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Hemiparetic Stepping to the Beat: Asymmetric Response to Metronome Phase Shift During Treadmill Gait

Trudy A. Pelton, MRes,[1] Leif Johannsen, PhD,[2] Huiya Chen, PhD,[2] and Alan M. Wing, PhD[1]

Abstract

Background. Walking in time with a metronome is associated with improved spatiotemporal parameters in hemiparetic gait; however, the mechanism linking auditory and motor systems is poorly understood. *Objective*. Hemiparetic cadence control with metronome synchronization was examined to determine specific influences of metronome timing on treadmill walking. *Methods*. A within-participant experiment examined correction processes used to maintain heel strike synchrony with the beat by applying perturbations to the timing of a metronome. Eight chronic hemiparetic participants (mean age = 70 years; standard deviation = 12) were required to synchronize heel strikes with metronome pulses set according to each individual's comfortable speed (mean 0.4 m/s). During five 100-pulse trials, a fixed-phase baseline was followed by 4 unpredictable metronome phase shifts (20% of the interpulse interval), which amounted to 10 phase shifts on each foot. Infrared cameras recorded the motion of bilateral heel markers at 120 Hz. Relative asynchrony between heel strike responses and metronome pulses was used to index compensation for metronome phase shifts. *Results*. Participants demonstrated compensation for phase shifts with convergence back to pre–phase shift asynchrony. This was significantly slower when the error occurred on the nonparetic side (requiring initial correction with the paretic limb) compared with when the error occurred on the paretic side (requiring initial nonparetic correction). *Conclusions*. Although phase correction of gait is slowed when the phase shift is delivered to the nonparetic side compared with the paretic side, phase correction is still present. This may underlie the utility of rhythmic auditory cueing in hemiparetic gait rehabilitation.

Keywords

hemiparetic stroke, metronome, synchronization, walking, correction

Introduction

Hemiparetic gait is associated with spatiotemporal changes, which include decreased walking speed, reduced overall stride length, and asymmetrical step amplitude and timing.[1] Published research suggests that overground walking in time with a metronome[2,3] or musical beat[4] can improve spatiotemporal parameters in hemiparetic gait and reduce gait asymmetry. Moreover, treadmill walking combined with acoustic pacing has been shown to result in increased symmetry of step length in hemiparetic gait.[5] Based on step responses to rhythm perturbations,[6] recent work[7] provides qualitative estimates of recovery functions in auditory-motor coordination poststroke. We extend previous research and provide quantitative analysis of the hemiparetic step response to perturbations in acoustic pacing.

Synchronization with a metronome involves control over the asynchrony between the time of the response relative to the time of the beat (phase), and control of the interval between successive responses (period).[8] In studies of synchronization involving finger tapping, a first-order linear feedback model, in which a proportion of the discrepancy between metronome pulse and associated tapping response is used to adjust the phase of the next response, explains the maintenance of phase.[9] One method used to

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[1]University of Birmingham, Birmingham, United Kingdom
[2]Chung Shan Medical University, Taichung, Taiwan

Corresponding Author:
Alan M. Wing, School of Psychology, University of Birmingham, Edgbaston, Birmingham B15 2TT, UK
E-mail:a.m.wing@bham.ac.uk

Table 1. Demographics and Clinical Features of the Stroke Participants

Stroke Participant	1	2	3	4	5	6	7	8
Age (years)	71	52	76	65	91	58	68	81
Gender	M	F	M	F	M	M	F	M
Time since lesion (months)	63	108	18	46	9	36	37	15
Lesioned hemisphere	L	L	R	R	R	R	L	L
Lesioned brain structure	PL, PG	FL	N/A	FL, BG, CI	N/A	BG, CI	TH, BG, CI	CI
FM paretic leg (maximum 34)	32	33	29	30	28	29	29	28
RMA GF (maximum 13)	10	10	10	11	10	11	10	7
Walking Speed (m/s)	0.53	0.59	0.32	0.34	0.51	0.51	0.33	0.41
IPI	640	740	740	680	720	940	540	880

