and index finger PIP).

Approximate Entropy

The temporal structure of variability of the shoulder, elbow, wrist and index finger PIP joint angle time series was determined by computing the ApEn using MATLAB (R2009a, the MathWorks, Natick, MA) code developed by Kaplan and Staffin (1996) using the algorithms provided by Pincus, Gladstone, and Ehrenkranz (1991). in the PREFERRED and FAST conditions, the time series for each joint angle was analyzed from the start of the reach through the entire length of the time series including the pauses between the three discrete trials. This approach was adopted because ApEn is effectively a measure of probability, developed to identify whether small patterns of a time series repeat later across the entire time series. These small patterns might not be repeated in a single cycle of reach-topoint movement. in previous studies, we have successfully used this approach to analyze ApEn in reach-to-grasp movements with the same number of trials in healthy controls and participants with stroke (Sethi, Patterson, et al., 2013). For the RHYTHM condition, the time series for each joint angle was analyzed from the start of the reach through the entire length of the time series of the continuous three trials.

The most common method employed in computation of ApEn is to identify repeating vectors of length m across the entire time series using the following equation:

$$\text{ApEn}(X \ m.r) = \log \left[C_{m,} \ (r)/C_{m+1,} \ (r) \right]$$

Table 2. Comparison of approximate entropy values for a subset of sample (n = 3) between three and 30 trials in each condition

	Healthy								Stroke							
	Shoulder		Elbow		Wrist		PIP		Shoulder		Elbow		Wrist		PIP	
	30	3	30	3	30	3	30	3	30	3	30	3	30	3	30	3
Preferred																
Trials	30	3	30	3	30	3	30	3	30	3	30	3	30	3	30	3
Median	0.13	0.15	0.18	0.16	0.35	0.38	1.2	1.1	0.05	0.06	0.07	0.06	0.16	0.15	0.48	0.5
Range	0.12–	0.13–	0.17–	0.15–	0.34–	0.35–	1.1–	1.07–	0.04–	0.05–	0.05–	0.05–	0.1–	0.1–	0.43–	0.46–
	0.14	0.16	0.22	0.2	0.39	0.4	1.3	1.2	0.06	0.06	0.08	0.07	0.21	0.2	0.52	0.5
Fast																
Trials	30	3	30	3	30	3	30	3	30	3	30	3	30	3	30	3
Median	.25	0.28	0.17	0.16	0.35	0.78	1.35	1.35	0.12	0.12	0.13	0.12	0.3	0.3	0.5	0.5
Range	0.24–	0.26–	0.17–	0.15–	0.34–	0.69–	1.29–	1.3–	0.1–	0.1–	0.1–	0.1–	0.25–	0.22–	0.48–	0.46–
	0.3	0.3	0.2	0.2	0.39	0.8	1.4	1.4	0.13	0.15	0.15	0.14	0.31	0.35	0.6	0.5
Rhythm																
Trials	30	3	30	3	30	3	30	3	30	3	30	3	30	3	30	3
Median	0.24	0.26	0.17	0.18	0.53	0.52	1.35	1.35	0.17	0.17	0.22	0.21	0.49	0.47	0.54	0.58
Range	0.23–	0.24–	0.17–	0.16–	0.52–	0.5–	1.29–	1.3–	.016–	0.17–	0.21–	0.21–	0.45–	0.45–	0.51–	0.53–
	0.25	0.26	0.2	0.2	0.54	0.55	1.36	1.4	0.19	0.18	0.23	0.24	0.5	0.47	0.6	0.6

Note. PIP = proximal interphalangeal.

