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RESEARCH ARTICLE

Stroke survivors exhibit stronger lower extremity synergies in more challenging walking conditions

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Abstract
The aim of this study was to examine how kinematic synergies are utilised as compensatory movements to stabilise foot positions under different walking task constraints in people with stroke. Ten (Males = 6, Females = 4) hemiplegic chronic stroke survivors volunteered to participate in this study, recruited from a rehabilitation centre. They completed a consent form and participated in treadmill walking tasks; flat, uphill, and crossing over a moving obstacle. The uncontrolled manifold method was used to quantify kinematic synergies in the paretic and non-paretic legs during their swing phase. The results of this study showed the strength of synergies was significantly greater in the obstacle task than in the uphill walking tasks at mid and terminal swing phases. In conclusion, the results suggest that walking in the challenging situations caused people with stroke to control step stability with greater compensation between lower extremity joints. Participants adapted to the increased challenge by increasing the amount of 'good variability', which could be a strategy to reduce the risks of falling.

Keywords Gait · Stroke · Uncontrolled manifold · Synergies

Introduction

The complexity of human movements is a natural and remarkable phenomenon. The question of how the central nervous system (CNS) governs this complexity and utilises it during voluntary actions has been a fundamental concern of motor control theories. It has been suggested that the complexity is controlled through an increase or decrease in the system dimensionality (Van Emmerik et al. 2005), which enables the system to "solve the degrees of freedom (DoFs) problem" (Bernstein 1967; Newell et al. 2003). Several motor control theories propose different ways the CNS solves the degrees of freedom problem. For example, motor redundancy theory places emphasis on eliminating the redundant (unnecessary) elements to optimise performance (Bernstein 1967). The motor abundance principle (Gelfand and Latash 1998) focuses on how DoFs are coordinated as families of solutions (synergies), by imposing constraints to interact with demanding environments. Utilisation of all available DoFs in a motor system provides an adequate and effective system that can react quickly to perturbations and meet the requirements of a task (Latash et al. 2007).

Scholz and Schöner (1999) proposed the uncontrolled manifold (UCM) method to quantify synergy in a multi-segment motor system. The UCM method partitions system variability into "good" and "bad" based on its contribution to the performance variable (e.g. foot position for stability). Variability in elemental variables (e.g. joint angles) that has no effect on the performance variable is classified as "good" (Latash et al 2007). On the other hand, variability in elemental variables that causes variability in the performance variable is considered "bad" (Latash et al 2007). According to the UCM method, greater "good" than "bad" variability suggests evidence of motor synergy (Scholz and Schöner 1999).

Some studies have reported the increased synergies as an adaptive strategy, under task and environment constraints (Qu 2012; Rosenblatt et al. 2014, 2015; Kao and Srivastava 2018). Kao and Srivastava (2018) showed that stroke survivors demonstrated greater kinematic synergies at slower speeds (60–80% preferred speed) than faster speeds (100% and fastest possible). Qu (2012) studied kinematic synergies

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and gait variability in fatigue and load carriage conditions, in able-bodied young males during treadmill walking. Load carriage and fatigue increased the magnitude of "good" variability and increased the strength of synergy in the frontal plane—which were suggested to be adaptations to stabilise the whole-body centre of mass (Qu 2012). However, sagittal plane stability control was only affected by load carriage (Qu 2012). Rosenblatt et al. (2014) used the UCM model to investigate the effect of task precision demands on motor synergy in able-bodied individuals during over ground walking. The need for motor synergy increased as the demands for precision in foot placement increased and participants achieved this by increasing the magnitude of "good" variability (Rosenblatt et al. 2014). In another treadmill walking experiment, Rosenblatt et al. (2015) manipulated the task in terms of treadmill width with a group of able-bodied young adults. They showed that the need for motor synergy during mid-swing was increased when walking on a treadmill with a narrower belt because the swing foot moves closer to the stance leg in the frontal plane, and there is a greater need to stabilise the foot trajectory to prevent the risk of collision between the foot and stance leg. Finally, Tokuda et al. (2018) studied gait synergies in people with osteoarthritis at normal walking and lateral trunk lean walking (participants were required to lean the trunk by a specific target angle after the initial contact). Synergy indices were not different between conditions; however, the amount of "good" variability was greater in the lateral lean condition.

It is necessary for stroke survivors to adapt to environmental constraints, such as uneven surfaces, obstacles, slops and stairs, in daily living for postural stability and the prevention of falls. Body adaptations in response to environment and task constraints have been shown in people with stroke (Liao et al. 2014; Novak and Deshpande 2014; Lowrey et al. 2007; MacLellan et al. 2015; Plummer-D'Amato et al. 2008, 2012; Manaf et al. 2015). Stroke survivors adapt frontal (e.g. hip abduction and circumduction) and sagittal (e.g. hip flexion, knee flexion) plane movements whilst clearing obstacles during walking (Lu et al. 2010; MacLellan et al. 2015) and walking uphill (Pandy et al. 2010; Perry and Burnfield 2010). More specifically, a compensation seen in hemiplegic stroke survivors (Balasubramanian et al. 2009; Hutin et al. 2011) is "frontal plane strategy"—increased hip abduction and pelvis tilt in the affected leg, as a response to muscle weaknesses in the sagittal plane (Stanhope et al. 2014). One way to understand the effectiveness of such strategy for postural stability and fall prevention is to examine the roles of movement variability and kinematic synergies under different task conditions. For example, Janshen et al. (2017) showed that the muscle synergies among active muscles in flat and inclined walking conditions were adjusted by the CNS to provide stable walking ability under varying mechanical demands.

Stroke survivors encounter walking conditions during activities of daily living which are challenging and require effective control of foot trajectory during the swing phase. There is evidence that stroke survivors altered kinematic synergies during walking with an ankle orthosis (Papi et al. 2015) or walking different from their preferred pace (Kao and Srivastava 2018). However, the effects of challenging walking conditions on swing phase foot trajectory and kinematic synergies in people with stroke has not been reported. Therefore, the aim of this study was to examine the effects of challenging walking task and environmental constraints on lower extremity synergies to stabilise swing foot trajectory in stroke survivors. We also were interested to know if synergy index and its variability components could predict the spatiotemporal gait parameters such as step width variability as indicator of risk of falling.

Methods

Participants

Ten participants (6 males and 4 females) volunteered to take part in this study and provided written informed consent. Participants were chronic stroke survivors that were referred to an outpatient rehabilitation clinic (Table 1). The hospital and university ethics committees approved all stages of the study. Inclusion criteria were the ability to walk independently (with or without an assistant device) on the treadmill for 2 min, the absence of orthopaedic problems or significant pains in the lower extremities that limit their independent walking, and being hemiplegic in the right or left side of the body. All participants were assessed on motor ability (Motor Assessment Scale; MAS), balance (Timed Up and Go; TUG) and walking ability (10 meters walk test, 10MWT). Patients with inability to control their posture or legs (MAS score < 24), poor dynamic balance (TUG > 150 s), or inability to walk on the treadmill for 2 min were excluded from the study.

