Measurement 110 (2017) 84–89

Contents lists available at ScienceDirect

Measurement

journal homepage: www.elsevier.com/locate/measurement

Stroke-affected upper extremity movement assessment via continuous relative phase analysis

Kristina Daunoravičienė [a], Jurgita Žižienė [a], Jolanta Pauk [b,c], Adam Idzkowski [b], Inga Raudonytė [d,e], Alvydas Juocevičius [d,e], Artūras Linkel [a], Julius Griškevičius [a,*]

[a] Department of Biomechanics, Vilnius Gediminas Technical University, Vilnius, Lithuania
[b] Bialystok University of Technology, Bialystok, Poland
[c] Glenrose Rehabilitation Hospital, Edmonton, Canada
[d] Vilnius University Hospital Santariskiu Klinikos, Vilnius, Lithuania
[e] Faculty of Medicine, Vilnius University, Vilnius, Lithuania

ARTICLE INFO

Article history:
Received 20 January 2017
Received in revised form 6 June 2017
Accepted 8 June 2017
Available online 15 June 2017

Keywords:
Coordination
Coupling
Continuous relative phase
Phase angle
Upper extremity movement
Stroke
Biomechanics

ABSTRACT

Paper presents the quantified assessment of stroke-affected upper extremity (UE) coordination via continuous relative phase (CRP) analysis. 14 post-stroke patients were divided into 3 groups based on the severity of impairment according to Wolf Motor Function Test (WMFT). CRP was determined based on UE kinematics parameters measured using inertial measurement units fixed on arm, forearm and hand while subjects performed designated movement. UE movement cycles were analysed based on the metrics derived from phase planes, the phase angle and CRP plots, as well as the calculated range of motions and CRP variability rates. It was found that CRP variability is associated with impairment level, i.e. it is decreasing with a higher level of dysfunction. Therefore, the CRP might serve as measurable quantity and could be valuable for supporting clinical assessment and quantifying impairment severity of UE motor functions.

1. Introduction

Various motor dysfunctions caused by stroke or other neurodegenerative diseases negatively affect the quality of life. Sufferers of stroke demonstrate slower, less smooth, less efficient and less precise upper limb movements compared to non-affected persons [1–2]. In addition, stroke sufferers may have decreased coordination between shoulder flexion and elbow extension, and may use compensatory movements such as excessive trunk and shoulder movements [16]. It has been shown, that various stroke rehabilitation strategies like constraint-induced movement therapy [23] or robot-assisted rehabilitation [15] facilitate better recovery. It is common for rehabilitation specialists to evaluate the progress and effect of a rehabilitation strategy on motor function recovery. General examination of motor dysfunction includes assessment of strength, muscle tone, muscle bulk, coordination, abnormal movements and various reflexes. Many of these are better detected through simple observations. Wolf Motor Function Test (WMFT) is one of the tools regularly used in clinical practice for post-stroke upper extremity (UE) motor assessment and it provides clinician with the score on the ability of the patient to perform the motion (ranging from 0 – no motion to 5 – normal) [18,23–26]. Depending on the WMFT task, the time elapsed from the start to end is most frequently determined. However, there are several other kinematic characteristics of upper limb movements after a stroke that can be measured [1,5]. Kinematic analysis of upper and lower limb motion, as well as coordination, is usually performed while post-stroke individuals walk [4,20]. When evaluating WMFT, UE coordination is often only assessed visually, i.e. without instrumentation to measure kinematics. However, additional instrumentation and a quantitative assessment might facilitate improved diagnostics or more detailed assessment of the rehabilitation progress [14]. Due to its vast amount of muscles and joints the human body has multiple degrees of freedom that must be controlled in order to achieve goal-oriented movements. Such redundancy of actuators results in increased motor variability. Taking into account that human movement is a variable, kinematic

---

[*] Corresponding author.
E-mail address: julius.griskevicius@vgtu.lt (J. Griškevičius).