Biomechanical data analysis conventionally utilizes r D 0.2 times the standard deviation of the time series, lag D 1 and m D 2 (Slifkin & Newell, 1999). Because the length of the time series could affect ApEn values, we normalized individual participants' ApEn values to the length of their time series and then multiplied the ratio (ApEn/length) with a constant equal to 100. Generally, a vector of shorter length repeats more often within a time series than a longer one, thus the lowest possible ApEn value can be the natural logarithm of 1, which is 0. ApEn values range from 0 to 2. in a highly periodic time series, values of Cm(r) can be similar to CmC1(r), such that ApEn is closer to 0. Hence, smaller values characterize a more regular time series where similar patterns are more likely to follow one another. in contrast, higher ApEn values suggest an irregular time series, where the predictability of subsequent patterns is low and ApEn could approach two (Stergiou, Buzzi, Kurz, & Heidel, 2004).

Surrogation Analysis

A surrogation procedure was also applied prior to computing ApEn utilizing Theiler, Eubank, Longtin, Galdrikian, and Farmer's (1992) procedure. Surrogation, performed prior to computing ApEn, is a critical step to verify whether fluctuations present in the kinematic data are deterministic in nature and not a source of noise. The surrogation procedure utilizes a phase randomization technique which removes the deterministic structure from the original shoulder, elbow, wrist, and PIP joint angle time series creating 20 surrogate time series of each trial with the same mean, variance, and power spectrum as the original time series. ApEn is then computed on both the original and each of the 20 surrogate time series. Significant differences in ApEn between the original and 19 of 20 surrogate time series could suggest that the original data are different than stochastic noise.

Variable Error Analysis

We computed variable error of the index finger marker to examine whether the index finger position upon contact with the target was different when reaching at a faster speed and/or with rhythmic auditory cues than preferred speed. Variable error was calculated in millimeters using the following equation:

$$\sqrt{\left(\sum \frac{(xi - M)^2}{N}\right)}$$ (Dounskaia, Wisleder, & Johnson, 2005)

where M is the average position of the index finger marker and N is the number of the reaches. We expected the variable error to be similar across the three conditions, suggesting that reaching at a faster speed and with auditory cues did not increase the error while reaching to the target.

Peak Velocity

Peak velocity (PV) was computed to confirm the differences in velocity between the three reaching conditions. PV was defined as the highest velocity during the entire reach trajectory and typically occurred at the transition from acceleration to deceleration.

Statistical Analysis

Data were checked for normality using the Shapiro-Wilk test. We employed nonparametric analyses using SPSS (Version 17.0) because the data did not meet the normality assumption (p   .05).

Determinism in Joint Angle Time Series Using Surrogation Analysis

Wilcoxon Signed- rank tests were performed to compare ApEn shoulder, elbow, wrist, and PIP (index finger) values between the original and surrogate time series.

Temporal Structure of Variability Between Healthy Controls and Participants With Stroke in Preferred Speed Condition

A Mann-Whitney U test was employed to compare ApEn shoulder, elbow, wrist, and PIP (index finger) in the PREFERRED condition between healthy controls and participants with stroke. This was performed to examine differences in the baseline condition (PREFERRED) between healthy controls and participants with stroke.

Temporal Structure of Variability Between Conditions in Healthy Controls and Between Conditions After Stroke

Wilcoxon signed-rank tests were also employed to differentiate ApEn shoulder, elbow, wrist, and PIP (index finger) between the FAST and PREFERRED conditions, and between RHYTHM and PREFERRED conditions within controls and within participants with stroke.

PV Between Conditions in Healthy Controls and Between Conditions After Stroke

Friedman's analysis of variance was conducted to examine the differences in PV across the three reaching conditions in healthy controls and participants with stroke. Post hoc analyses were conducted using the Wilcoxon signed-rank test to differentiate significant differences in PV between PREFERRED and FAST conditions and PREFERRED and RHYTHM conditions within controls and participants poststroke.

Variable Error Between Conditions in Healthy Controls and Between Conditions After Stroke

Wilcoxon signed-rank tests were also employed to differentiate variable error between the FAST and PREFERRED conditions, and between RHYTHM and PREFERRED conditions within controls and within participants with stroke.

