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Adaptive changes in foot placement for split-belt treadmill walking in individuals with stroke

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ABSTRACT

Background: Adaptation to split-belt treadmill walking differs between individual stroke survivors. Many discussions only address spatiotemporal parameters that are related to movement, and the changes in interlimb spatiotemporal parameters as a consequence of adaptation are poorly understood.
Objectives: To investigate symmetry of the center of pressure (CoP) position relative to the center of mass (CoM), and ascertain whether this can be used to identify differences in adaptation of interlimb spatiotemporal parameters among stroke survivors during split-belt walking.
Methods: Twenty-two chronic post-stroke patients and nine elderly controls walked in tied- then split-belt (2:1 ratio of fast:slow) conditions. Spatiotemporal parameters were compared within groups to assess symmetry of the CoM-CoP angle at foot contact.
Results: Asymmetry of the CoM-CoP angle was associated with asymmetry of spatiotemporal parameters. Re-establishment of symmetry of CoM-CoP angle was reflected in re-established symmetry of spatiotemporal parameters in post-stroke and control participants.
Conclusions: Post-stroke patients who re-establish symmetry of the COM-COP angle are able to adapt their walking for split-belt perturbation. This suggests that predictively symmetric foot placements on the fast and slow sides are necessary for adaptation in walking. Symmetrical foot placement is achieved by interlimb coordination and may contribute to dynamic stability.

1. Introduction

Stroke survivor typically walk asymmetrically, and assessment of spatiotemporal and kinetic variables in such individuals will reveal differences between the paretic and non-paretic limbs. Examples of asymmetrical variables include stance and swing times (Wall and Turnbull, 1986), double-support time (Olney et al., 1994), joint power (Knutsson and Richards, 1979), joint excursion (Olney et al., 1991), and step length (Allen et al., 2011). Asymmetry of individuals with stroke is thought to be caused by various obstacles to locomotion (Patterson et al., 2008, 2010), and split-belt treadmill training has recently emerged to induce the learning of a new walking pattern with reduced asymmetry of step length and double-support time (Reisman et al., 2007, 2013; Betschart et al., 2017; Miéville et al., 2018). This is achieved by improvement in interlimb coordination due to patients' adaptability (Tyrell et al., 2015; Wutzke et al., 2015).

Split-belt treadmills use separate double belts to drive the legs at different speeds, and the adaptation protocol is used to demonstrate the adaptability of human bipedal locomotion by time-series changes in the symmetry or asymmetry of gait spatiotemporal parameters (Dietz et al., 1994; Choi and Bastian, 2007). During split-belt walking, interlimb (i.e., step length, double-support time) and intralimb (i.e., stride length, stance time) spatiotemporal parameters show different changes. Interlimb parameters gradually adapt and re-establish symmetry, whereas intralimb parameters adjust merely reactively, and asymmetry is maintained (Reisman et al., 2005).

However, not all individuals with stroke completely re-establish symmetry during split-belt walking. Previous studies have reported that

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the magnitude (Savin et al., 2013), time (Tyrell et al., 2015), and reaction manner (Reisman et al., 2007; Malone and Bastian, 2014) of interlimb spatiotemporal parameters are different depending on the individual patient and belt conditions (i.e., paretic side slow or fast). In these studies that dealt with spatiotemporal parameters, the causes of differences in the adaptation of interlimb parameters following split-belt walking remain unclear. This is because spatiotemporal parameters product as result of movement. Kinetic analysis of gait adaptation in split-belt conditions may provide clarification of these causes. For adaptation during split-belt treadmill walking, adequate control of the position of the center of mass (CoM) contributes to stability during locomotion (Morton and Bastian, 2006; Mawase et al., 2013; Jansen et al., 2013). Moreover, studies of split-belt treadmill training in individuals with stroke have shown that foot placement affects dynamic stability (Hak et al., 2013; Reisman et al., 2013; Miéville et al., 2018). We hypothesized that differences in the adaptation of interlimb parameters are related to dynamic stability in individuals with stroke.

Dynamic stability during walking relies on the interactions that affect foot placement, and the positions of the CoM and center of pressure (CoP) at the moment of foot contact are crucial (Hsue et al., 2009). In our preliminary study of the inverted pendulum model of CoM and CoP, healthy subjects re-established symmetry of leading-foot placement relative to the whole body at foot contact during split-belt walking (Hirata et al., 2019a). The present study aimed to verify that symmetry of the CoP position relative to the CoM can indicate differences in adaptation of interlimb spatiotemporal parameters in individuals with stroke during split-belt walking. Moreover, we use the peak braking force of the leading leg (required coefficient of friction, RCOF), which occurs shortly after ground contact by the leading foot, to evaluate dynamic stability (Yamaguchi and Masani, 2016). The RCOF is an important factor with regards to slipping during walking (Nagano et al., 2013). This verification could add kinetic explanations of the differences that are observed in the dynamic stability adaptability of individuals with stroke.