Abbreviations: F, female; M, male; L, left hemisphere/right paretic; R, right hemisphere/left paretic; PL, parietal lobe; PG, precentral gyrus; FL, frontal lobe; BG, basal ganglia; CI, internal capsule; TH, thalamus; FM paretic leg, Fugl-Meyer lower-extremity score; RMA GF, Rivermead Motor Assessment Gross Function Scale; N/A, no scan available but presumed capsular lesion from symptomatology; IPI, interpulse interval.

investigate this feedback correction process is to unexpectedly perturb the phase of the metronome and examine how the resulting phase error is progressively reduced over the following sequence of responses.[10] Phase shift compensation functions produced by trained healthy normal participants have revealed that even the smallest change in metronome phase, well below the threshold for awareness, can induce corrections.[11]

It has been suggested that the linear phase correction model described for finger tapping also applies to gait. Compensation functions for 10% phase shifts in a metronome with a period of 500 ms were presented for healthy adult participants tapping with the feet (while seated) and stepping on the spot (while standing).[6] Correction was slower in stepping while standing. It was suggested that the increased demand of maintaining balance resulted in slowed correction. This, in turn, caused deterioration in synchronization performance while the added demand of bilateral coordination increased the variability of synchronization.

The stepping synchronization study identified the potential of using phase shifts to analyze the mechanisms underlying metronome-assisted pathological gait. Such an approach was used with auditory paced treadmill walking in hemiparetic stroke.[7] The main focus was to show improved performance with the provision of a metronome beat on every step compared with alternate steps. The study also included qualitative analysis of phase shift trials, which indicated correction in hemiparetic stroke participants, albeit less reliable compared with controls. The present study was designed to provide quantitative estimates of phase shift correction after hemiparetic stroke and to contrast correction when the phase shift coincided with affected and nonaffected stance. It was expected that correction might be slower when the phase shift occurred on the unaffected side (requiring the first adjustment step by the affected limb during the swing phase) compared with phase shift on the affected side (requiring initial adjustment by the unaffected limb). Further understanding of the potential for the metronome to implement temporal changes to walking following stroke may lead to more effective use of auditory cues in gait rehabilitation.

Methods

Participants

Eight community dwelling adults (of whom 3 were women) aged 52 to 91 years (mean [M] = 70; standard deviation [SD] = 12) who had experienced a stroke 6 or more months previously, gave written consent to participate. Ethical approval was granted by the local research ethics committee. Stroke participants with hemiparesis who were able to walk 10 m independently with or without a walking aid were recruited via a consortium of local GPs. Volunteers were excluded if they reported any other serious medical condition or complication that would preclude safe participation in the study, dementia, or cognitive problems (Mini-Mental State Examination < 21) that would prevent them from following instructions.

The participants were assessed at the onset of the study by a physiotherapist on the Fugl-Meyer Lower Extremity Scale[12] and Rivermead Motor Assessment Gross Function subscale.[13] A summary of the clinical profiles of the stroke participants is displayed in Table 1. The average entry post-cerebrovascular accident (CVA) for the study was 45 months (SD = 32), and there were equal numbers of cases with right- and left-hemisphere lesions. Average walking speed over a distance of 10 m was 0.81 m/s (SD = 0.39). For comparison, normal overground walking speeds of 0.80 to 1.61 m/s for 65- to 80-year-olds have previously been reported.[14]

Equipment

Walking was performed on a treadmill (Kettler Toronto; minimum speed 0.28 m/s). An overhead harness (Arjo Bianca) was used for safety; this harness provided no bodyweight support other than in the case of a fall. Custom-written software running on a PC provided clearly audible metronome pulses, which were set approximately to each individual's heel strike cadence when walking at a comfortable rate (M = 0.37 m/s, SD = 0.09). An Oxford Metrics Vicon tracking system with 6 infrared cameras captured the motion of ankle markers at 120 Hz.

Procedure

An initial familiarization period allowed each participant to adjust to walking on the treadmill and the experimenter to determine what constituted a comfortable cadence. To begin, the treadmill speed was gradually increased from zero until the participant stated that it was set at a comfortable pace. Fine adjustments to speed were then made according to the individual's preference in a 5-minute practice period, during which the experimenter matched the metronome beats to heel strike intervals. Participants were allowed to hold the hand rail in front of them if necessary. The task required participants to maintain the speed over a series of trials and to synchronize individual heel strikes with the metronome pulse.