http://dx.doi.org/10.1016/j.measurement.2017.06.011
0263-2241/

data from isolated joints (e.g. angle, displacement, velocity, etc.) are analysed as functions of time. Early studies regarded variability as error or "noise" in the performance of movement [25]. Others explained variability not as good or bad phenomena, but simply stated that it reflects the variety of coordination patterns used to complete the task [9,17]. Within the context of human body movement, the definition of the coordination varied from selective activation of degrees of freedom resulting in organized motor activity, or mastering redundant degrees of freedom in order to create coordinated locomotion patterns [31], to the most recent understanding by introducing the concept of abundancy, which means that all degrees of freedom contribute to the stability and flexibility of the task [32]. It has been suggested that the coordination or coupling between segments may be an important line of investigation [6]. Quantifying the coordination between two body segments depends not only on the technique used for the assessment but also on the researcher's particular approach. The most popular methods for quantifying coordination appear to be vector coding and continuous relative phase (CRP) [1,3,6,10,17,21,28] or principal component analysis [33]. The CRP quantifies the coupling (coordination) relation between the kinematics of two body segments that are linked in this case both anatomically and mechanically. When analyzing CRP graph, the one can evaluate whether the segments are moving in-phase (CRP is closer to 0°) or antiphase (CRP is closer to 180°). In gait analysis, most common measures derived from CRP data are averages over a functional unit of the movement cycle [28]. CRP measures have been used to quantify the coordination between different body segments and joints during various activities [13]. Advances in the field of non-linear dynamics have shown that collective variables, such as relative phase, are able to capture the underlying spatial-temporal dynamics of coordination [7–9,27]. CRP is usually applied when analyzing cyclic movements [13,28]. However, studies show that it has the potential to provide quantitative information on multi-joint coordination of discrete movements [3,19], which is the case when performing simple WMFT motions. Unfortunately, there is lack of studies where CRP was used for separately quantifying coordination and motor function states in the movement assessment of damaged UE. Coordination plays a significant role in important daily activities, which are also the focus of advanced UE stroke rehabilitation. Quantitative information gained from CRP analysis may further facilitate clinical assessment and guide personalized stroke rehabilitation.

This study is focused on the assessment of stroke-affected UE movement based on intra-limb coupling strength and coordination analysis. The main purpose is to quantitatively represent UE coordination while performing non-cyclic movements during clinical motor function assessment.

2. Methods

2.1. Subjects

Kinematic data from UE were collected at Vilnius University Hospital "Santariskių Klinikos" Center of Rehabilitation, Physical and Sports Medicine. Fourteen stroke patients (age 60.8 ± 12.5 (mean ± SD)) were recruited for the study. The inclusion criteria were: all participants suffered their first ischemic stroke, paresis of the most affected upper extremity – 2 points of elbow and shoulder flexor and extensors muscles force according the Lovett scale, had no previous orthopaedic surgery or rehabilitation treatment, had the ability to sit in the wheelchair or on the chair and move the affected UE, and had the ability to understand and follow verbal instructions. The exclusion criteria were as follows: haemorrhagic stroke, repetitive rehabilitation, affected limb plegia, muscle tone of affected limb more than 1 point according to Modified Ashworth Scale, Mini-Mental State Examination less than 24 points and other diseases or states influencing motor control of upper extremities. Since the aim of this study is to investigate whether CRP is correlated to WMFT scores as provided by clinicians, no healthy subjects were included as a control group.

All procedures performed in studies involving human participants were in accordance with ethical standards of the national bioethics committee (protocol No. 65-11-95) and with the 1964 Helsinki declaration.

2.2. Experimental setup

A total of ten motor tasks based on WMFT and disability and health ICF guidelines were included in a clinical test series [26]. The motor tasks were selected regardless of the patients' level of UE function. Three wireless inertial measurement units (IMU) (Shimmer Research, Dublin, Ireland) were fixed on the segments of the most stroke affected upper extremity (Fig. 1) and were used for measuring the kinematics of the upper extremity during the movement. Each motor task was evaluated by WMFT score based on the performance and all scores were summed up. However, not all subjects were able to complete all ten tasks, i.e. if they failed to perform a task the corresponding score was 0 and the kinematic data from IMUs were not collected. The item task 3 of part A of WMFT was one of the easiest tasks which all 14 subjects were able to perform. Therefore in this study only this one task will be analysed in detail. This motor task focusses on the extension of the elbow on a table. The patient is sitting on a chair in front of a table and places his/her hand on the table. Then he/she attempts to reach across the table by pushing the forearm forward. The movement should be initiated by the shoulder and upper arm leaning