Materials

A walking treadmill (STEX, 8100TD Medical, Fitness Europe, Germany) that was adjustable in terms of slope and speed was used in this study. For safety reasons, there were two handles and a body-support harness to prevent any accident because of failure to walk or losing balance in some patients. The patients who were not able to walk independently (without supports) on the treadmill were excluded. A

Table 1 Demographic measures of participants on age, sex and motor functions

No	Sex	Age (year)	Affected side	Stroke Duration (months)	TUG (second)	MAS (score)	10MWT (score)
1	Male	78	Left	18	27.19	47	19.32
2	Male	74	Right	24	33	31	25.86
3	Male	34	Left	18	23	45	9
4	Male	55	Right	36	76	41	90
5	Female	45	Left	36	30	36	19.59
6	Female	72	Left	14	39	40	31
7	Female	70	Left	6	42	42	56.16
8	Male	75	Left	60	46	27	29.16
9	Male	57	Right	24	65	37	65
10	Male	58	Right	7	10.11	40	10.93
Mean		61.8 (14.51)	40% Right, 60% Left	23.9 (16.7)	39.13 (19.5)	38.6 (6.1)	45.6 (26.3)

TUG timed up and go, *MAS* Motor Assessment Scale, *10MWT* 10 meters walk test

physiotherapist accompanied the patients during the walking task and other clinical assessment tests.

A 3D motion analysis system was used to analyse gait. The system used three digital cameras (Panasonic V160 HD Camcorder, Japan), with a 6-megapixel resolution and 50 Hz frame rate. Cameras were focussed on the right side, left side and front of the treadmill (with a separation of approximately 60°). The cameras were placed 170 cm from the treadmill and at a height of 65 cm from the floor.

Reflective adhesive markers (20 mm) were attached on bony landmarks of both the right and left legs: second metatarsal head (foot), lateral malleolus (ankle), between lateral femoral epicondyle and fibula apex (knee), and greater trochanter. Anthropometric measures were used to calculate the joints centre. A calibration cube was used to calibrate the motion capture volume. The 3D motion data were analysed through Simi motion software (Simi Co, Germany). Raw data were smoothed using a Butterworth 2nd order low pass (dual-pass) filter with a 6 Hz cut-off frequency.

Procedure

Participants individually met a qualified physiotherapist and the principal investigator for clinical assessments. Limb function was assessed by MAS (Carr et al. 1985), which includes eight different movements, and it assesses the quality of motions that require balance, mobility, positioning and ambulation based on the 6-point rating scale (lowest score represents the most severe disability). Dynamic balance was assessed with the TUG test, a valid and reliable test for assessing balance and walking abilities (Shamay and Hui-Chan 2005). Participants sat on a chair, then stood up and walked a distance of 3 m and returned to the initial sitting position, with the time taken to complete this movement recorded. Each participant performed one trial after familiarisation with the procedures. The 10MWT was used to assess walking ability in participants. 10MWT is a valid and reliable mobility test in adults with neurological disorders (Jackson et al. 2008). Only one trial was undertaken to demonstrate the capacity for independent walking and participants were able to use assistive devices (e.g. stick) if they needed.

Preferred walking speed was determined during a familiarisation trial in which the participants walked on a flat treadmill for 1 min and investigators recorded the preferred speed of the participants (0.21 ± 0.09 m/s). There were three main walking tasks with 2 min duration for further analysis in this study: flat (FW), inclined uphill (UW) and obstacle (OW). In the UW condition, the participants were required to walk at their preferred speed with the slope of treadmill was set at + 5°. In OW condition, a 3D paper obstacle (L:50, W:3, H:1 cm) was attached orthogonally to the treadmill belt and participants were asked to cross the foot over the obstacle to avoid any contact whilst walking. Only the steps over the obstacle were included in the analysis for OW. In FW condition, they walked on the flat treadmill, as in the familiarisation phase. The order of the conditions was counterbalanced among participants. Twelve successive strides of each leg from the last moments of the walking trial were selected from each condition for further analysis. There was a 3-min break between conditions.

Data analysis

A spline interpolation method was used to normalise the raw kinematics data of a swing phase with respect to time (toe-off: 0% to initial contact: 100%). The UCM model and analysis methods described by Krishnan et al. (2013) were used to calculate variability components which affect (V_{ORT}—“bad” variability) and those that do not affect (V_{UCM}—“good” variability) mediolateral position of the ankle joint. Kinematic synergy index (ΔV_Z) was then calculated.

Kinematic synergy index was calculated using the UCM methods (see appendix) adapted from Krishnan et al. (2013). The adapted model separated the thigh and shank segments in the stance and swing legs due to kinematic characterises of hemiplegic gait such as decreased knee extension in swing leg (Moore et al. 1993) and increased hip abduction and knee flexion in the stance leg (Moseley et al. 1993). As gait stability is different and dependent whether the paretic or non-paretic leg is in swing (Stanhope et al. 2014), two UCM models were used; one for paretic leg and one for non-paretic leg swing phases. For the current model, $\Delta V_Z = 0.477$ represents the absence of a synergy, whereas $\Delta V_Z > 0.477$ represents the presence of a synergy and $\Delta V_Z < 0.477$ refers to anti-synergy.

The interpolated swing phase (0–100%) was also divided into three sub-phases: initial-swing (0–33%), mid-swing (34–67%) and terminal (68–100%) swing (Olney and Eng 2011). Initial contact and toe-off moments were determined when the swing leg was in its minimum and maximum length (in X axis), respectively. Two step widths were calculated; one for which the step was initiated by the paretic leg and one for which the step was initiated by the non-paretic leg. Step width variability was calculated based on the standard deviations of the absolute distance between the right and left foot markers in the mediolateral direction at initial contact. All values of ΔV_Z, V_{UCM} and V_{ORT} were averaged across swing sub-phases (initial, mid and terminal) and tasks.

A 2 (Leg) × 3 (Task) × 3 (Swing Phase) within-participants analysis of variance (ANOVA) with repeated measures on all factors was used to compare the selected dependent variables. A 2 (Leg) × 3 (Task) within-participants ANOVA with repeated measures on both factors was used to compare the spatiotemporal parameters. The Bonferroni post hoc test was used to follow-up the significant main effects and interactions. The Pearson correlation coefficient method was used to examine the association between ΔV_Z, V_{UCM}, V_{ORT}, and step width variability in paretic and non-paretic legs at initial contact in flat walking condition. The same method was used to examine the association between functional tests (MAS, TUG, 10MWT) and spatiotemporal variability during treadmill walking in the flat walking condition (Fig. 1).

The confidence interval was set at 95%, two-tailed. All statistical analysis was performed using SPSS IBM 22 (IBM, USA).

Results

Spatiotemporal parameters

The results of gait performance during treadmill walking are presented in Table 2. The results of 2-way repeated measures ANOVA showed that there was a significant main effect of task on variability in stride time interval ($F_{2,8} = 3.47$, $p < 0.05$, $\omega^2 = 0.44$), stride length ($F_{2,8} = 22.52$, $p < 0.05$, $\omega^2 = 0.81$), step length ($F_{2,8} = 24.95$, $p < 0.05$, $\omega^2 = 0.82$) and step width ($F_{2,8} = 5.12$, $p < 0.05$, $\omega^2 = 0.46$). The main effect of leg and the interaction between task and leg was not significant ($p > 0.05$). The results of Bonferroni post hoc test showed that the obstacle condition relative to other walking conditions had significantly higher stride time variability (OW = 0.08 ± 0.001, FW = 0.06 ± 0.008, UW = 0.05 ± 0.002), stride length variability (OW = 0.07 ± 0.001, FW = 0.04 ± 0.003, UW = 0.03 ± 0.004), step length variability (OW = 0.07 ± 0.005, FW = 0.03 ± 0.007, UW = 0.03 ± 0.006) and step width variability (OW = 0.03 ± 0.004, FW = 0.02 ± 0.005, UW = 0.02 ± 0.008).