For all analyses, ApEn, PV, and variable error data were analyzed with statistical significance set at $p < .05$. Holm's step-down correction procedure was used to correct

for multiple comparisons between conditions for ApEn, PV and variable error data (Holm, 1979).

Results

Determinism in Joint Angle Time Series Using Surrogation Analysis
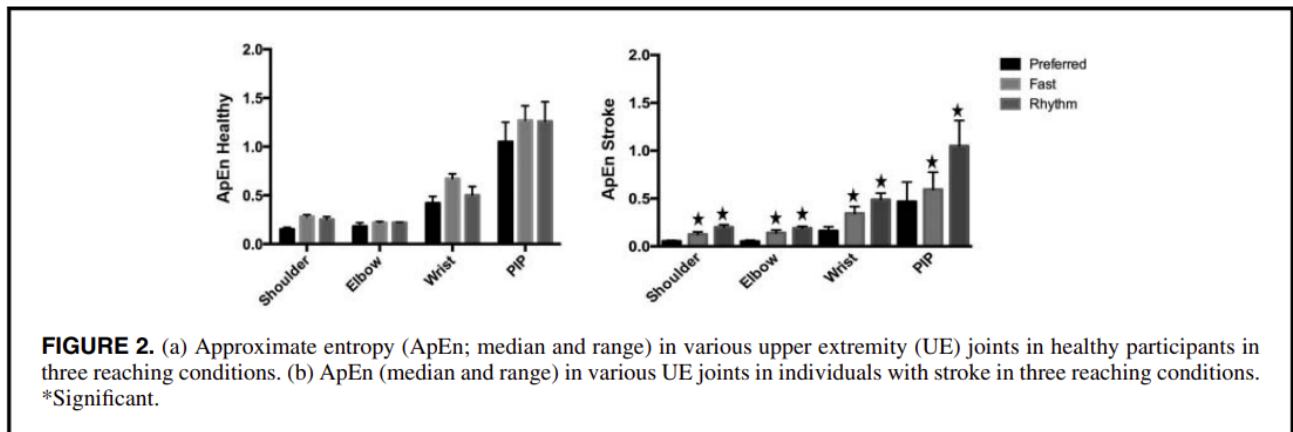
Differences between the original and the surrogated data were found in the joint angle time series in all three conditions for both control (p D .001) and stroke (p D .000) groups. These findings suggest that the fluctuations present in our data could be deterministic in nature (i.e., there is a certain structure) and not just random data.

Temporal Structure of Variability Between Conditions in Healthy Controls and Between Conditions After Stroke

Healthy control participants exhibited no significant differences in the ApEn values of the shoulder (p D .07), elbow (p D .10), wrist (p D .04), and PIP (p D .05) joints between the FAST and PREFERRED conditions (p D .10). Similarly, significant differences were not observed in the ApEn values of the shoulder (p D .09), elbow (p D .10), wrist (p D .07), and PIP (p D .05) joints between the RHYTHM and the PREFERRED conditions (p D .12; see Figure 2a). Although the statistical test values were < .05, these values did not reach significance after the Holm's step-down correction procedure.

For participants after stroke, ApEn at the shoulder (z D ¡3.18, p < .012), elbow (z D ¡3.18, p < .025), and wrist (z D ¡3.18, p < .016) was significantly greater in FAST as compared to the PREFERRED, condition. However, ApEn of the PIP joint did not differ (p > .05) between the FAST and the PREFERRED condition (see Figure 2b).

ApEn at the shoulder (z D ¡3.18, p < .016), elbow (z D ¡3.18, p < .05), wrist (z D ¡3.18, p < .012), and PIP joint (z D ¡2.51, p < .025) was significantly greater in the RHYTHM as compared with the PREFERRED condition (see Figure 2b).

FIGURE 2. (a) Approximate entropy (ApEn; median and range) in various upper extremity (UE) joints in healthy participants in three reaching conditions. (b) ApEn (median and range) in various UE joints in individuals with stroke in three reaching conditions. *Significant.