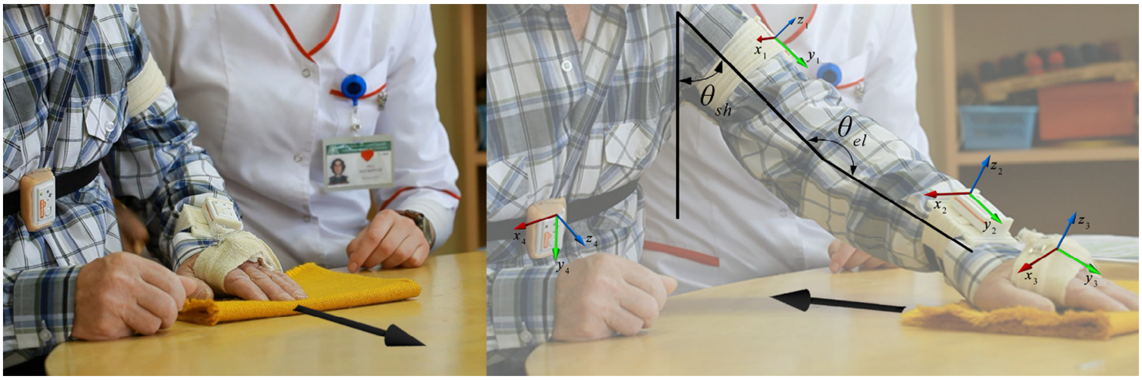

Fig. 1. Placement of the IMU sensors on the upper extremity. Left: UE at initial position. Right: the elbow in extended position before moving the hand back. Black arrows indicate direction of movement.

the upper body forward. Afterwards the patient is pulling the UE back without reclining the whole body. The movement is similar to wiping a cloth over the table surface while sitting in an upright position, i.e. it makes only use of flexion/extension of the shoulder and elbow joints. The subjects were divided into groups W2, W3 and W4, which corresponded to clinical evaluation in points 2, 3 and 4 according to the WMFT scale.

The IMU sensor on the upper arm was placed in the segment's center of the mass (according to anthropometric tables and physical measures of the individual length of the patient's segments), while the sensor on the forearm was placed at the distal end. A third sensor was fixed on the dorsal surface of the palm. The same sensor placement procedure was used for all patients. The acquisition of the data was implemented via Labview (National Instruments, USA) based application, which controlled and synchronized all sensors.

The subjects were instructed to follow verbal commands and a sound cue from the software to start the task. All subjects repeated the task three times. The kinematic data of the sensors (linear acceleration and angular velocity) was sampled at a rate of 51.2 Hz and stored on a personal computer for processing.

2.3. Data processing and analysis

Matlab (Mathworks Inc., Sweden) was used for all signal and data processing. A 2nd order Butterworth low-pass filter with a cut-off frequency of 5 Hz was used to remove noise from the raw data. A complementary filter algorithm for integrating accelerometer and gyroscope measurements based on white paper by Colton [29] was adapted for estimating the angles of the shoulder (θ_{sh}) and elbow (θ_{el}) joints [30]. The angle of the j-th joint in sagittal plane (zy plane, Fig. 1) was calculated as follows:

$$\theta_j = \underbrace{b\cdot\left[\theta_{0,j} + \underbrace{\int \omega_{z,j}dt}_{\text{integration}}\right]}_{\text{high-pass filter}} + \underbrace{(1-b)\cdot\theta_{a,yz}}_{\text{low-pass filter}}, \qquad (1)$$

where $j = 1$ (shoulder), 2 (elbow); b – constant of the complementary filter which relates sampling frequency and relative duration of the signal it will act on, and it equals to 0.911 for the selected time constant 0.2 s and it is calculated as $b = \frac{0.2}{0.2+1/51.2}$, where 51.2 is the sampling rate in Hz; $\omega_{z,j}$ – angular velocity of the arm's segment about axis perpendicular to zy plane; $\theta_{0,j}$ – initial angle of the upper extremity segment with respect to y axis and it is estimated from two-argument inverse tangent of the acceleration readings when arm is stationary (from initial time 0 to 0.5 s of measurement) and $\theta_{a,yz}$ when arm is in motion:

$$\theta_{0\text{ or }a,j} = \tan^{-1}\left(\frac{a_{yj}}{\sqrt{a_{xj}^2 + a_{zj}^2}}\right), \qquad (2)$$

where a_{xj}, a_{yj}, a_{zj} – linear acceleration data from 3-axial accelerometer.

Since the Eq. (1) refers to joint angles in global reference frame, the angle of the elbow joint is expressed as relative to the shoulder (θ_{el} in Fig. 1) and it was calculated as follows:

$$\theta_{el} = \theta_2 - \theta_1. \qquad (3)$$

Each trial was processed individually and then the average of the individual trials was used for further comparison between different groups. The time scale was normalized to 100 percent of the motion cycle. A movement cycle represents the two main parts of the task: forward motion and back motion. Forward motion consists of pushing the hand forward by extending the elbow and is determined as the relative time duration from beginning of the motion to the maximum of the joint amplitude and the whole movement is completed by pulling the hand back.