2. Methods

2.1. Participants

Twenty-two individuals who had sustained a single stroke more than 6 months prior to the study (mean age: 67.0 years, standard deviation (SD): 9.3; 17 females, Table 1), nine age- and gender-matched healthy elderly control subjects (0.4 matched controls per case) were recruited to participate in the study. Exclusion criteria were: other neurological conditions, orthopedic conditions affecting the legs or back, uncontrolled hypertension, pacemaker or automatic defibrillator fitted, active cancer, evidence of damage to the cerebellum by radiological and/or physical examination, or inability to complete the task. Subjects who customarily wear an ankle-foot orthosis were allowed to wear it during testing. The experiments were explained in detail to participants, and written informed consent obtained. This study and its protocols were approved by the Saitama prefectural university ethics review committee (No. 29501).

2.2. Clinical evaluation

Prior to the experiments, participants walked on the treadmill to determine their comfortable and maximum walking speeds. Lower-limb motor recovery was evaluated with the lower extremity portion of the Fugl-Meyer Assessment (LE FMA) (Fugl-Meyer et al., 1975). The Timed Up and Go test (Ng and Hui-Chan, 2005) and Functional Independence Measure (FIM) were also carried out.

2.3. Experimental protocol

The split-belt treadmill protocol followed a published protocol (Hirata et al., 2019b) (Fig. 1A). Post-stroke and healthy elderly participants walked on the treadmill under the following conditions: (1) "tied-belt" at half-maximum walking speed for 3 min, then (2) "split-belt" with the slow belt at half-maximum and fast belt at maximum walking speed (2:1 ratio of fast:slow) for 6 min. Individuals with stroke performed the task twice. First, the non-paretic leg was assigned to the fast belt during split-belt conditions (denoted "paretic leg slow"). Second, the paretic leg was assigned to the fast belt (denoted "paretic leg fast"). One trial was performed in each session. For elderly control participants, the fast and slow belts were randomly assigned to either the right or left leg.

Table 1
Post-stroke patient characteristics (n = 22).

Subject	Age	Sex	Hemiparetic side	LE FMA (/34)	Months since stroke	Speed slow/fast (m/s)	FIM	TUG (s)
S1	68	M	R	21	128	0.25/0.5	115	17.8
S2	63	F	L	20	27	0.3/0.6	115	31.1
S3	69	M	R	29	142	0.45/0.9	126	7.0
S4	75	M	R	22	120	0.35/0.7	115	15.8
S5	70	F	R	21	65	0.3/0.6	120	14.0
S6	79	M	L	31	13	0.2/0.4	105	12.0
S7	73	M	L	26	36	0.25/0.5	120	16.0
S8	78	M	R	34	9	0.4/0.8	126	10.4
S9	65	M	L	25	7	0.35/0.7	120	14.0
S10	71	M	R	20	18	0.2/0.4	114	18.8
S11	73	M	R	22	8	0.2/0.4	107	25.9
S12	70	M	L	19	141	0.2/0.4	104	45.1
S13	57	M	L	19	18	0.2/0.4	118	30.3
S14	45	F	L	21	66	0.2/0.4	114	15.8
S15	48	F	L	14	132	0.2/0.4	119	28.6
S16	84	M	R	29	7	0.2/0.4	121	24.8
S17	71	M	R	27	162	0.2/0.4	109	21.8
S18	70	M	R	20	33	0.2/0.4	106	32.5
S19	63	M	L	28	102	0.2/0.4	83	46.9
S20	65	F	L	28	47	0.2/0.4	121	21.8
S21	69	M	R	25	50	0.2/0.4	102	27.0
S22	55	M	L	14	103	0.2/0.4	116	37.7

Abbreviations: F, female; FIM, functional independence measure; M, male; L, left; LE FMA, lower extremity Fugl-Meyer Assessment; R, right; TUG, Timed Up and Go test.