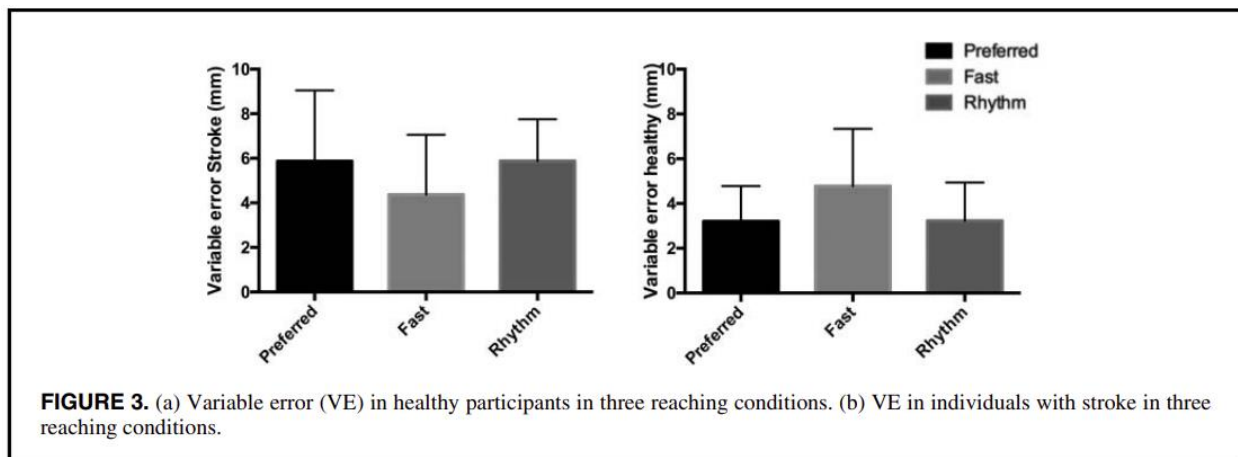### PV Between Conditions in Healthy Controls and Between Conditions After Stroke

Healthy participants revealed significantly different PV across the three conditions, $x2$ (2, N D 9) D 14.25, p < .0001. Healthy participants demonstrated significantly greater PV in FAST (Mdn D 1.2) as compared with PREFERRED condition (Mdn D 0.90; z D ¡2.52, p D .007). Healthy participants also showed significantly decreased PV in RHYTHM (Mdn D 0.77) as compared to PREFERRED condition (Mdn D 0.90; z D ¡2.52, p D .02).

Participants with stroke revealed significantly different PV across the three conditions, $x2$ (2, N D 10) D 18, p D .0001. Specifically, PV was greater in FAST (Mdn D 0.80) as compared with PREFERRED condition (Mdn D 0.50; z D ¡3.18, p D .002). Participants with stroke also showed significantly greater PV in RHYTHM (Mdn D 0.67) as compared with PREFERRED condition (Mdn D 0.50; z D ¡2.41, p D .01).

Variable Error Between Conditions in Healthy Controls and Between Conditions After Stroke

Variable error is a linear measure of variability and was computed to quantify the accuracy of reaches across conditions. Healthy control participants exhibited no significant differences in the variable error values of the index finger marker between the FAST and PREFERRED conditions (p D .08). Similarly, significant differences were not observed between the RHYTHM and the PREFERRED conditions (p D 1.06; see Figure 3a).

Participants with stroke exhibited no significant differences in the variable error values of the index finger marker between the FAST and PREFERRED conditions (p D .50). Similarly, significant differences were not observed between the RHYTHM and the PREFERRED conditions (p D .80; see Figure 3b).

FIGURE 3. (a) Variable error (VE) in healthy participants in three reaching conditions. (b) VE in individuals with stroke in three reaching conditions.