Normalization of two signals that make up the phase plane of the joint is necessary to account for the amplitude differences in the signals [3,8,13,17,21,22]. Angular velocity and angular displacement were normalized according to a method A described in [11]:

$$\tilde{\omega}_j = \left(\frac{\omega_j}{\max\{|\omega_j|\}}\right), \tilde{\theta}_j = \left(\frac{\theta_j}{\max\{|\theta_j|\}}\right). \qquad (4)$$

The phase angle of a segment is then given by the following equation:

$$\varphi_j(t) = \tan^{-1}\left(\frac{\tilde{\omega}_j}{\tilde{\theta}_j}\right). \qquad (5)$$

The mean CRP provides quantitative information about the spatial organization of segments during a given task. The CRP was assessed over intra-limb coupling between shoulder and elbow. The CRP was calculated as the difference between the shoulder and elbow phase angles:

$$\Phi(t) = \varphi_{sh}(t) - \varphi_{el}(t). \qquad (6)$$

In order to better reflect the differences in CRP between the groups, a coordination variability measure was determined. Variability is expressed as between-cycle standard deviation of CRP across all trials within each group for each portion (forward and back) of the movement cycle. Furthermore, a root-mean-square of average CRP across functional phase of movement was calculated. Additionally the amount of angular displacement during the task as functional range of motion *fROM* of single joint within the task was calculated as follows:

$$fROM_j = \max(\theta_j) - \min(\theta_j). \qquad (7)$$

2.4. Statistical analysis

Using Matlab software (Mathworks Inc., Sweden) Pearson's correlation coefficient r_{int} was calculated to assess intra-limb coupling between shoulder and elbow joints during the motor task.

3. Results

Average combined WMFT score of all motor tasks was calculated for each of the groups and is presented in Table 1. Subjects from W2 group were those who weren't able to complete all ten motor tasks and their relatively low average score reflects the

Table 1
Calculated parameters.

Parameters	W2 ($n = 4$)	W3 ($n = 5$)	W4 ($n = 5$)
Average WMFT score ± SD	6 ± 2.92	27.33 ± 8.26	38 ± 2.76
Average $fROM_{el}$, ° ± SD	0.895 ± 0.393	3.060 ± 1.44	4.155 ± 1.127
Average $fROM_{sh}$, ° ± SD	3.262 ± 0.814	8.607 ± 0.62	17.381 ± 4.26
r_{int}	0.862	0.925	0.96
Average t_{el}, % of cycle ± SD	69.0 ± 3.61	58.6 ± 7.80	46.0 ± 7.31
Average t_{sh}, % of cycle ± SD	47.3 ± 24.09	55.4 ± 10.33	43.8 ± 6.54
$T_{forward}$, % of cycle ± SD	58.17 ± 15.32	57.0 ± 2.26	44.9 ± 1.60
Average $T_{forward}$, % of cycle ± SD	53.36 ± 6.38		

t_{el} – Average time of the elbow joint extending from initial position to maximum amplitude.

t_{sh} – Average time of the shoulder joint flexing from initial position to maximum amplitude.

$T_{forward}$, – Relative time of dissected movement of both segments respectively forward.

severity of the impairment and associated smaller functional *fROM* (7) of the joints. The intra-limb invariance in the cross-correlation measures was strong overall, where r_{int} values were greater than 0.8 for couplings in all impairment levels (Table 1). The correlation coefficient between two segments increases from a strong level in W2 to a very strong level in W4 as it could be subjectively predicted by visual observation of smoothness, speed and accuracy of the UE movement. Average duration of forward motion for all groups was 53.36 % of the cycle.

The differences in phase plane and ROM can clearly be seen from Table 1 and in Fig. 2. Observing the smoothness of the phase plane diagrams, it is apparent that patients in the W4 group had stronger control and coupling of their UE segments. A vertical line drawn from zero on the horizontal axis divides the plot into two sides – negative and positive velocities. The motion trajectories of the elbow in W2 group mostly appear on a negative side of phase plane (Fig. 2).

Noticeably distorted curve in W2 group could be explained by the fact that UE movement during the task was carried out mostly by pushing the shoulder forward. Nevertheless, with less UE impairment (in groups W3 and W4) the phase plane plots show more smooth curves and faster movements of the UE. Also visually the motion is more in-phase. The smoothness of the phase plane is increasing with movement that is more accurate and the shape of the UE segment trajectories is nearly the same (W4).

The average CRP (the difference between the two phase angles) of shoulder and elbow joints $\Phi(t)$ (6) is presented in Fig. 3.

In this study the movement cycle (given as movement in % in Figs. 3 and 4) represents forward (up to 53.36 % of the cycle while extending an elbow) and backwards (from 53.36 to 100 % of the cycle while pulling the hand back) motions of UE. The measure of coordination variability is expressed as between-cycle standard deviation of the CRP across all trials within each group for each portion (forward and back) of movement cycle was calculated together with a root-mean-square of average CRP across functional phase of movement and the results are presented in Fig. 4.

It can be noted, that the highest variability is in the W4 group and it stays almost the same during the whole movement cycle.