Participants performed five 100-pulse trials, comprising 100 steps each. Trials consisted of 20 metronome pulses without phase shift as baseline, followed by 4 sections of 20 pulses, in each of which there was 1 phase shift occurring at an unpredictable time. Motion capture and metronome pulses were initiated for each trial once the participant was walking comfortably. Positive metronome phase shifts equal to 20% of the interpulse interval (IPI; ie, 36° of the gait cycle) were manually triggered at times that were randomized for each participant between the first and the fifth pulse of each section. These were selected pseudorandomly by the experimenter to yield 10 phase shifts on the paretic leg and 10 phase shifts on the nonparetic leg to assess interlimb phase correction response. The phase shift value of 20% was based on the 10% shifts used in previous work with healthy participants[6] but increased to allow for the greater task difficulty and increased variability expected in stroke. Given that stroke patients demonstrate delayed muscle latencies in response to perturbation,[15] we expected difficulty in accelerating steps, which is required by negative phase shift, and therefore only included positive phase shifts. We predicted that a beat that occurred 20% later than expected would encourage a lengthened step duration, without being so large as to promote the insertion of additional steps.

To determine the influence of the auditory pacing on temporal control, the first trial included an additional (1 minute) continuation period of walking without the metronome. During this period, participants were encouraged to maintain the frequency of heel strikes set up in the synchronization phase.

Data Analysis

Baseline analysis included measures from all 8 participants. A total of 40 phase shift trials (8 participants × 5 trials) were run. Operator error resulted in loss of 3 trials. The remaining 37 trials included 148 phase shifts (4 phase shifts per trial). One participant in the group (no. 6) was discarded from the analysis of phase correction because of failure to synchronize back to the pre–phase shift asynchrony, instead exhibiting a continuous drift of the phase away from baseline, leaving (148 − 20 = 128) phase shifts. After visually checking the data, a further 21 phase shifts (10 paretic and 11 nonparetic) were lost or discarded on the basis that no correction was observed. The remaining 107 correction responses were analyzed. Data for the continuation period of 1-minute walking without the metronome were successfully collected for each of the 8 participants.

Performance measures included average phase control, period control, limb asymmetry, and the compensation function. Phase control was defined in milliseconds (ms) as the asynchrony between each metronome pulse and the associated heel strike response (defined as the time of the most anterior position of the ankle marker in each cycle). Negative asynchrony signified that the response occurred in advance of the metronome pulse and positive that it occurred after the pulse. Period control was calculated as the proportional asynchrony error from the target pulse—(IPI [in ms] − Interresponse Interval [IRI; in ms])/IPI—for each participant.[6] Limb symmetry was measured as the difference between paretic to nonparetic and nonparetic to paretic heel strike intervals (ms).

Each response asynchrony following the phase shift was subtracted from the average of the 3 steps prior to phase shift asynchrony (T-3 to T-1, where T is the response at which phase shift occurs) to determine the relative asynchrony.[11] Compensation for the phase shift was determined by fitting the function $Y = r + q e^{-t/p}$; where Y is the estimated asynchrony, p is the time constant, q is the point at which the curve intercepts the vertical axis, and r is the final asymptote at which the compensation function settles.

Mean data were analyzed using repeated measures ANOVA with the 2 factors—namely, limb (paresis/nonparesis) and condition (baseline/continuation)—to determine significant effects ($P < .05$) of the metronome. Paired t tests with Bonferroni correction were used to assess significant intralimb differences ($P < .05$).

Results

The average walking speed on the treadmill (0.44 m/s) was reliably slower ($t(7) = 7.162$; $P < .001$) than the average recorded time for overground walking of 0.87 m/s. A section of treadmill walking for 1 participant is shown in