Discussion

The objective of this study was to investigate whether movement variables (i.e., speed and rhythm) have potent effects that alter the temporal structure of variability of reaching movements in individuals with stroke within a single session. The temporal

structure of variability was assessed using ApEn from the shoulder, elbow, wrist, and index finger proximal PIP flexion–extension joint angles. In our sample, individuals with stroke showed significantly lower values of the temporal structural variability (ApEn) across all joints at preferred speed as compared with the healthy controls. Both auditory rhythm cues and forced speed instructions increased the temporal structural variability of the reaching in individuals with stroke, suggesting that such variables may influence the temporal structural variability, potentially leading to greater movement adaptability. The lack of such effects on the temporal structural variability in healthy individuals suggests that when variability is optimal such cues have no marked effect on movement composition.

Our finding that reaching at a faster speed increased ApEn of the UE joints compared with reaching at preferred speed after stroke is consistent with observations by DeJong et al. (2012) in which linear measures of UE kinematics, including movement time, reach trajectories, reach path, and grasp aperture path ratios, increased when individuals with stroke reached at faster speeds. The underlying neurological mechanisms responsible for altering ApEn as a function of increased speed remain unclear. Investigation of this question is beyond the scope of the current study. However, reaching at faster speed might result in a more effective recruitment and activation of UE muscles due to the utilization of inertial forces poststroke (Wagner, Rhodes, & Patten, 2008).

We also discovered immediate effects of rhythmic auditory cues in altering ApEn in UE movements poststroke. Reaches with auditory cues demonstrated greater temporal structure of variability across all joints in individuals with stroke. One of the possible reasons for these effects with rhythmic auditory cueing could be attributed to the interaction of the auditory and motor systems. Auditory signals are known to raise the excitability of spinal motor neurons mediated by the auditory-motor circuitry at the reticulospinal level (Rossignol & Jones, 1976). Thaut et al. (2002) argued that rhythmic auditory cues allow the brain to map and scale smoother temporal variables of changes in position of the paretic arm throughout the entire movement cycle. Furthermore, external auditory cues provide a temporal constraint to the reaching movement (Thaut, Kenyon, Schauer, & McIntosh, 1999). Once the temporal constraint is added to the movement, the rhythmic cue may act as an external forcing function and serve to simplify the motor task of reaching between two targets. Moreover, the structured time information in rhythmic auditory cues alters the spatiotemporal characteristics of movement by entraining the timing of muscle activation patterns (Thaut et al., 1999). Entrainment of the timing of muscle activation patterns due to the rhythmic auditory cues might be one of the mechanisms contributing to changes in the temporal variability of reaching movements poststroke.

An alternative explanation for the increase in ApEn in the rhythmic auditory cueing condition is that PV of reaches in the auditory cueing condition was greater for individuals with stroke than in the preferred condition. Because our data show that

speed alone increased the temporal structural variability, we are not able to determine whether the increase in the temporal structure of variability resulted from the rhythmic auditory cueing or simply the increased speed of movement. Future researchers will need to investigate rhythm cueing at various speeds to decompose the potential effects of these two conditions. The increase in the temporal structure of variability found for individuals with stroke during fast and auditory rhythm conditions did not compromise the accuracy of reaching performance. Variable error of the index finger marker was not significantly different when reaching in these conditions for either the healthy controls or individuals with stroke compared to when reaching at preferred speed.

We acknowledge certain limitations of this study. We did not control for trunk displacement during reaching movements across conditions because we wanted to mimic natural reaching in which a trunk restraint would be atypical. Although the effect of trunk displacement may compromise the quality of reaching movements, we focused on the completion of the goal to touch the target with the affected hand. This can be explained by the contemporary motor control theories (dynamic systems theory), and rehabilitation interventions such as task-oriented approaches (Davis & Burton, 1991; Horak, 1991; Wolf et al., 2006) in which the emphasis to accomplish the goal is considered more important than the process of accomplishing the goal. We also did not control for the differences in age, gender, and hand-dominance between groups. Given the heterogeneity observed in persons poststroke, the sample size was relatively small. Future researchers can use the results of this pilot study to estimate the sample size for larger studies with more evenly matched participants to confirm these findings. Future researchers are also warranted to determine the critical speed and frequency of rhythmic auditory cues to optimize the temporal structure of variability of UE movements following stroke. In addition, the long-term effects and clinical feasibility of augmenting the intensity of interventions either by training the affected hand to reach at a faster speed or with rhythmic auditory cues in individuals with moderate impairments after stroke needs to be determined.