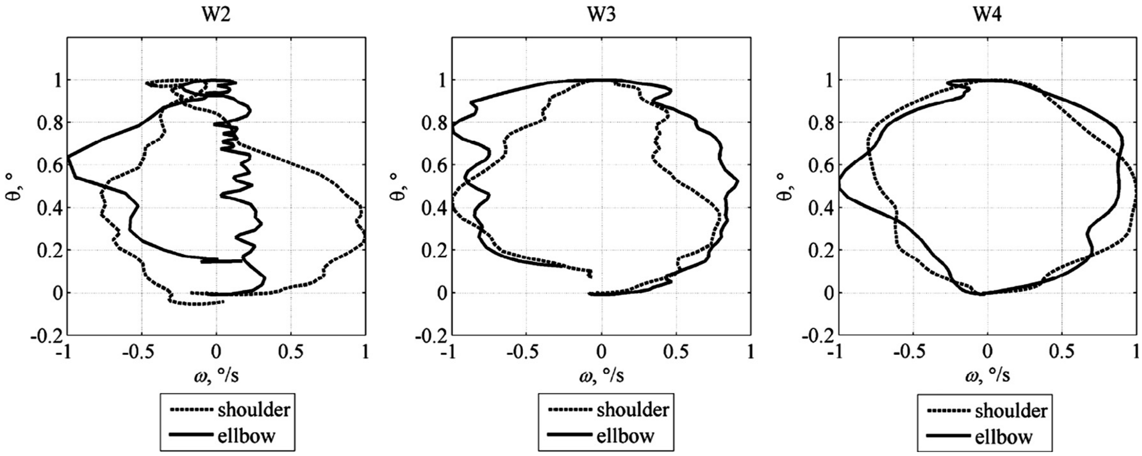

Fig. 2. Phase planes of shoulder-elbow at different impairment levels.

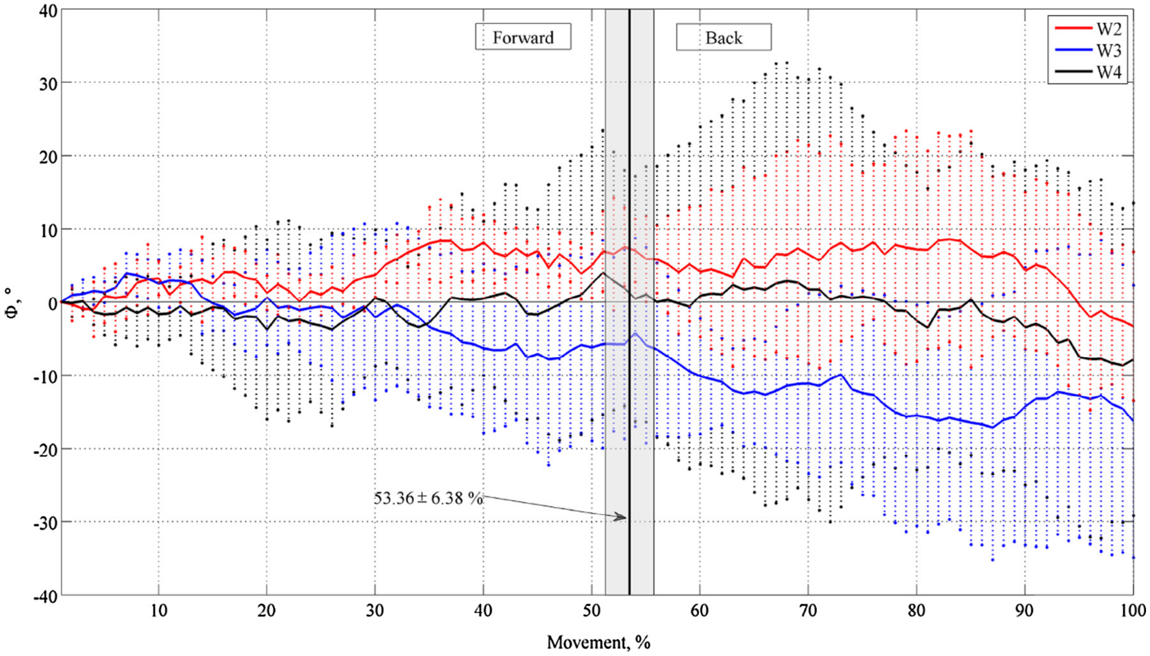

Fig. 3. Average CRP $\Phi(t)$ of UE segments within the groups. Vertical lines (error bars) indicate standard deviation of the CRP around the mean (bold solid lines).