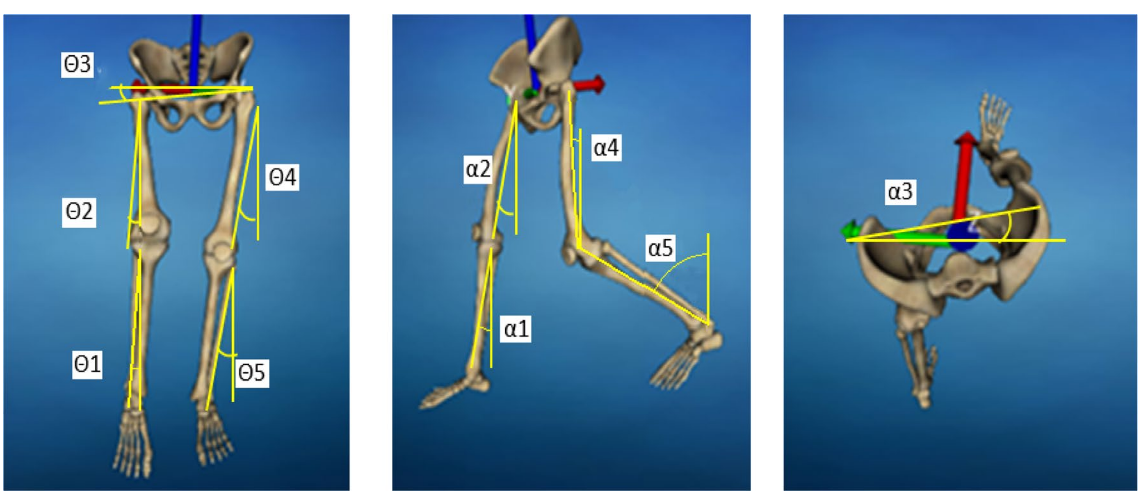

Fig. 1 The geometric model was used to quantify the multi-segment synergy

Table 2 Mean and variability (SD, CV) of spatiotemporal parameters during treadmill walking in different conditions

No.	Flat	Stride Time (s)		Flat	Stride Length (m)		Flat	Step Length (m)		Flat	Step Width (m)	
		Uphill	Obstacle		Uphill	Obstacle		Uphill	Obstacle		Uphill	Obstacle
Paretic leg												
1	0.718 (0.14,19.5)	0.543 (0.19,35)	0.565 (0.12,21.2)	0.47 (0.05,10.6)	0.45 (0.07,15.6)	0.49 (0.12,24.5)	0.27 (0.06,22.2)	0.25 (0.05,20)	0.25 (0.12,48)	0.23 (0.02,8.7)	0.25 (0.04,16)	0.25 (0.02,8)
2	0.440 (0.02,4.5)	0.416 (0.03,7.2)	0.345 (0.05,14.5)	0.34 (0.03,8.8)	0.35 (0.03,8.6)	0.33 (0.09,27.3)	0.20 (0.01,5)	0.18 (0.02,11.1)	0.19 (0.07,36.8)	0.24 (0.01,4.2)	0.24 (0.01,4.2)	0.22 (0.04,18.2)
3	0.650 (0.04,6.1)	0.618 (0.03,4.9)	0.525 (0.08,15.2)	0.38 (0.03,7.9)	0.46 (0.03,6.5)	0.43 (0.12,27.9)	0.25 (0.03,12)	0.22 (0.03,13.6)	0.24 (0.14,58.3)	0.18 (0.01,5.6)	0.18 (0.01,5.6)	0.18 (0.01,5.6)
4	0.441 (0.06,13.6)	0.541 (0.08,14.8)	0.481 (0.10,20.8)	0.50 (0.03,6)	0.48 (0.02,4.2)	0.52 (0.04,7.7)	0.22 (0.05,22.7)	0.21 (0.03,14.3)	0.21 (0.11,52.4)	0.23 (0.04,17.4)	0.26 (0.03,11.5)	0.25 (0.05,20)
5	0.450 (0.04,8.8)	0.393 (0.04,10.2)	0.390 (0.05,12.8)	0.38 (0.04,10.5)	0.36 (0.04,11.1)	0.34 (0.07,20.6)	0.21 (0.03,14.3)	0.25 (0.03,12)	0.23 (0.04,17.4)	0.22 (0.02,9.1)	0.25 (0.01,4)	0.24 (0.01,4.2)
6	0.316 (0.09,28.4)	0.300 (0.02,6.7)	0.320 (0.08,25)	0.41 (0.07,17.1)	0.44 (0.04,9.1)	0.45 (0.09,20)	0.27 (0.05,18.5)	0.28 (0.03,10.7)	0.23 (0.08,34.8)	0.20 (0.02,10)	0.20 (0.03,15)	0.21 (0.04,19)
7	0.319 (0.1,31.3)	0.325 (0.08,24.6)	0.332 (0.11,33.1)	0.43 (0.06,14)	0.45 (0.03,6.7)	0.44 (0.07,15.9)	0.21 (0.02,9.5)	0.19 (0.03,15.8)	0.20 (0.03,15)	0.21 (0.02,9.5)	0.20 (0.01,5)	0.20 (0.01,5)
8	0.189 (0.04,21.1)	0.220 (0.03,13.6)	0.215 (0.06,27.9)	0.32 (0.04,12.5)	0.34 (0.03,8.8)	0.39 (0.05,12.8)	0.10 (0.04,40)	0.15 (0.02,13.3)	0.14 (0.03,21.4)	0.25 (0.01,4)	0.25 (0.02,8)	0.24 (0.03,12.5)
9	0.700 (0.25,35.7)	0.510 (0.05,9.8)	0.663 (0.12,18.1)	0.41 (0.07,17.1)	0.35 (0.05,14.3)	0.49 (0.10,20.4)	0.23 (0.10,43.5)	0.21 (0.06,28.6)	0.25 (0.09,36)	0.24 (0.08,33.3)	0.23 (0.05,21.7)	0.22 (0.09,40.9)
10	0.315 (0.04,12.7)	0.283 (0.06,21.2)	0.368 (0.11,29.9)	0.31 (0.03,9.7)	0.40 (0.05,12.5)	0.36 (0.14,38.9)	0.20 (0.03,15)	0.18 (0.04,22.2)	0.19 (0.11,57.9)	0.27 (0.01,3.7)	0.26 (0.01,3.8)	0.23 (0.01,4.3)
Mean	0.453	0.415	0.421	0.39	0.41	0.42	0.21	0.21	0.21	0.22	0.23	0.22
CV (%)	18.2	14.8	22	11.4	9.7	21.6	20.3	16.2	37.8	10.5	9.5	13.5
Non-paretic leg												
1	0.655 (0.10,15.3)	0.572 (0.08,14)	0.575 (0.11,19.1)	0.55 (0.05,9.1)	0.48 (0.06,12.5)	0.53 (0.10,18.9)	0.29 (0.03,10.3)	0.25 (0.05,20)	0.28 (0.11,39.3)	0.23 (0.01,4.3)	0.24 (0.02,8.3)	0.25 (0.03,12)
2	0.288 (0.01,3.5)	0.275 (0.04,14.5)	0.273 (0.02,7.3)	0.30 (0.1,33.3)	0.27 (0.04,14.8)	0.27 (0.04,14.8)	0.18 (0.02,11.1)	0.19 (0.05,26.3)	0.17 (0.08,47.1)	0.18 (0.01,5.6)	0.19 (0.01,5.3)	0.15 (0.01,6.7)
3	0.516 (0.04,7.8)	0.516 (0.03,5.8)	0.490 (0.05,10.2)	0.50 (0.03,6)	0.62 (0.03,4.8)	0.57 (0.07,12.3)	0.34 (0.02,6)	0.25 (0.04,16)	0.26 (0.05,19.2)	0.15 (0.01,6.7)	0.15 (0.01,6.7)	0.15 (0.02,13.3)
4	0.366 (0.04,10.9)	0.337 (0.02,5.9)	0.321 (0.04,12.5)	0.41 (0.02,4.9)	0.42 (0.03,7.1)	0.40 (0.05,12.5)	0.21 (0.01,4.8)	0.22 (0.06,27.3)	0.24 (0.08,33.3)	0.20 (0.02,10)	0.25 (0.01,4)	0.24 (0.04,16.7)
5	0.288 (0.03,10.4)	0.235 (0.04,17)	0.265 (0.06,22.6)	0.40 (0.04,10)	0.37 (0.04,10.8)	0.36 (0.06,16.7)	0.20 (0.03,15)	0.21 (0.03,14.3)	0.20 (0.09,45)	0.16 (0.01,6.3)	0.23 (0.02,8.7)	0.17 (0.01,6)
6	0.359 (0.05,13.9)	0.308 (0.06,19.5)	0.333 (0.09,27)	0.50 (0.07,14)	0.52 (0.08,15.4)	0.56 (0.10,18)	0.21 (0.06,28.6)	0.20 (0.08,40)	0.20 (0.09,45)	0.19 (0.02,10.5)	0.19 (0.02,10.5)	0.18 (0.04,22.2)
7	0.363 (0.02,5.5)	0.360 (0.03,8.3)	0.364 (0.04,11)	0.31 (0.02,6.5)	0.32 (0.03,9.4)	0.30 (0.02,6.7)	0.11 (0.04,36.4)	0.12 (0.01,8.3)	0.16 (0.04,25)	0.18 (0.02,11.1)	0.19 (0.02,10.5)	0.17 (0.02,11.8)