In summary, the present study demonstrated that reaching at either faster speed or with external rhythmic auditory cues alters a measure of the temporal structure of variability, without compromising the accuracy of the reaching movements in individuals poststroke. Although rehabilitation goals are not explicitly stated to address the temporal structure of variability, an implied expectation of therapists is to achieve and enhance functional movement ability. Importantly, these goals can be met through acquisition of adaptable movements that successfully respond to the changing demands of everyday tasks. This study provides empirical evidence that simple movement variables promote short-term gains in the adaptability of UE movements poststroke.

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References

American Heart Association. (2016). Heart disease and stroke statistics—update: a report from the American Heart Association. Circulation, 133, e168–e169.

Bernstein, N. A. (1967). The co-ordination and regulation of movements. Oxford, England: Pergamon Press.

Bobath, B. (1990). Adult hemiplegia: Evaluation and treatment (3rd ed.). Oxford, England: Butterworth-Heinemann.

Cirstea, M. C., & Levin, M. F. (2000). Compensatory strategies for reaching in stroke. Brain, 123, 940–953.

Corti, M., McGuirk, T. E., Wu, S. S., & Patten, C. (2012). Differential effects of power training versus functional task practice on compensation and restoration of arm function after stroke, Neurorehabilitation and Neural Repair, 26, 842–854.

Davids, K., Button, C., & Benett, S. (2008). Dynamics of skill acquisition: A constraints led approach. Champaign, IL: Human Kinetics.

Davis, W. E., & Burton, A. W. (1991). Ecological task analysis: Translating movement behavior theory into practice. Adapted Physical Activity Quarterly, 8, 154–177.

DeJong, S. L., Schaefer, S. Y., & Lang, C. E. (2012). Need for speed: Better movement quality during faster task performance after stroke. Neurorehabilitation and Neural Repair, 26, 362–373.

Dounskaia, N., Wisleder, D., & Johnson, T. (2005). Influence of biomechanical factors on substructure of pointing movements. Experimental Brain Research, 164, 505–516.

Fugl-Meyer, A. R., Jaasko, L., Leyman, I., Olsson, S., & Steglind, S. (1975). Poststroke hemiplegic patient. 1. Method for evaluation of physical performance. Scandinavian Journal of Rehabilitation Medicine, 7, 13–31.

Harbourne, R. T., & Stergiou, N. (2009). Movement variability and the use of nonlinear tools: Principles to guide physical therapist practice response. Physical Therapy, 89, 284–285. doi: 10.2522/ptj.20080130.ar

Holm, S. (1979). A simple sequentially rejective multiple test procedure. Scandinavian Journal of Statistics, 6, 65–70.

Horak, F. B. (1991). Assumptions underlying mototr control for neurologic rehabilitation. in M. J. Lister (Ed.), Contemporary management of motor control problems: Proceedings of the II Step Conference (pp. 11–27). Alexandria, VA: Foundations for Physical Therapy.

Jang, S. H., Kim, Y. H., Cho, S. H., Lee, J. H., Park, J. W., & Kwon, Y. H. (2003). Cortical reorganization induced by taskoriented training in chronic hemiplegic stroke patients. Neuroreport, 14, 137–141. doi: 10.1097/01.wnr.0000051544.96524. f2

Kamm, K., Thelen, E., & Jensen, J. L. (1990). A dynamical systems approach to motor development. Physical Therapy, 70, 763–775.