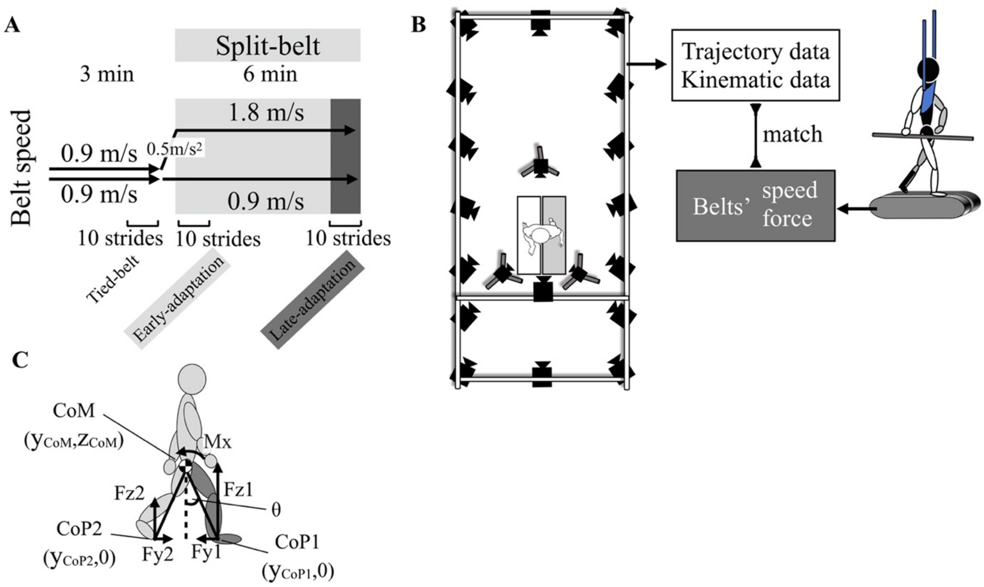

Fig. 1. (A) Illustration of the split-belt walking protocol. (B) The motion capture system and split-belt treadmill. Kinematic and trajectory data as well as belt speed and force data were synchronized. (C) The bipedal inverted pendulum model. Abbreviations: CoM, whole-body center of mass; yCoM (zCoM), anterioposterior (vertical) position of the CoM; Mx, moment around the CoM; Fy1/Fz1 (Fy2/Fz2), braking force for the leading (trailing) leg; CoP1 (CoP2), center of pressure of the leading (trailing) leg; yCoP1 (yCoP2), anterioposterior position of the CoP of the leading (trailing) leg; θ, the CoM-CoP angle of the leading leg.

2.4. Data collection

A motion-capture system involving 22 cameras (Vicon Motion Systems, Oxford, UK, Fig. 1B) was used to capture three-dimensional marker data and calculate the CoM during the task. Marker placement followed the Plug-in-Gait full-body AI model. The ground reaction force (GRF) exerted on the instrumented double-belt treadmill (Bertec Corp., OH, USA) and belt speeds were recorded bilaterally (1000 Hz) and resampled to match the kinematic data (100 Hz). Recording of kinematic and kinetic signals was conducted for 30 s at the end (150–180 s) of the tied-belt period, and at the beginning (180–210 s, defined as “early-adaptation”) and end (510–540 s, defined as “late-adaptation”) of the split-belt period. We calculated each CoP from the recorded GRFs and moments.

2.5. Data analysis

2.5.1. Spatiotemporal parameters and center of mass-center of pressure angle

We calculated the following spatiotemporal gait parameters: (1) Percent stance time, defined as the duration from foot contact to toe off, expressed as a percentage of stride time; (2) Step length, defined as the anteroposterior distance between the ankle markers of each leg at foot contact; (3) Percent double-support time, defined as the duration from foot contact on one side to toe off on the other side, expressed as a percentage of stride time. The angle between a vertical line (against-earth vertical) and the vector from the CoM to the CoP of the leading leg on a sagittal plane at foot contact on the slow and fast sides was calculated (θ, Fig. 1C). Based on a previous study (Malone and Bastian, 2014), we calculated the symmetrical indices of slow- and fast-side data for all parameters using Eq. (1):

$$Symmetrical \quad index = \frac{fast \quad side - slow \quad side}{fast \quad side + slow \quad side} \times 100 \quad (\%) \tag{1}$$

2.5.2. Kinetic parameters

The tangent of the CoM-CoP angle is strongly correlated with the peak braking force of the leading leg (required coefficient of friction, RCOF) during walking without instability (Yamaguchi and Masani, 2016). Using bipedal inverted pendulum modeling with the CoM and CoP in the sagittal plane (Fig. 1C), Yamaguchi et al. estimated the CoM moment (Eq. (2)):

$$Mx = Fz1(y_{COP1} - y_{COM}) + Fz2(y_{COP2} - y_{COM}) - Fy1{*}Z_{COM} + Fy2{*}Z_{COM} \tag{2}$$

From this, the braking force for the leading leg (Fy1/Fz1) can be defined by Eqs. (3) and (4):

$$\frac{Fy1}{Fz1} = \tan\theta + \frac{Fz2}{Fz1}{*}\frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} + \frac{Mx}{Fz1{*}Z_{COM}} \tag{3}$$

$$RT = \frac{Fz2}{Fz1}{*}\frac{y_{COP2} - y_{COM}}{Z_{COM}} + \frac{Fy2}{Fz1} + \frac{Mx}{Fz1{*}Z_{COM}} \tag{4}$$

Eqs. (3) and (4) separate the tangent of CoM-CoP angle of the leading leg (tan θ) and terms relating to the effect of the trailing leg (residual term (RT)). Yamaguchi et al. clarified that the peak braking force of the leading leg (Fy1/Fz1) is approximately equal to tan θ at the time of peak braking force on the leading leg (Yamaguchi and Masani, 2016).

We used these kinetic parameters to explain the differences in the dynamic stability of individuals with stroke who either did or did not show adaptation of spatiotemporal parameters. We assessed the difference between the braking force for the leading leg (Fy1/Fz1) and tan θ during walking adaptation.

2.6. Subgroup classification for individuals with stroke

Individuals with stroke were classified as “responders” or “non-responders” by the change in CoM-CoP angle symmetry during late-adaptation. We determined the threshold for classification as a “responder” by the mean symmetrical index during late-adaptation in elderly control subjects.

2.7. Statistics

We compared each parameter of the last 10 strides of the tied-belt and late-adaptation periods and the first 10 strides of the early-adaptation period of each treadmill session. The symmetrical index of spatiotemporal parameters normalized to tied-belt conditions (i.e., early- and late-adaptation minus the tied-belt period) was analyzed to evaluate the change from the tied-belt period. Within-subjects one-way repeated measures analysis of variance (ANOVA) was used to compare the symmetry of spatiotemporal parameters among the different periods (tied-belt, early-adaptation, and late-adaptation). Two-way repeated measures ANOVA was used to identify statistically significant interactions in the normalized index between periods and groups. If ANOVA revealed significant main effects or interactions, Bonferroni's

Table 2
Comparisons of parameters between responders and non-responders for paretic leg fast and slow settings.