Table 2 (continued)

No.	Stride Time (s)			Stride Length (m)			Step Length (m)			Step Width (m)		
	Flat	Uphill	Obstacle	Flat	Uphill	Obstacle	Flat	Uphill	Obstacle	Flat	Uphill	Obstacle
8	0.210 (0.04,19)	0.225 (0.04,17.8)	0.224 (0.03,13.4)	0.33 (0.03,9.1)	0.35 (0.02,5.7)	0.36 (0.04,11.1)	0.18 (0.03,16.7)	0.19 (0.02,10.5)	0.18 (0.02,11.1)	0.25 (0.01,4)	0.26 (0.02,7.7)	0.27 (0.03,11.1)
9	0.672 (0.15,22.3)	0.453 (0.10,22.1)	0.571 (0.19,33.3)	0.41 (0.07,17.1)	0.45 (0.04,8.9)	0.35 (0.08,23)	0.25 (0.05,20)	0.25 (0.03,12)	0.20 (0.08,40)	0.18 (0.05,27.8)	0.12 (0.02,16.7)	0.19 (0.04,21.1)
10	0.290 (0.02,7)	0.243 (0.02,8.2)	0.280 (0.07,25)	0.47 (0.03,6.4)	0.46 (0.02,4.3)	0.41 (0.09,22)	0.21 (0.02,9.5)	0.21 (0.01,4.8)	0.22 (0.07,31.8)	0.28 (0.01,3.6)	0.26 (0.01,3.8)	0.23 (0.01,4.3)
Mean	0.401	0.352	0.349	0.41	0.42	0.41	0.21	0.20	0.21	0.21	0.20	0.20
CV (%)	11.6	13.3	18.1	11.6	9.4	15.6	15.8	17.9	33.3	9	8.2	12.5

Pearson correlation coefficients showed that only MAS score significantly associated with step width variability in both paretic leg ($r = -0.65$, $p < 0.05$) and non-paretic leg ($r = -0.64$, $p < 0.05$). The relationship between other spatiotemporal gait parameters and motor functions were not significant ($p > 0.05$).

Synergy index in stroke survivors

The interaction of task, leg and swing phase on synergy index was significant ($F_{4,36} = 3.1$, $p < 0.05$, $\omega^2 = 0.35$), but the main effects of task, leg, swing phase and other interactions among factors were not significant (see Fig. 2). Post hoc analysis revealed that the highest synergy index were in the obstacle condition in paretic leg at terminal and mid-swing and in non-paretic leg at initial, mid and terminal swing. The paretic and non-paretic legs in uphill conditions had lowest synergy index.

There was a significant interaction of task and swing phase ($F_{2,18} = 4.8$, $p < 0.05$, $\omega^2 = 0.45$) and a significant main effect of task ($F_{2,9} = 33.5$, $p < 0.05$, $\omega^2 = 0.92$) on V_{UCM}. Post hoc tests showed that V_{UCM} was greater in the obstacle at terminal and mid swing phases than the uphill and flat walking tasks at all swing sub-phases (Fig. 3). There were no significant main effects of the leg, phase and interaction among them ($p > 0.05$). In addition, the main effect and interaction of all factors on V_{ORT} were non-significant ($p > 0.05$).

Correlation between motor synergy and step width variability

In the paretic leg, there was an inverse significant correlation between step width variability and ΔV_Z ($r = -0.68$, $R^2 = 46\%$, $p < 0.05$), and there were positive significant correlation coefficients between step width variability and V_{UCM} ($r = 0.65$, $R^2 = 42\%$, $p < 0.05$) and V_{ORT} ($r = 0.63$, $R^2 = 40\%$, $p < 0.05$). In other words, participants with more stable gait performance (less step width variability) were benefited more through increased ΔV_Z that was achieved through larger reduction of V_{ORT} than V_{UCM}. In the non-paretic leg, there were significant correlation coefficients between step width variability and V_{UCM} ($r = 0.88$, $R^2 = 77\%$, $p < 0.05$), but not ΔV_Z and V_{ORT} ($p > 0.05$).