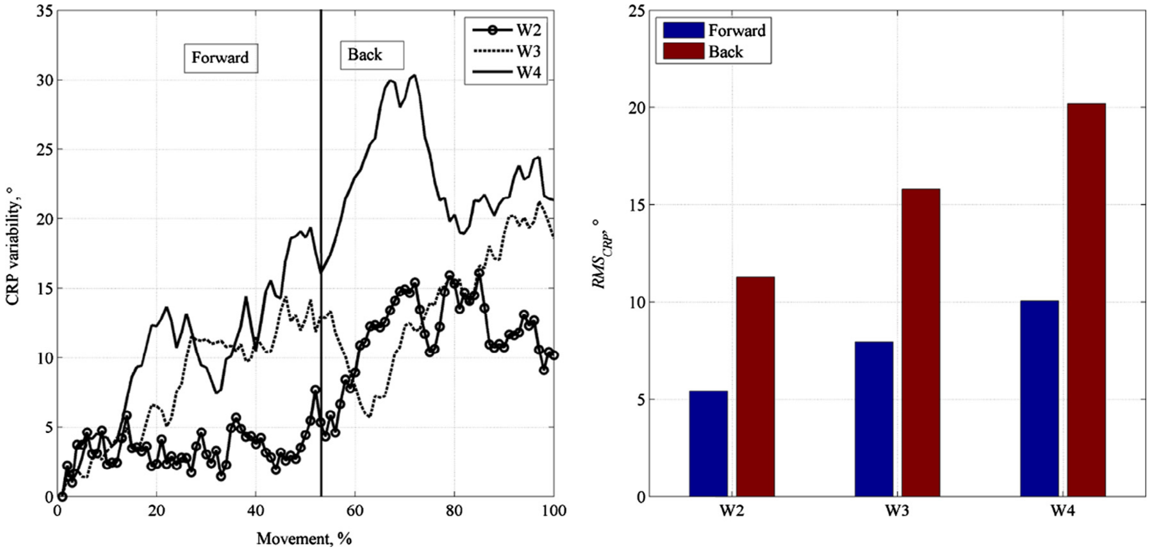

Fig. 4. Variability of CRP between the groups expressed as standard deviation of CRP (left) and RMS_{CRP} (right).

The lowest overall variability is in the W2 group, however when dissecting the movement into forward and back portions, the average RMS_{CRP} increases two times (from 5.14° to 11.09°) and indicating that pulling the hand back is much easier task than pushing it forward. Observing the test procedure visually it was clearly seen that patients pulled the hand back much faster than pushing it forward.

4. Discussion and conclusions

Usually UE coordination is tested with a sequence of movements. This can be repeated a few times assessing the smoothness and accuracy of the movement. Further assessment can be obtained by having the patient perform a rapidly repeated movement. The clinical screening of the upper extremity motor function might be improved by adding the quantitative data to support clinical rate after visual inspection. For quantifying a coordination pattern in movement, CRP was used and joint angle-angle diagrams, phase angle and additional methods supplement the results [6,10,17,21]. CRP analysis was previously used to assess locomotion [8], gait symmetry and coordination [6] as well as upper and lower extremities coordination [12]. To date, CRP is not commonly used for separately quantifying coordination and motor function states in the damaged upper extremity's movement assessment. Previously CRP was used to evaluate changes in limb coordination under each experimental load condition compared to a no load baseline condition [6]. Both changes in magnitude of CRP's RMS as well as temporal changes in CRP across the movement cycle were used for assessment. Coordination changes in RMS were also observed for the inter-limb couplings [6,8]. In the current study correlation was also used to quantify changes in the CRP. Intra-limb couplings in the cross-correlation measures was strong overall, where r_{int} values were greater than 0.8 for couplings in all impairment levels (Table 1). Lower coordinative variability (i.e. tighter coupling) is the norm for individuals with knee pain [28]. In this study, a similar phenomenon can be observed when analyzing CRP differences between the groups via CRP variability measures as the standard deviation of the average CRP between the WMFT groups or average RMS_{CRP} during the forward-backward phases of the movement. Decrease of variability was greater in the W2 group and indicates a decreased function of upper extremity. Generally, greater CRP variability is associated with a more unstable movement pattern, indicating the generation of new movement or a switch to a different movement pattern. Decreased variability of the CRP supports the "loss of complexity" hypothesis, which suggests that looser (i.e. increased variability) coupling between selected segments is the norm for a healthy individual [27]. In our measurement, we also found wide variations in impairment UE movements in all groups. One of the limitations of our study is that the CRP was developed and is usually applied for the coordination analysis of cyclic movements of limbs during gait. Therefore, the boundaries for higher or lower variability are still undetermined and required more studies with more subjects for supporting our hypothesis related to UE analysis by the CRP and, moreover, when analyzing semi-cyclic motions of the WMFT. Also, measuring a control group of healthy subjects would support better results of the increased variability.