	Paretic leg slow			Paretic leg fast		
	Responder (n = 8)	Non-responder (n = 14)	p Value	Responder (n = 5)	Non-responder (n = 17)	p Value
LE FMA (/34)	22.2 (7.0)	23.8 (4.6)	0.52	21.2 (7.1)	23.9 (5.0)	0.35
TUG (s)	21.9 (10.7)	24.3 (11.1)	0.63	22.0 (10.1)	23.8 (11.2)	0.75
FIM (/124)	114.7 (7.8)	112.7 (10.8)	0.64	111.4 (5.0)	114.0 (10.7)	0.60
Maximum walking speed (m/s)	0.5 (0.2)	0.5 (0.1)	0.38	0.5 (0.1)	0.5 (0.1)	0.59
Symmetrical index of percent stance time during tied-belt period	7.9 (5.5)	6.3 (5.9)	0.54	8.4 (4.1)	7.0 (5.3)	0.49
Symmetrical index of step length during tied-belt period	32.7 (30.4)	36.3 (29.0)	0.70	50.9 (33.8)	31.7 (29.9)	0.44
Symmetrical index of percent double-support time during tied-belt period	18.9 (12.1)	13.6 (11.8)	0.16	16.2 (15.0)	15.1 (9.4)	0.18
Symmetrical index of CoM-CoP angle during late-adaptation period	0.3 (0.2)	2.4 (2.1)	< 0.01	0.4 (0.2)	2.7 (2.2)	< 0.05

Data are presented as number (%). Abbreviations: CoM, center of mass; CoP, center of pressure; RIM, functional independence measure; LE FMA, lower extremity Fugl-Meyer Assessment; TUG, Timed Up and Go test.

post-hoc comparisons were carried out to identify significant differences among variables. Paired t-tests were used to evaluate the difference between the peak braking force of the leading leg and tan θ during late-adaptation within subgroups. To evaluate the difference in the symmetry of CoM-CoP angle during late-adaptation between subgroups, we used non-paired t-tests among subgroups. All analyses were performed with a significance level set at $p = 0.05$, and using MATLAB (MathWorks Inc., MA, USA).

3. Results

3.1. Center of mass-center of pressure angle and classification of individuals with stroke

Table 1 shows the characteristics of individuals with stroke. The mean symmetrical index of CoM-CoP angle during late-adaptation was 0.78 in elderly control participants. Table 2 details the recorded values for each parameter in the post-stroke subgroups, determined using the threshold of 0.78. A significant difference was seen in the symmetry of the CoM-CoP angle between subgroups ($p < 0.05$), although no other parameters or functions were significantly different.

Fig. 2A illustrates the changes in the CoM-CoP angle observed for each side during the testing period at foot contact for responders, non-responders, and elderly control participants. Responders and elderly control subjects established symmetry of the CoM-CoP angle through split-belt walking, while non-responders remained asymmetrical. Fig. 2B is a representative example of the CoP profile relative to CoM position in the horizontal plane during late-adaptation. At foot contact, the foot placement of responders was symmetrical, while that of non-responders was asymmetrical.

Furthermore, the correlation coefficients between the symmetry of CoM-CoP angle and symmetry of step length were low during each period (Table 3). The symmetry of the CoM-CoP angle was not a spatial variable that is dependent on the symmetry of step length.

3.2. Spatiotemporal parameters

Fig. 3 shows the changes in the symmetrical indices of stance time, step length, and double-support time at different time points during the testing period. For all participants, the symmetrical index for stance