Discussion

The aim of this study was to examine how kinematic synergies are utilised as compensatory movements to stabilise foot positions under different walking task constraints in people with stroke. The results suggested that kinematic synergies were stronger when participants had to adapt to a more

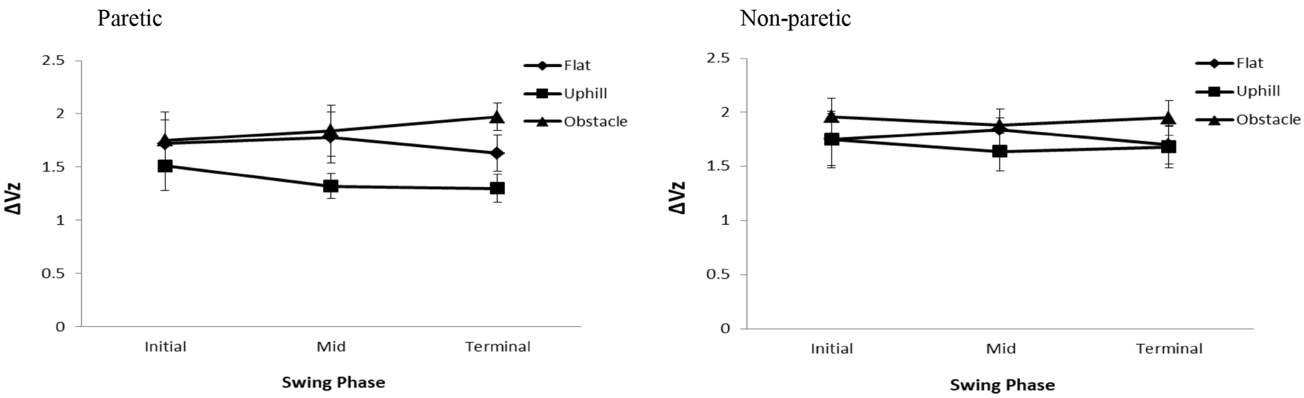

Fig. 2 The mean (± SD) synergy index in different tasks and different phases of swing and in paretic and non-paretic legs

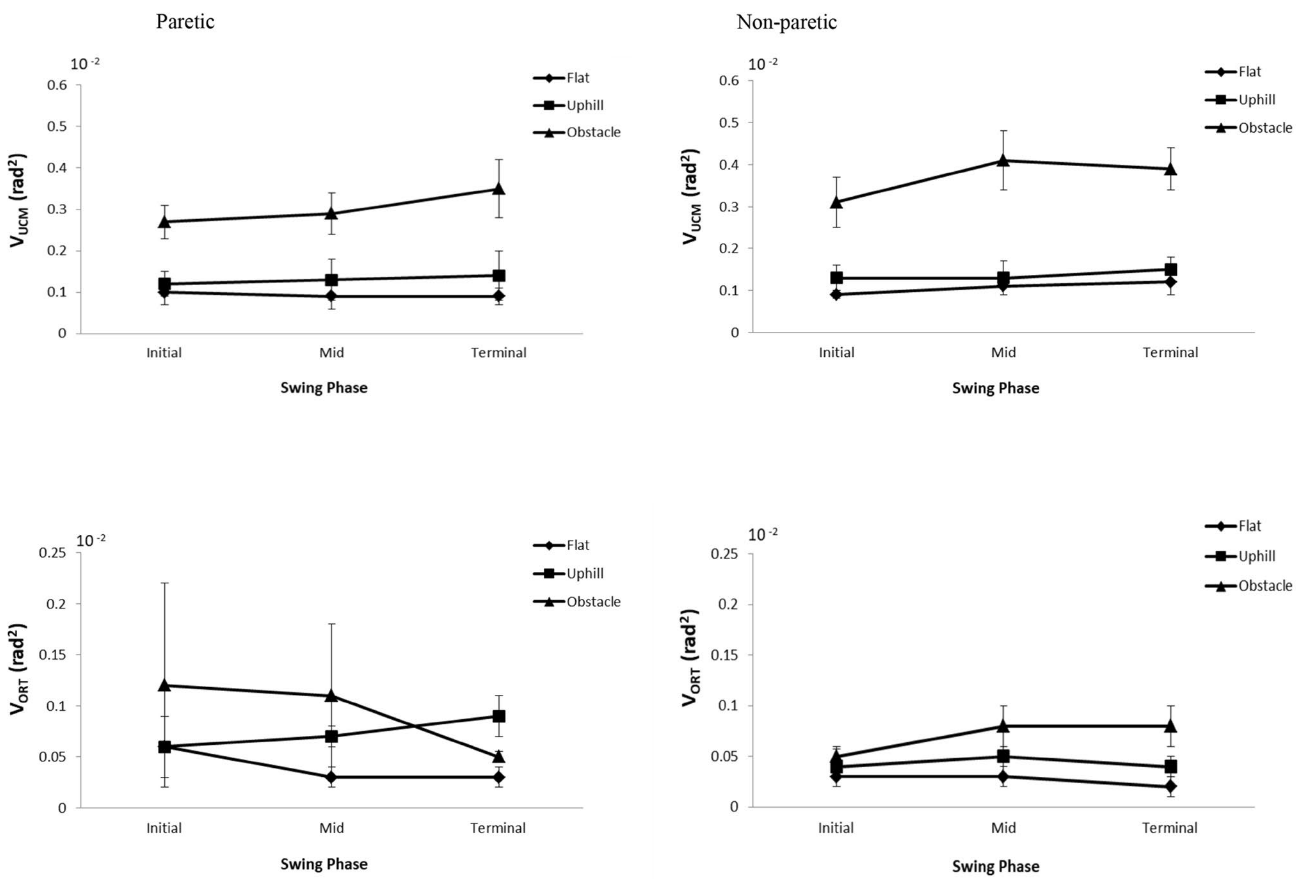

Fig. 3 The mean (± SD) V_{UCM} and V_{ORT} in paretic and non-paretic legs during different walking tasks and swing phases

challenging task—obstacle crossing—which increased the requirement for high mediolateral precision in foot trajectory especially at mid-swing and terminal swing of both legs. Furthermore, the stronger synergies during obstacle walking were related to an increase in V_{UCM} (good variability) rather than decrease in V_{ORT} (bad variability) at mid-swing and terminal swing phases.

Previous studies showed that the more challenging tasks resulted in a greater need to use kinematic synergies to stabilise the foot placement in terminal swing. For example, in young participants, changing walking task constraints (unconstrained/narrow pathway/beam walking) has been shown to affect motor synergy, with synergy index increasing from 0.58 to 0.67 and 0.77 as the task became more difficult (Rosenblatt et al. 2014). In a similar study, Rosenblatt et al. (2015) examined synergy index in double-belt and single-belt treadmill walking and showed that the single-belt, due to narrow walking width, required greater synergy (0.91) than double-belt (0.75).