In this study, the CRP was calculated using data normalization techniques based on prior recommendations [3,8,11,13,22]. Normalized series in CRPs present the information about trajectories and symmetry of movement cycle. In the present study, the approach carries valuable information of damaged UE coordination for different impairment levels, which cannot be observed visually during the task. The CRP quantifies the coordination between the kinematics of two body segments that are linked in this case both anatomically and mechanically. The differences in CRP between the groups represents differences in movement coordination along severity of the impairment. Unstable movement was indicated in higher impairment level (W2 group), the motion of the UE is more out of phase and taking more time. However, RMS of the CRP within each group for each part (forward and backward) of the movement cycle measures variability of CRP and quantifies differences in groups. The lowest overall variability, as it was predicted from CRP plots, was found in the W2 group and is associated with less coordinated movement of the UE. When dissecting the movement into forward and backward portions, the increased average RMS_{CRP} and increased CRP variability during the backward motion in all groups shows that pulling the hand back is much easier task than pushing it forward and the movement is less coordinated.

As it is clear from previous scientific work that it is not so easy and simple to identify differences between separate impairment levels visually. The evaluation of UE motor function is sometimes challenging even for experienced physicians. Despite the limitations and relatively small group of subjects, our study showed that instrumented motion analysis might serve as tool for decision making and improve clinical screening procedures. The CRP analysis enables to distinguish the performance of a stroke patient according the Wolf Motor Function Test. The coordination variability measures could be helpful in describing the motion in detail from beginning to an end, and prove the observations from the CRP plots. The future steps towards improving the UE function assessment would be implementation of the automated real-time evaluation and application during the rehabilitation process. More accurate assessment of the motor function might lead to the more personalized rehabilitation. The quantitative assessment of the therapeutic progress could be used to automatically increase the difficulty of the task in line with the performance, or design patient-specific motor tasks and build the basis for the continuous rehabilitation and training off-site.

Conflict of interest

The authors have no conflicts of interest related to this study

Acknowledgment

This research was funded from own resources.

References

[1] M. Alt Murphy, C. Willen, K.S. Sunnerhagen, Kinematic variables quantifying upper-extremity performance after stroke during reaching and drinking from a glass title, Neurorehabilit. Neural Rep. 25 (1) (2010) 71–80.

[2] M. Alt Murphy, C.K. Häger., Kinematic analysis of the upper extremity after stroke – how far have we reached and what have we grasped?, Phys Therap. Rev. 15 (3) (2015) 137–155.

[3] R. Burgess-Limerick, B. Abernethy, R.J. Neal., Relative phase quantifies interjoint coordination, J. Biomec. 26 (1) (1993) 91–94.

[4] A. Carmo, A.F.R. Kleiner, P.H.L. da Costa, R.M.L. Barros., Three-dimensional kinematic analysis of upper and lower limb motion during gait of post-stroke patients, Brazil. J. Med. Biol. Res. 45 (6) (2012) 537–545.

[5] M.C. Cirstea, A.B. Mitnitski, A.G. Feldman, M.F. Levin., Interjoint coordination dynamics during reaching in stroke, Experimental Brain Research 151 (3) (2003) 289–300.

[6] J.M. Haddad, R.E.A. van Emmerik, S.N. Whittlesey, J. hamill., Adaptations in interlimb and intralimb coordination to asymmetrical loading in human walking, Gait Posture 23 (4) (2006) 429–434.

[7] J. Hamill, J.M. Haddad, W.J. Mcdermott., Issues in quantifying variability from a dynamical systems perspective, J. Appl. Biomech. 16 (2016) 407–418.

[8] J. Hamill, R.E.A. van Emmerik, B.C. Heiderscheit, L. LI., A dynamical systems approach to lower extremity running injuries, Clin. Biomech. 14 (5) (1999) 297–308.

[9] H. Haken, J.A.S. Kelso, H. Bunz, A theoretical model of phase transitions in human hand movements, Biol. Cybernet. 51 (5) (1985) 347–356.

[10] C. Heiderscheit, J. Hamill, R.E.A. van Emmerik., Variability of stride characteristics and joint coordination among individuals with unilateral Patellofemoral pain, J. Appl. Biomech. 18 (2) (2002) 110–121.

[11] M.J. Kurz, N. Stergiou, Effect of normalization and phase angle calculations on continuous relative phase, J. Biomech. 35 (3) (2002) 369–374.

[12] G. Kwakkel R.C. Wagenaar, Effect of Duration of Upper- and Lower-Extremity Rehabilitation Sessions and Walking Speed on Recovery of Interlimb Coordination in Hemiplegic Gait, Physical Therapy, 82, 5, p. 432, 2002.