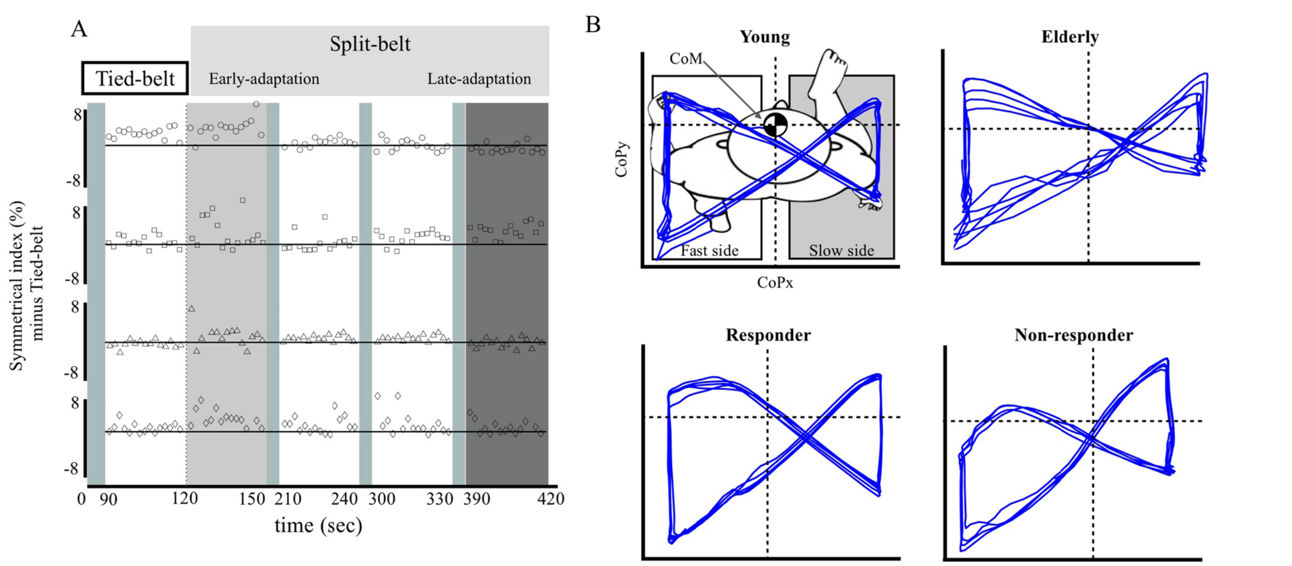

Fig. 2. (A) Symmetrical index of center-of-mass (CoM)-center-of-pressure (CoP) angle measurements for sequential strides from representative elderly healthy controls (top row) and individuals with stroke (middle row, responder; bottom row, non-responder) across all testing periods. (B) The CoP profile relative to CoM for the same representative subjects as in (A); representative elderly controls (top right), and individuals with stroke (bottom left, responder; bottom right, non-responder) during the same cycle of late-adaptation. Definitions: CoM, centre of mass; CoPx, mediolateral position of the CoP; CoPy, anterioposterior position of the CoP.

Table 3
Correlation coefficients between the symmetry of CoM-CoP angle and that of step length.

	Baseline	Early-adaptation	Late-adaptation
Post-stroke in paretic leg slow	R = 0.39, p = 0.07	R = 0.28, p = 0.21	R = 0.38, p = 0.08
Post-stroke in paretic leg fast	R = 0.48, p = 0.10	R = 0.49, p = 0.07	R = 0.39, p = 0.07
Elderly	R = −0.31, p = 0.41	R = 0.31, p = 0.41	R = −0.49, p = 0.17

time was significantly different during early- and late-adaptation to those recorded in the tied-belt period ($p < 0.05$). Stance time remained asymmetrical in the split-belt period. In the responder and elderly control groups, the symmetrical index of step length in early-adaptation showed significant differences to the tied-belt values ($p < 0.001$). No significant difference in this parameter was seen between early and late-adaptation ($p = 0.53$). Symmetry of step length was re-established during split-belt walking in the responder and elderly control groups, and there were no interactions between the period and group (F (2, 42) = 0.17, $p = 0.85$, and F (2, 36) = 0.15, $p = 0.86$). Similarly, symmetry of the double-support time was also re-established through split-belt walking in the non-responder and elderly control groups. Double-support times of these two groups indicated that there were no interactions between the period and group (F (2, 42) = 0.32, $p = 0.73$, and F (2, 36) = 0.36, $p = 0.51$). On the other hand, the step length and double-support time of non-responders were significantly different in late-adaptation compared with the tied-belt period ($p < 0.01$). Step lengths of the non-responder and elderly groups indicated that there were significant interactions between the period and group (F (2, 66) = 5.67, $p = 0.01$, and F (2, 72) = 3.73, $p = 0.03$). The stance time remained asymmetrical in the non-responder group; however, the double support time of the non-responder and elderly groups indicated that there were no interactions between the period and group (F (2, 66) = 2.40, $p = 0.09$, and F (2, 72) = 2.50, $p = 0.09$).

3.3. Kinetic parameters

Fig. 4 shows the difference between Fy1/Fz1 and tan θ during late-adaptation for individuals with stroke. No significant differences were observed between the fast and slow sides among responders (paretic leg slow $p = 0.30$ and 0.65, paretic leg fast $p = 0.11$ and 0.13, respectively). In the non-responder group, there was a significant difference between the fast and slow sides in the paretic leg fast setting ($p < 0.05$). In the non-responder group, there was no significant difference between the values for the fast and slow sides in the paretic leg slow setting ($p = 0.05$). All three terms in the calculation of RT (Eq. (4)) relate to the effect of the trailing leg. Fig. 5 shows the relationship between the first and second terms on the slow and fast sides. The elderly control group showed a negative correlation between these terms (Fig. 5A, R = −0.78; $p < 0.01$), which was also observed for