The increased motor synergy was necessary as a compensatory strategy during obstacle negotiation for several reasons. First, hemiplegic stroke survivors usually use a "frontal plane strategy" as a compensation for muscle weaknesses (Stanhope et al. 2014). This, in turn, could affect mediolateral stability and subsequently the risk of falling. In other words, the reduced ability to stabilise the paretic swing leg (Dean and Kautz 2015) could result in a lateral movement if the centre of mass away from the stance foot. Both changes might demand stronger synergies in stroke survivors in order to adjust foot trajectory because motor synergies provide an effective system to react quickly against external and internal perturbations (Latash et al. 2007). Second, large joint angle variability is associated with an increased need for precision in mediolateral foot placement (Rosenblatt et al. 2014). The stronger synergies in the more challenging tasks could be related to specific biomechanical adaptations in people with stroke during obstacle negotiating tasks. For example, it was shown that, during obstacle crossing, stroke survivors demonstrated greater variation in centre of pressure placement in a mediolateral direction (Novak and Deshpande 2014), reduced crossing step velocity, shorter landing distance (Lowrey et al. 2007) and greater hip vertical elevation and hip abduction (MacLellan et al. 2015; Lu et al. 2010). In addition, the results showed evidence of less stable gait (increased variability in step width, stride time interval, step and stride length) in the obstacle condition for the stroke survivors regardless of their motor function (e.g. TUG and 10MWT). This might be a possible reason to explain the need for stronger motor synergies, as an effective adaptation, to stabilise foot placement in the obstacle condition. The increased synergies are adaptations that stem from feedback from stance leg during swing (Rankin et al. 2014) or peripheral information from the swing leg itself. Synergy helps to ensure that the foot trajectory is controlled appropriately during swing (Rosenblatt et al. 2015).

Previous studies in older and young healthy adults (Krishnan et al. 2013) showed that synergy was stronger in mid-swing because of a need to avoid a collision in mediolateral direction or in response to increased precision before initial contact (Rosenblatt et al. 2014, 2015). Similarly, the current study showed stronger motor synergies to control foot mediolateral foot trajectory in people with stroke in both mid and terminal swing sub-phases than in early swing. It is likely that the greater need for kinematic synergies in people with hemiplegic stroke in paretic leg at mid-swing and terminal-swing is due to decreased hip flexion at initial and mid-swing phases and decreased knee extension and ankle dorsiflexion in terminal phase that is required for fluent limbs movements (Moore et al. 1993).

The results of this study also showed that greater synergy indices were related to increases in V_{UCM} rather than decreases in V_{ORT} (see Fig. 3) at mid-swing and terminal swing. This finding supports previous studies that have shown the increased good variability in response to increased task difficulty (Rosenblatt et al. 2014, 2015) in able-bodied participants such as practice of a novel task (Yang and Scholz 2005), walking with a loaded backpack (Qu 2012), walking with a cognitive-secondary task (Zhang et al. 2008) and unexpected perturbations during walking (Mattos et al. 2011). The increased motor synergies through an increase in V_{UCM} variability could represent how the CNS uses more compensatory movements to solve the redundant joints DoFs in challenging tasks and in some critical periods of the swing phase.

There was no difference in synergies between paretic and non-paretic swing legs; however, the current findings indicate that the difference between legs depended on the swing sub-phase and the nature of task, suggesting that hemiplegic gait is not a unilateral functional weakness; in contrast, both legs act as synergy to accomplish the goals of walking in normal and challenging situations. One reason could be related to the dynamic nature of a gait cycle. In fact, gait—as a bilateral cyclic movement pattern—requires use of both legs at the same time for different roles. It seems that the inter-limb coupling of both legs through the whole gait cycle increases the kinematic synergy when the body requires postural control by the stance leg and body transporting by the swing leg (Winter 1987). Since the risk of falling is mainly determined by between-limb collision (Rosenblatt et al. 2015), the need for stability (by the stance leg) and forward progression (by swing leg) are equally important. In addition, after stroke, 15% of motor neurones from the affected cortical hemisphere are sent to the ipsilateral side of the body and cause impairment in the unaffected side (Wong et al. 2004). This might explain why the strength of motor synergy was different in the swing phases of both the paretic and non-paretic legs under different task constraints.

The results of this study also showed an association between motor synergy index and step width variability at initial contact in paretic leg but not in non-paretic leg. In other words, a decrease in step width variability was associated with an increase in motor synergy index in the paretic but not in the non-paretic leg. These findings showed that the compensatory movement in the paretic leg is effective, to some extent, in reducing the risk of collision between legs and increasing foot stability before initial contact; however, stroke survivors were not able to control the step width variability in the non-paretic leg in swing. This finding in the non-paretic leg was similar to older adults study (Krishnan et al. 2013). The findings of previous study in stroke survivors (Kao and Srivastava 2018) also showed that higher synergies were accompanied with less step width variability and an increased margin of stability. Additionally, Rosenblatt et al. (2014) and Qu (2012) showed an association between joint angle variability and step width variability at initial contact in young adults. Qu (2012) showed that the need for greater motor synergies was greater at initial contact than toe-off in both sagittal and frontal planes. Whether synergy index could be a robust clinical indicator to quantify gait stability still is not yet known, but the findings of the current and previous studies support the notion that the synergy index could play an important role in predicting gait stability in pathological gait patterns.

A possible reason for different findings in paretic and non-paretic legs could be the need for more active control in the paretic leg to compensate for a frontal plane strategy and sagittal plane weaknesses in hemiplegic gait (Dean and Kautz 2015; Stanhope et al. 2014). In fact, foot placement with the paretic leg at initial contact requires more synergies among the elements, whereas in the non-paretic leg the foot placement is controlled by individual elements acting in isolation rather than synergy (Krishnan et al. 2013). It seems that the situations with highest risk of fall—like initial contact—require that hemiplegic patients make more compensation through exploiting motor redundancy and channelling more variance into V_{UCM} to stabilise foot placement (Rosenblatt et al. 2014). Generally, the relationship between synergy index and step width variability supports the notion that motor synergies play an important compensatory role to control medio-lateral foot position.

The increased V_{UCM} and V_{ORT} in participants with greater step width variability are also interesting and indicate the different types of adaptation to stabilise mediolateral gait stability at initial contact. According to UCM theory (Latash et al. 2007), the stabilisation of a performance variable could be related to three possible scenarios in the variance of elemental variables: (1) V_{ORT} decreases relative to V_{UCM} – V_{UCM} could be constant, decreased or increased, leading to an increase in motor synergies (2) V_{UCM} is reduced to the same extent as V_{ORT}, leading to no change in the motor synergies; (3) V_{UCM} decreases relative to V_{ORT}, leading to a reduction in motor synergies. Our results showed that stronger motor synergies were associated with decreases in both V_{UCM} and V_{ORT} (scenario 1). This was confirmed by the results of supplementary analysis which revealed a negative and significant relationship between ΔVz, V_{ORT} and V_{UCM} in this study.

This study has implications for gait retraining in people with stroke. An effective strategy to control foot trajectory during treadmill walking was channelling variability into V_{UCM}. In other words, the stroke survivors were able to walk on the treadmill in more challenging conditions by exploiting considerable compensatory movements. It has been demonstrated that practice can have the effect of changing the magnitude of V_{UCM} (Wu et al. 2012). Designing gait re-training interventions that help stroke patients to explore joint variability through V_{UCM} could provide a safe context for independent walking. It is suggested that promoting increased V_{UCM}—more elemental variability—and minimising the V_{ORT}—less performance variability—could be achieved by extensive task-related practices on the treadmill. The variations in gait training could be implemented in conditions such as stair climbing, which requires foot adaptations for both uphill walking and obstacle crossing (stair nosing) at the same time.