[13] P.F. Lamb, M. Stöckl., On the use of continuous relative phase: review of current approaches and outline for a new standard, Clin. Biomech. 29 (5) (2014) 484–493.

[14] A. Linkel, K. Daunoravičienė, J. Griškevičius, J. Žižienė, A. Juocevičius, I. Raudonytė, Applications of inertial sensors in medicine: towards model-based rehabilitation of stroke, IFAC-PapersOnLine 48 (4) (2015) 442–447.

[15] C. Lo et al., Multicenter Randomized trial of robot-assisted rehabilitation for chronic stroke: methods and entry characteristics for VA ROBOTICS, Neurorehabilit. Neural Rep. 23 (8) (2009) 775–783.

[16] S. Michaelsen, S. Jacobs, A. Roby-Brami, M. Levin., Compensation for distal impairments of grasping in adults with hemiparesis, Exper. Brain Res. 157 (2004) 2.

[17] R.H. Miller, R. Chang, J.L. Baird, R.E.A. van Emmerik, J. Hamill, Variability in kinematic coupling assessed by vector coding and continuous relative phase, J. Biomech. 43 (13) (2010) 2554–2560.

[18] E. Van wegen, R. Nijland, J. Verbunt, R. Van wijk, J. Van kordelaar, G. Kwakkel, A comparison of two validated tests for upper limb function after stroke: The wolf motor function test and the action research arm test, Journal of Rehabilitation Medicine, 42, 7, pp. 694–696, 2010.

[19] R. Robins, J. Wheat, G. Irwin, R.M. Bartlett, The effect of shooting distance on movement variability in basketball, J. Human Move. Stud. 50 (4) (2006) 217–238.

[20] J.L. Stephenson, A. Lamontagne, S.J. de Serres, The coordination of upper and lower limb movements during gait in healthy and stroke individuals, Gait Posture 29 (1) (2009) 11–16.

[21] M. Varlet, M.J. Richardson, Computation of continuous relative phase and modulation of frequency of human movement, J. Biomech. 44 (6) (2011) 1200–1204.

[22] R.E.A. Van emmerik R.C. Wagenaar, Effects of walking velocity on relative phase dynamics in the trunk in human walking, Journal of Biomechanics. 29, 9, pp. 1175–1184, 1996.

[23] S.L. Wolf et al., Effect of constraint-induced movement therapy on upper extremity function 3 to 9 months after stroke, JAMA, 296, 17, p. 2095, 2006.

[24] M. Woodbury et al., Measurement structure of the wolf motor function test: Implications for motor control theory, Neurorehabilit. Neural Rep. 24 (9) (2010) 791–801.

[25] K. Davids, P. Glazier, D. Araujo, R. Bartlett., Movement systems as dynamical systems – the functional role of variability and its implications for sports medicine, Sports Med. 33 (4) (2003) 245–260.

[26] R. Nijland, E. van Wegen, J. Verbunt, R. van Wijk, J. van Kordelaar, G. Kwakkel, A comparison of two validated tests for upper limb function after stroke: the wolf motor function test and the action research arm test, J. Rehabilitat. Med. 42 (7) (2010) 694–696.

[27] L.A. Lipsitz, Dynamics of stability: the physiologic basis of functional health and frailty, J. Gerontol. Series A: Biolog. Sci. Med. Sci. 57 (3) (2002) B115–B125.

[28] J. Hamill, C. Palmer, R. Van emmerik, Coordinative variability and overuse injury, SMARTT, 27, 1, pp. 4:45, 2012.

[29] S. Colton, The Balance Filter A Simple Solution for Integrating Accelerometer and Gyroscope Measurements for a Balancing Platform, Massachusetts Institute of Technology, Cambridge, 20. 2007. Retrieved from http://web.mit.edu/scolton/www/filter.pdf

[30] R.E. Mayagoitia, A.V. Nene, P.H. Veltink, Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems, J. Biomech. 35 (4) (2002) 537–542.

[31] M.T. Turvey, Coordination, Am. Psychol. (1990), http://dx.doi.org/10.1037/0003-066X.45.8.938.

[32] M.L. Latash, J.P. Scholz, G. Schöner, Motor control strategies revealed in the structure of motor variability, Exerc. Sport Sci. Rev. 30 (1) (2002) 26–31, http://dx.doi.org/10.1097/00003677-200201000-00006.

[33] A. Ozturk, A. Tartar, B. ersoz huseyinsinoglu, A.H. Ertas, A clinically feasible kinematic assessment method of upper extremity motor function impairment after stroke, Measurement, 80, pp. 207–216, 2016. https://doi.org/10.1016/j.measurement.2015.11.026