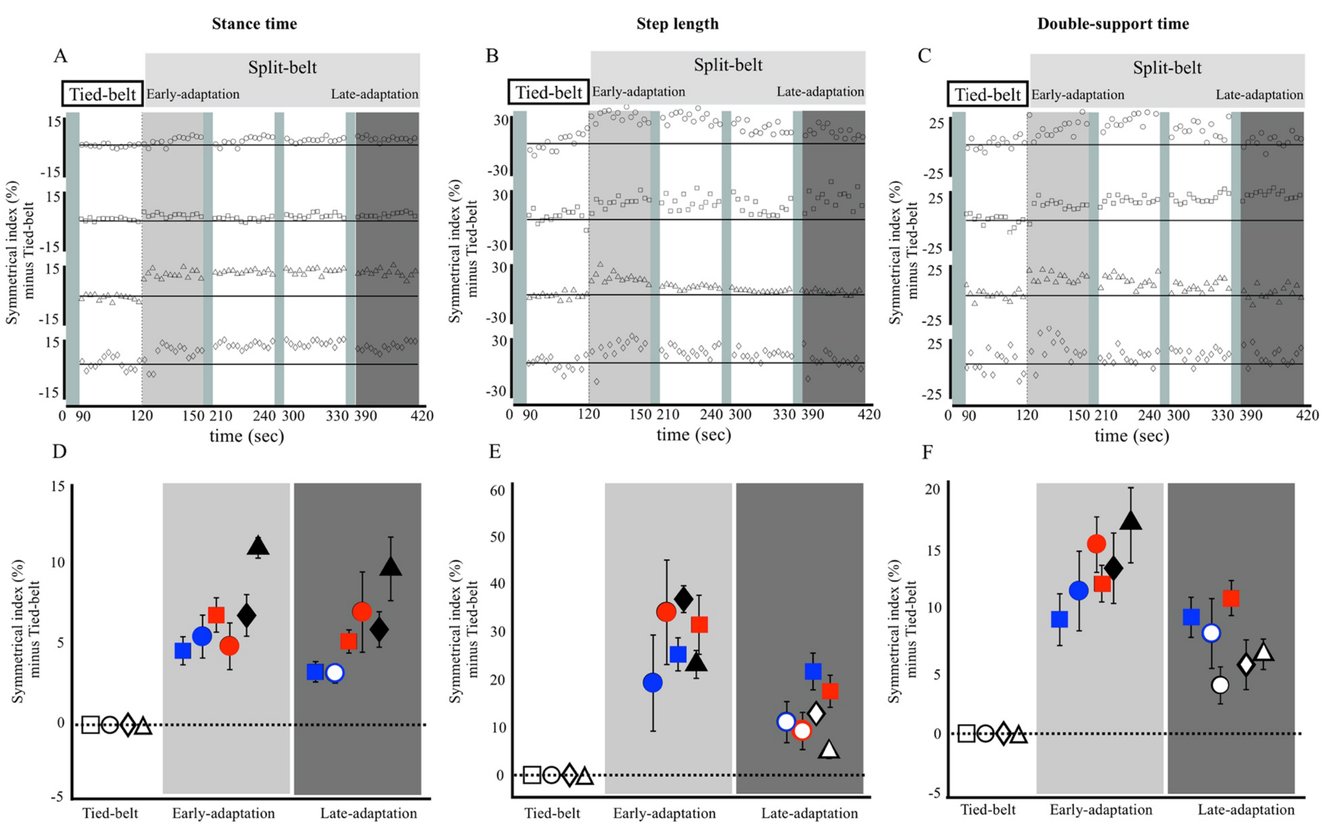

Fig. 3. The symmetrical index of (A) stance time, (B) step length, and (C) double-support time measurements for sequential strides from representative elderly healthy controls (top row) and individuals with stroke (middle row, responder; bottom row, non-responder) across all testing periods. (D) Stance time, (E) step length, and (F) double-support time differences for post-stroke (red and blue circles represent responders with paretic leg slow and fast settings, respectively; red and blue squares represent non-responders with paretic leg slow and fast settings, respectively) and elderly (diamond) groups. Each data point indicates the difference from the tied-belt period (early- or late-adaptation minus the tied-belt data), averaged across all subjects in the group. Error bars indicate mean standard error. Filled objects indicate statistically significant differences compared with tied-belt values. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

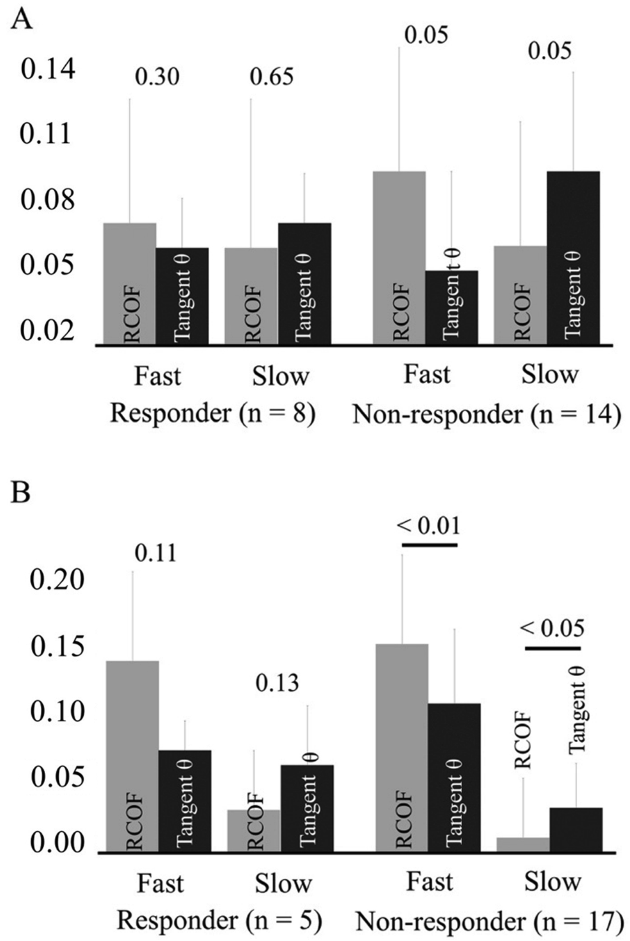

Fig. 4. Comparisons between required coefficient of friction (gray) and tan θ (black) values of responders and non-responders for fast and slow belt sides in (A) paretic leg slow and (B) paretic leg fast settings. Error bars indicate mean standard deviation. Horizontal bars indicate significant differences ($p < 0.05$). Abbreviations: RCOF, required coefficient of friction.

individuals with stroke (Fig. 5B, C, $R = -0.74$–-0.72; $p < 0.01$). Thus, the first and second terms almost cancel each other out.