We acknowledge some limitations to this study. This study only used the UCM method for one performance variable—medio-lateral foot trajectory—during gait. It is possible to configure the elemental variables in multiple planes of motion for postural stability and whole body transport. For example, gait stability during swing could be affected by toe clearance in different conditions that could cause tripping and risks of fall. Future studies could examine a more complex model of UCM for other functions of gait in stroke survivors. Further, patients in this study did not practice the treadmill walking in different slopes and with an obstacle as a part of their rehabilitation programme. As the treadmill walking task was new to patients, it provided sufficient challenge for exploring the DoFs, but future studies could be carried out in a natural context that is similar to activities of daily living.

Conclusion

In conclusion, the results of this study showed that negotiating an obstacle during walking requires stroke survivors to control swing leg foot trajectory with stronger kinematic synergies than less challenging walking conditions. Stroke survivors adapted to this increased challenge by channelling variability through V_{UCM}, which was beneficial for frontal plane stability.

Compliance with ethical standards

Conflict of interest The authors have not received any financial support in this study and declare no conflict of interest.

Appendix: Details of the UCM method adapted from Krishnan et al (2013)

Creating a model for elemental variables

The UCM method that is used in this study has 10 elemental variables and 1 performance variable (mediolateral trajectory of swing foot). A geometric model was created to relate elemental variables to the performance variable. This comprised a five-segment and 10 DoFs model (Fig. 1). The segments were stance leg, pelvis, swing-leg thigh and swing-leg shank with lengths L_1, L_2, L_3, and L_4 respectively. For the paretic swing leg in gait cycle, 5 DoFs are in the frontal plane, Θ_1 = stance-leg shank$_{\text{non-paretic}}$, Θ_2 = stance-leg thigh$_{\text{non-paretic}}$, Θ_3 = pelvis, Θ_4 swing-leg thigh$_{\text{paretic}}$ and Θ_5 = swing-leg shank$_{\text{paretic}}$, and 4 DoFs are out of the frontal plane; α_1 = stance-leg shank$_{\text{non-paretic}}$, α_2 = stance-leg thigh$_{\text{non-paretic}}$, α_3 = pelvis, α_4 = swing-leg thigh$_{\text{paretic}}$ and s_5 = swing-leg shank$_{\text{paretic}}$. The same model is used when for the non-paretic swing leg.

Segmental configuration

An ankle joint trajectory (AJT) and functions of ten DoFs ($\underset{\sim}{\Theta}$) from the segmental model was created by a custom-written code in Matlab 2018a (Mathworks, Natick, MA) according to the UCM procedure in Krishnan et al. (2013). Prior to UCM analysis, all segmental configuration and ankle joint trajectory (*AJT*) data were normalised for swing phases (0–100%):

$$\begin{aligned}\text{AJT} = &-L_1 \cos\alpha_1 \sin\theta_1 - L_2 \cos\alpha_2 \sin\theta_2 + L_3 \cos\alpha_3 \cos\theta_3 \\ &+ L_4 \cos\alpha_4 \sin\theta_4 + L_5 \cos\alpha_5 \sin\theta_5\end{aligned}$$

$$\underset{\sim}{\Theta} = \left[\theta_1\theta_2\theta_3\theta_4\theta_5\alpha_1\alpha_2\alpha_3\alpha_4\alpha_5\right]. \tag{1}$$

Calculation of V_{UCM} and V_{ORT}

As geometric models describing the position of an end effector are generally non-linear, uncontrolled manifolds are often curved. Therefore, the first step in calculating V_{UCM} and V_{ORT} is to perform a linearization around a reference configuration. A linear approximation of the model can be obtained by calculating the Jacobian matrix with respect to a reference configuration—the mean segment configuration across trials (Scholz and Schoner 1999). Deviations of segment vectors, for a particular trial, from the mean segment configuration can then be projected onto the null space (ε) of this Jacobian matrix:

$$J = \frac{\partial D}{\partial \Theta} = \begin{bmatrix} -L_1 \cos\alpha_1 \cos\theta_1, -L_2 \cos\alpha_2 \cos\theta_2, -L_3 \cos\alpha_3 \sin\theta_3, L_4 \cos\alpha_4 \cos\theta_4, L_5 \cos\alpha_5 \cos\theta_5, \\ L_1 \sin\alpha_1 \sin\theta_1, L_2 \sin\alpha_2 \sin\theta_2, -L_3 \sin\alpha_3 \cos\theta_3, -L_4 \sin\alpha_4 \sin\theta_4 - L_5 \sin\alpha_5 \sin\theta_5 \end{bmatrix}$$

$$\underset{\sim}{\Theta}_{\parallel} = \sum_{i=1}^{n-d} \left(\varepsilon_i \cdot \left(\underset{\sim}{\Theta} - \underset{\sim}{\Theta}^0\right)\right)\varepsilon_i s, \tag{2}$$

where n is the number of degrees of freedom of the model, d is the number of dimensions of the performance variable, $\underset{\sim}{\Theta}$ is a vector of segment angles for a particular trial and $\underset{\sim}{\Theta}^0$ is a vector of segment angles in the reference configuration. For this work, $n = 10$ and $d = 1$. A component of the deviation of $\underset{\sim}{\Theta}$ from $\underset{\sim}{\Theta}^0$ that is perpendicular to the UCM can also be calculated as:

$$\underset{\sim}{\Theta}_{\perp} = \left(\underset{\sim}{\Theta} - \underset{\sim}{\Theta}^0\right) - \underset{\sim}{\Theta}_{\parallel}. \tag{3}$$

The variability per degree of freedom parallel (V_{UCM}) and perpendicular (V_{ORT}) to the UCM was then calculated using:

$$V_{\text{UCM}} = \left(\frac{\sum \Theta_{\parallel}^2}{(n-d)N}\right), \tag{4}$$

and

$$V_{\text{ORT}} = \left(\frac{\sum \Theta_{\perp}^2}{dN}\right), \tag{5}$$

where N is the number of stride. V_{UCM} does not affect the variance in the mediolateral foot trajectory (Good variability), whereas V_{ORT} causes the variance in the foot trajectory (Bad variability).

Motor synergy index

If $V_{\text{UCM}} > V_{\text{ORT}}$, then lower limb synergy stabilise the foot trajectory in a gait cycle. A synergy index was calculated as:

$$\Delta V = \frac{V_{\text{UCM}} - V_{\text{ORT}}}{V_{\text{ToT}}}, \tag{6}$$

where

$$V_{\text{ToT}} = \left(\frac{1}{n}\right)\left(dV_{\text{ORT}} + (n-d)V_{\text{UCM}}\right), \tag{7}$$

n = number of DoFs; d = number of performance variable.