Kaplan, D., & Staffin, P. (1996). Software for heart rate variability. Retrieved from http://www.macalester.edu/»kaplan/hrv/ doc/

Kelso, J. A. S. (1984). Phase-transitions and critical-behavior in human bimanual coordination. American Journal of Physiology, 246, 1000–1004.

Kitago, T., Liang, J., Huang, V. S., Hayes, S., Simon, P., Tenteromano, L., ... Krakauer, J. W. (2013). Improvement after constraint-induced movement therapy: Recovery of normal motor control or task-specific comepnsation? Neurorehabilitation and Neural Repair, 27, 99–109.

Kwakkel, G., & Kollen, B. (2007). Predicting improvement in the upper paretic limb after stroke: A longitudinal prospective study. Restorative Neurology and Neuroscience, 25, 453–460.

Kwakkel, G., Kollen, B. J., van der Grond, J., & Prevo, A. J. (2003). Probability of regaining dexterity in the flaccid upper limb: Impact of severity of paresis and time since onset in acute stroke. Stroke, 34, 2181–2186.

Kyvelidou, A., Harbourne, R., Shostrom, V. K., & Stergiou, N. (2010). Reliability of center of pressure measures for assessing the development of sitting postural control in infants with or at risk of cerebral palsy. Archives of Physical Medicine & Rehabilitation, 91, 1593–1601.

Levin, M. F., Kleim, J. A., & Wolf, S. L. (2009). What do motor "recovery" and "compensation" mean in patients following stroke? Neurorehabilitation and Neural Repair, 23, 313–319.

Liepert, J., Graef, S., Uhde, I., Leidner, O., & Weiller, C. (2000). Training-induced changes of motor cortex representations in stroke patients. Acta Neurologica Scandinavica, 101, 321–326.

Lodha, N., Naik, S. K., Coombes, S. A., & Cauraugh, J. H. (2010). Force control and degree of motor impairments in chronic stroke. Clinical Neurophysiology, 121, 1952–1961.

Malcolm, M. P., Massie, C., & Thaut, M. (2009). Rhythmic auditory-motor entrainment improves hemiparetic arm kinematics during reaching movements: A pilot study. Topics in Stroke Rehabilitation, 16, 69–79.

Mark, L. S., Nemeth, K., Gardner, D., Dainoff, M. J., Paasche, J., Duffy, M., & Grandt, K. (1997). Postural dynamics and the preferred critical boundary for visually guided reaching. Journal of Experimental Psychology: Human Perception and Performance, 23, 1365–1379.

Morrison, S., & Newell, K. M. (2012). Aging, neuromuscular decline, and the change in physiological and behavioral complexity of upper-limb movement dynamics. Journal of Aging Research, 8, 912–18.

Newell, K. M. (1976). More on absolute error, etc. Jounral of Motor Behavior, 8, 139–142.

Newell, K. M., & Corcos, D. M. (2003). Variability and motor control. Champaign, IL: Human Kinetics.

Newell, K. M., & Slikfin, A. B. (1998). The nature of movement variability. in J. P. Piek (Ed.), Motor behavior and human skill: A multidisciplinary perspective (pp. 143–160). Champaign, IL: Human Kinetics.

Olsen, T. S. (1990). Arm and leg paresis as outcome predictors in stroke rehabilitation. Stroke, 21, 247–251.

Patterson, T. S., Bishop, M. D., McGuirk, T. E., Sethi, A., & Richards, L. G. (2011). Reliability of upper extremity kinematics while performing different tasks in individuals with stroke. Journal of Motor Behavior, 43, 121–130.

Pincus, S. M., Gladstone, I. M., & Ehrenkranz, R. A. (1991). A regularity statistic for medical data-analysis. Jounral of Clinical Monitoring, 7, 335–345.

Rapp, P. E., Albano, A. M., Schmah, T. I., & Farwell, L. A. (1993). Filtered noise can mimic low-dimensional chaotic attractors. Physical Review E., Statistical Physics, Plasmas Fluids and Related Interdisciplinary Topics, 47, 2289–2297.