4. Discussion

We analyzed the differences in adaptation of interlimb spatiotemporal parameters by evaluating the symmetry of the CoP position relative to CoM in individuals with stroke during split-belt walking. We found that symmetry of the CoM-CoP angle is associated with symmetry of spatiotemporal parameters. This suggests that predictively symmetric foot placements on the fast and slow sides are necessary for adaptation in walking to achieve high dynamic stability.

4.1. Difference in adaptability of gait parameters between post-stroke subgroups

Classification by symmetry of the CoM-CoP angle during split-belt treadmill walking meant that adaptabilities of spatiotemporal parameters differed between subgroups. For step length and double-support time, the interlimb parameters of non-responders who did not re-establish symmetry of the CoM-CoP angle remained asymmetrical, as did the intralimb parameters. On the other hand, individuals with stroke who re-established symmetry of the CoM-CoP angle through split-belt walking also re-established baseline symmetry of interlimb parameters, as did the matched control participants. In healthy humans, symmetry of interlimb parameters is gradually re-established by trial-and-error during split-belt treadmill walking (Reisman et al., 2005). However, responders did not completely re-establish symmetry of interlimb parameters, in line with a previous study (Reisman et al., 2013). Our results support those of a previous study that suggested an asymmetric walking pattern could be the optimal pattern for certain individuals, as a byproduct of asymmetries in neural control or limb mechanics (Finley and Bastian, 2017). Our results clarify that complete re-establishment of CoM-CoP angle symmetry through split-belt walking depends on the adaptability of spatiotemporal parameters in individuals with stroke. Analyses of GRF (Ogawa et al., 2014) and the phase of gait cycle (Fujiki et al., 2015) have identified foot contact as an important element in adaptation in split-belt walking. Therefore, differences in spatiotemporal parameters suggest that the position of foot contact represents an important kinetic contribution to adaptation.

4.2. Kinetic factors in the symmetry of center of mass-center of pressure angle

Bipedal inverted pendulum modeling with the CoM and CoP revealed that the coincidence between the CoM-CoP angle (i.e., tan θ) and the peak vector of braking force (i.e., Fy1/Fz1) is related to the difference in CoM-CoP angle symmetry in individuals with stroke. The peak braking force of the leading leg, calculated from the vertical and anteroposterior components of the GRF, usually occurs shortly after ground contact by the leading foot. This is an important factor with regards to slipping during walking, as it must not exceed the RCOF (Nagano et al., 2013). Peak braking forces of the leading leg during straight and turning gaits are strongly positively correlated with tan θ (Yamaguchi et al., 2012; Yamaguchi and Masani, 2016). In the responder group, tan θ and Fy1/Fz1 were consistent, as is observed in healthy people, whereas non-responders showed significant differences between tan θ and Fy1/Fz1. However, the first and second terms of the RT almost canceled each other in all participants, indicating that the RT directly reflects the third term in Eq. (4), which defines the moment around the CoM. Therefore, in non-responders whose RT was not 0, the moment around the CoM from the leading and trailing legs was not canceled at the time of RCOF. The difference between tan θ and the RCOF is increased when dynamic stability is decreased; for example, during slow walking (Yamaguchi et al., 2018). Therefore, individuals with stroke who do not re-establish symmetry of the CoM-CoP angle might experience low dynamic stability in split-belt walking.

Bipedal locomotion is governed primarily by passive dynamics with minimal active energy costs (Collins et al., 2005). Foot contact of the leading leg represents negative work, which produces a backward rotational moment around the CoM in the sagittal plane from the leading leg. In an ideal dynamic walking model, this negative work can be reduced by pushing off with the trailing leg just before foot contact (Kuo, 2002). Kuo et al indicate that the trailing leg reduces the amount of active work that is required to predictively sustain steady gait (Kuo and Donelan, 2010). Therefore; at foot contact, the interaction between the leading and trailing legs (interlimb coordination) is also key in stabilization of the whole body.

4.3. Limitations and future study

This study had some limitations. First, our protocol did not include a tied-belt condition in the post-adaptation period. Assessments of the parameters after split-belt walking could provide evidence of motor learning. However, as our main goal was to provide information about the changes during adaptation in split-belt walking, this was not carried out. Therefore, we analyzed adaptive behavior through split-belt treadmill walking and do not discuss motor learning in this study.