The more positive ΔV, the stronger the motor synergy among segments. If $V_{\text{ORT}} = 0$, then $\Delta V = 10/9$ and all variance lie within the UCM. If all variance lies in the orthogonal sub-space ($V_{\mathrm{UCM}} = 0$), $\Delta V = -10$. For consistency among studies, ΔV was transformed to Fisher's z-transformation index:

$$\Delta V_Z = \frac{1}{2} \log \left[\frac{10 + \Delta V}{\frac{10}{9} - \Delta V} \right]. \quad (8)$$

References

Balasubramanian CK, Neptune RR, Kautz SA (2009) Variability in spatiotemporal step characteristics and its relationship to walking performance post-stroke. Gait Posture 29:408–414

Bernstein N (1967) The coordination and regulation of movements. Pergamon Press, London

Carr JH, Shepherd RB, Nordholm L, Lynne D (1985) Investigation of a new motor assessment scale for stroke patients. Phys Ther 65:175–180

Dean JC, Kautz SA (2015) Foot placement control and gait instability among people with stroke. J Rehabil Res Dev 52:577–590

Gelfand IM, Latash ML (1998) On the problem of adequate language in movement science. Mot Control 2:306–313

Hutin E, Pradon D, Barbier F, Gracies JM, Bussel B, Roche N (2011) Lower limb coordination patterns in hemiparetic gait: factors of knee flexion impairment. Clin Biomech 26:304–311

Jackson AB, Carnel CT, Read MS, Boninger ML, Schmeler MR, Williams SR, Donovan WH (2008) Outcome measures for gait and ambulation in the spinal cord injury population. J Spinal Cord Med 31:487–499

Janshen L, Santuz A, Ekizos A, Arampatzis A (2017) Modular control during incline and level walking in humans. J Exp Biol 220:807–813

Kao P, Srivastava S (2018) Mediolateral footpath stabilization during walking in people following stroke. PloS One 13:e0208120

Krishnan V, Rosneblatt NJ, Latash ML, Grabiner MD (2013) The effects of age on stabilisation of the mediolateral trajectory of the swing foot. Gait Posture 38:923–928

Latash ML, Scholz JP, Schoner G (2007) Toward a new theory of motor synergies. Mot Control 11:276–308

Liao Y, Yang Y, Wu Y, Wan R (2014) Factors influencing obstacle crossing performance in patients with Parkinson's Disease. PLoS One 9:e84245

Lowrey CR, Watson A, Vallis LA (2007) Age-related changes in avoidance strategies when negotiating single and multiple obstacles. Exp Brain Res 182:289–299

Lu TW, Yen HC, Chen HL, Hsu WC, Chen SC, Hong SW (2010) Symmetrical kinematic changes in highly functioning older patients post-stroke during obstacle-crossing. Gait Posture 31:511–516

MacLellan MJ, Richards CL, Fung J, McFadyen BJ (2015) Comparison of kinetic strategies for avoidance of an obstacle with either the affected or unaffected as leading limb in persons post stroke. Gait Posture 42:329–334

Manaf H, Justine M, Goh HT (2015) Effects of attentional loadings on gait performance before turning in stroke survivors. PM&R 7:1159–1166

Mattos D, Latash ML, Park E, Kuhl J, Scholz JP (2011) Unpredictable elbow joint perturbation during reaching results in multijoint motor equivalence. J Neurophysiol 106:1424–1436

Moore S, Schurr K, Wales A, Moseley A, Herbert R (1993) Observation and analysis of hemiplegic gait: swing phase. Aus J Physiother 39:271–278

Moseley A, Wales A, Herbert R, Schurr K, Moore S (1993) Observation and analysis of hemiplegic gait: stance phase. Aus J Physiother 39:259–267

Newell KM, Broderick MP, Deutsch KM, Slifkin AB (2003) Task goals and change in dynamical degrees of freedom with motor learning. J Exp Psychol Hum Percept Perform 29:379–387

Novak AC, Deshpande N (2014) Effects of aging on whole body and segmental control while obstacle crossing under impaired sensory conditions. Hum Mov Sci 35:121–130

Olney SJ, Eng J (2011) Gait. In: Levangie PK, Norkin CC (eds) Joint structure and function, 5th edn. Philadelphia, F. A. Davis Company, pp 524–569

Pandy MG, Lin YC, Kim HJ (2010) Muscle coordination of mediolateral balance in normal walking. J Biomech 43:2055–2064

Papi E, Rowe PJ, Pomeroy VM (2015) Analysis of gait within the uncontrolled manifold hypothesis: stabilisation of the centre of mass during gait. J Biomech 48:324–331

Perry J, Burnfield JM (2010) Larger contributions of the hip and knee extensors as well as the stabilizing knee muscles ST and BFL in Syn 1. NJ Slack Inc, Thorofare

Plummer-D'Amato P, Altmann LJP, Saracino D, Fox E, Behrman AL, Marsiske M (2008) Interactions between cognitive tasks and gait after stroke: a dual task study. Gait Posture 27:683–688

Plummer-D'Amato P, Kyvelidou A, Sternad D, Najafi B, Villalobos RM, Zurakowsk D (2012) Training dual-task walking in community-dwelling adults within 1 year of stroke: a protocol for a single-blind randomized controlled trial. BMC Neurol 12:129–137

Qu X (2012) Uncontrolled manifold analysis of gait variability: effects of load carriage and fatigue. Gait Posture 36:325–329

Rankin BL, Buffo SK, Dean JC (2014) A neuromechanical strategy for mediolateral foot placement in walking humans. J Neurophysiol 112:374–383

Rosenblatt NJ, Hurt CP, Latash ML, Grabiner MD (2014) An apparent contradiction: increasing variability to achieve greater precision? Exp Brain Res 232:403–413

Rosenblatt NJ, Latash ML, Hurt CP, Grabiner MD (2015) Challenging gait leads to stronger lower-limb kinematics synergies: the effects of walking within a more narrow pathway. Neurosci Lett 600:110–114

Scholz JP, Schöner G (1999) The uncontrolled manifold concept: identifying control variables for a functional task. Exp Brain Res 126:289–306

Shamay S, Hui-Chan CW (2005) The timed up and go test: its reliability and association with lower-limb impairments and locomotor capacities in people with chronic stroke. Arch Phys Med Rehabil 86:1641–1647

Stanhope VA, Knarr BA, Reisman DS, Higginson JS (2014) Frontal plane compensatory strategies associated with self-selected walking speed in individuals post-stroke. Clin Biomech 29:518–522

Tukoda K, Anan M, Takahashi M, Sawada T, Tanimoto K, Kito N, Shunkoda K (2018) Biomechanical mechanism of lateral trunk lean gait for knee osteoarthritis patients. J Biomech 66:10–17

Van Emmerik REA, Hamill J, McDermott WJ (2005) Variability and coordinative function in human gait. Quest 57:102–123

Winter DA (1987) The biomechanics and motor control of human gait. University of Waterloo Press, Waterloo

Wong AM, Pei YC, Hong WH, Chung CY, Lau YC, Chen CP (2004) Foot contact pattern analysis in hemiplegic stroke patients: an implication for neurologic status determination. Arch Phys Med Rehabil 85:1625–1630

Wu Y, Pazin N, Zatsiorsky VM, Latash ML (2012) Practicing elements versus practicing coordination: changes in the structure of variance. J Mot Behav 44:471–478

Yang JF, Scholz JP (2005) Learning a throwing task is associated with differential changes in the use of motor abundance. Exp Brain Res 163:137–158

Zhang W, Scholz JP, Zatsiorsky VM, Latash ML (2008) What do synergies do? Effects of secondary constraints on multi-digit synergies in accurate force-production tasks. J Neurophysiol 99:500–513