Rossignol, S., & Jones, G. M. (1976). Audio-spinal influence in man studied by the H-reflex and its possible role on rhythmic movements synchronized to sound. Electroencephalography and Clinical Neurophysiology, 41, 83–92.

Sawaki, L., Butler, A. J., Leng, X., Wassenaar, P. A., Mohammad, Y. M., Blanton, S., ... Wittenberg, G. F. (2008). Constraintinduced movement therapy results in increased motor map area in participants 3 to 9 months after stroke. Neurorehabilitation and Neural Repair, 22, 505–513.

Scholz, J. P. (1990). Dynamic pattern theory–some implications for therapeutics. Physical Therapy, 70, 827–843.

Slifkin, A. B., & Newell, K. M. (1999). Noise, information transmission, and force variability. Journal of Experimental Psychology: Human Perception Performance, 25, 837– 851.

Sethi, A., Davis, S., McGuirk, T., Patterson, T. S., & Richards, L. G. (2013). Effect of intense functional task traiing upon temporal structure of variability of upper extremity poststroke. Journal of Hand Therapy, 26, 132–138.

Sethi, A., Patterson, T., McGuirk, T., Patten, C., Richards, L. G., & Stergiou, N. (2013). Temporal structure of variability decreases in upper extremity movements poststroke. Clinical Biomechanics, 28, 134–139.

Stergiou, N., Buzzi, U. G., Kurz, M. S., & Heidel, J. (2004). Nonlinear tools in human movement. in N. Stergiou (Ed.), Innovative analysis of human movement (pp. 63–90). Champaign, IL: Human Kinetics.

Stergiou, N., & Decker, L. M. (2011). Human movement variability, nonlinear dynamics, and pathology: Is there a connection? Human Movement Sciences, 30, 869–88.

Stergiou, N., Moraiti, C., Giakas, G., Ristanis, S., & Georgoulis, A. D. (2004). The effect of the walking speed on the stability of the anterior cruciate ligament deficient knee. Clinical Biomechanics, 19, 957–963.

Stergiou, N., Harbourne, R., & Cavanaugh, J. (2006). Optimal movement variability: A new theoretical perspective for neurologic physical therapy. Journal of Neurologic Physical Therapy, 30, 120–129.

Thaut, M. H., Kenyon, G. P., Hurt, C. P., McIntosh, G. C., & Hoemberg, V. (2002). Kinematic optimization of spatiotemporal patterns in paretic arm training with stroke patients. Neuropsychologia, 40, 1073–1081.

Thaut, M. H., Kenyon, G. P., Schauer, M. L., & McIntosh, G. C. (1999). The connection between rhythmicity and brain function. IEEE Engineering in Medicine and Biology Magazine, 18, 101– 108.

Theiler, J., Eubank, S., Longtin, A., Galdrikian, B., & Farmer, J. D. (1992). Testing for nonlinearity in time-series: the method of surrogate data. Physica D, 58, 77–94

Vaillancourt, D. E., & Newell, K. M. (2002). Changing complexity in human behavior and physiology through aging and disease. Neurobiology of Aging, 23, 1–11.

Wagner, J. M., Rhodes, J. A., & Patten, C. (2008). Reproducibility and minimal detectable change of three-dimensional kinematic analysis of reaching tasks in people with hemiparesis after stroke. Physical Therapy, 88, 652–663.

Wolf, S. L., Catlin, P. A., Ellis, M., Archer, A. L., Morgan, B., & Piacentino, A. (2001). Assessing Wolf motor function test as outcome measure for research in patients after stroke. Stroke, 32, 1635–9.

Wolf, S. L., Winstein, C. J., Miller, J. P., Taub, E., Uswatte, G., Morris, D., ... Nichols-Larsen, D. (2006). Effect of constraintinduced movement therapy on upper extremity function 3 to 9 months after stroke - The EXCITE randomized clinical trial. Jama, 296, 2095–2104.