Second, we allowed individuals with stroke to use the handrail for support and safety. This has been suggested to affect symmetry in such patients (IJmker et al., 2015; Finley and Bastian, 2017). However, the

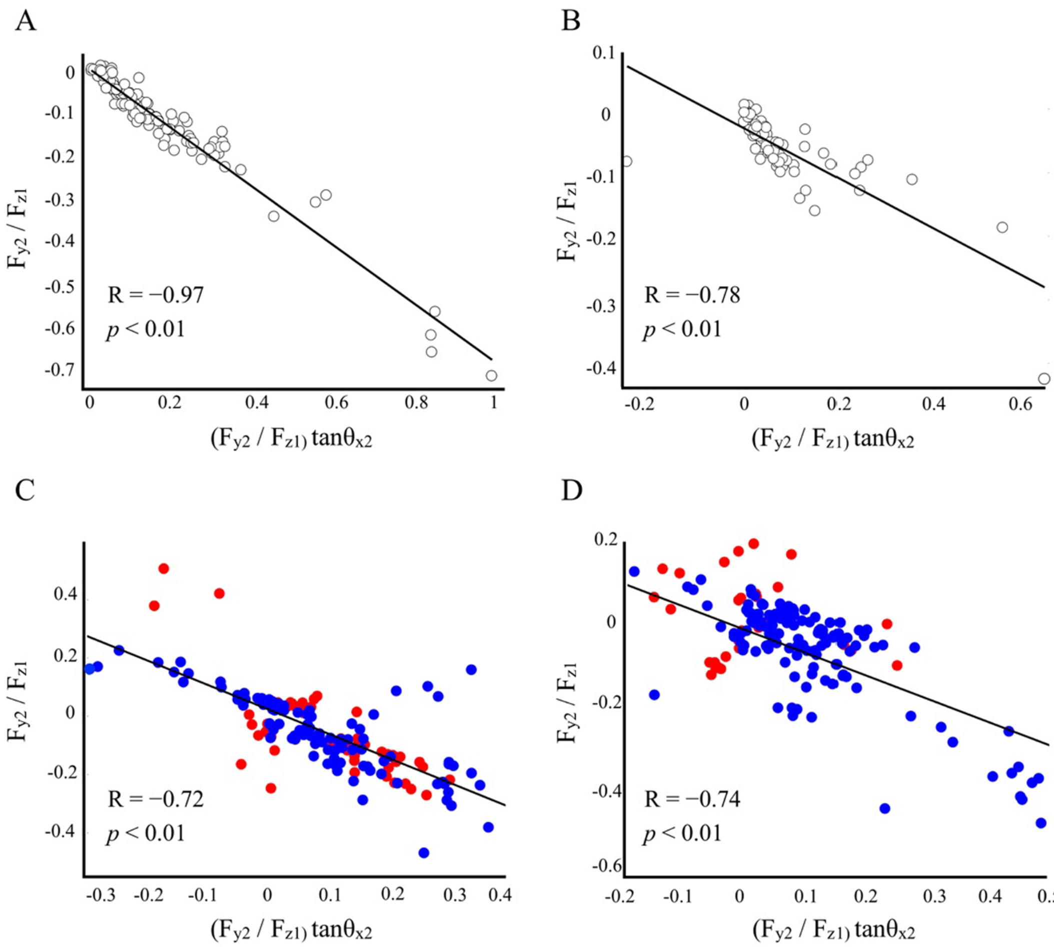

Fig. 5. Graphs of the negative relationship between the first and second terms of RCOF in Eq. (3) for (A) elderly healthy controls, (B) individuals with stroke under paretic leg slow conditions, and (C) individuals with stroke under paretic leg fast conditions. In C and D, red and blue dots represent data from responders and non-responders, respectively. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

activity levels of our individuals with stroke were, in general, lower than those of previous studies. The novelty of this study was the assessment of differences in patients with low levels of activity. Also, we did not observe a statistically significant effect of split-belt walking in patients of the non-responder group. This was particularly notable in the case of double-support time, and suggests that the classification of subjects may not always have been correct.

Finally, no differences were observed in the clinical evaluations or data of other clinical measures relating to dynamic stability among post-stroke subgroups. This limits the clinical application of our findings regarding the relationship between dynamic stability and gait adaptability. On the other hand, a previous study has highlighted the need to interpret such findings with caution, as high scores in evaluations of dynamic stability may not be associated with high performance in clinical balance assessments in post-stroke individuals (Nott et al., 2014; Vistamehr et al., 2016). The present study is the first to explain the relationship between gait adaptability and dynamic balance during split-belt treadmill walking among post-stroke individuals. Future studies involving larger sample sizes and measurements of more clinical parameters are required.

4.4. Conclusions

Our study shows that individuals with stroke who re-establish symmetry of the CoM-CoP angle at foot contact can adapt their walking for split-belt perturbation. This suggests that predictively symmetric foot placements on the fast and slow sides are necessary for adaptation in walking. Reisman et al argue that adapting the angle of the leading leg at weight transfer is critical to control the relative position between the CoM and the leading leg (Reisman et al., 2005). Symmetrical foot placement is achieved by interlimb coordination of both legs, and may contribute to dynamic stability. Because patients who have no limitations in forward placement of the paretic leg are able to use the split-belt treadmill, this may be one of the criteria for split-belt training in post-stroke patients.

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Declaration of Competing Interest

None of the authors have any conflicts of interest associated with this study.

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Appendix A. Supplementary material

Supplementary data to this article can be found online at https://

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