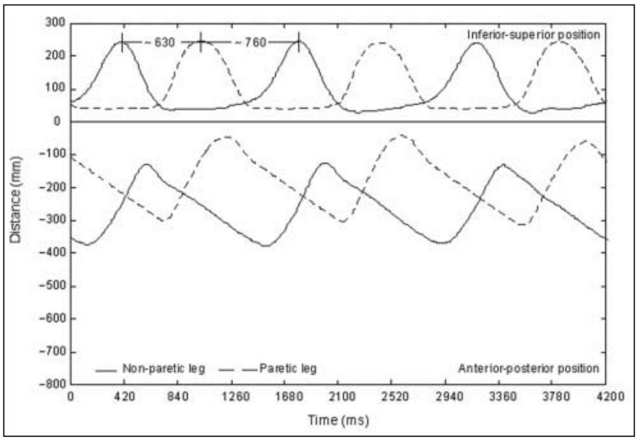

Figure 1. Illustrative data showing the step pattern of 1 participant: waveforms show heel motion for the nonparetic (left; solid line) and paretic (right; dashed line) sides. The figure illustrates the asymmetrical pattern following stroke

Table 2. Phase Timing Control (Synchronization Between Heel Strike and Metronome Rhythm) and Period Timing Control (Ability to Maintain Heel Strike Frequency Without the Metronome)[a]

	Paretic	Nonparetic
Mean baseline asynchrony (ms)	–47.4 (75.5)	–49.1 (65.0)
SD of baseline asynchrony (ms)	86.6 (54.0)	77.7 (38.2)
Percentage proportional error in period control: mean ([IPI – IRI]/[IPI]) × 100		
Baseline	5.3 (7.6)	5.3 (6.5)
Continuation	8.1 (9.3)	5.3 (3.5)

[a] The standard deviation (SD) across participants is given in parentheses.

Figure 1, and a degree of asymmetry between the timing of paretic and nonparetic heel strike is apparent. The average paretic–nonparetic interval was 744 ms (SD = 181), and the nonparetic to paretic interval was 722 ms (SD = 108); however, the difference in means was not statistically reliable. Nor did the difference between paretic and nonparetic intervals change significantly with (18 ms; SD = 143) and without (25 ms; SD = 140) the metronome.

The anterior–posterior displacement values in Figure 1 suggest that the interheel strike distance for the nonparetic swing is greater than that for the paretic swing. The mean overall distance was (M = 326.6 mm; SD = 103), but ANOVA revealed no main effect of paresis or metronome on step length for the group.

Measures of phase and period timing control are provided in Table 2. A negative onset asynchrony between heel strike and metronome beat (M = –48 ms; SD = 68) showed that overall, participants tended to step in advance of the metronome pulse during the baseline period. There was no significant difference in asynchrony between the paretic and nonparetic limbs. Nor was there any significant difference in variability between the paretic and nonparetic sides. Although mean proportional error on the paretic side appeared to be higher during the continuation phase, ANOVA showed no reliable interaction (Condition × Limb) for absolute proportional error in period timing control ([IPI − IRI]/[IPI × 100]).

A coarse-grained initial estimate of percentage correction for phase shift using group-averaged data (Figure 2) indicates progressive recovery of synchrony with successive steps after phase shift to the paretic and nonparetic sides. The first step response differs as a function of the side of perturbation. Immediate error correction by the nonparetic limb (23%) is apparent after phase shift is introduced at paretic heel strike, whereas initial correction by the paretic limb (6%) is small after phase shift on the nonparetic side.

Figure 3 illustrates, for a single trial, the compensation function (heel strike asynchrony over steps following a phase shift of the metronome), with phase shift occurring on the nonparetic side of 1 participant. The example demonstrates how correction for phase perturbation was spread over about 7 subsequent heel strikes, with overcorrection and variability around the predicted curve evident.

We found that 7 of the 8 participants showed correction for the phase shift, with phase timing returning to the pre–phase shift asynchrony values over subsequent heel strike responses. The mean correction parameter (p) was 5.2 ms/step (SD = 2.7), and the relative asymptote (r) was 50.2 ms (SD = 88.0). Average correction function parameters with the shift on either limb are summarized in Table 3. A significant difference between the side of limb was found ($t(6) = -3.00$; $P = .024$), indicating that participants needed longer to compensate when the phase shift occurred on the nonparetic side, requiring initial correction with the paretic limb. Although participants took longer to reach asymptote, the mean relative asymptote was the same ($t(6) = 0.564$; $P = .593$) whether the phase shift occurred on the paretic or nonparetic side.

Discussion

The aim of the present study was to evaluate timing control in hemiparetic, treadmill walking. This was analyzed both in terms of heel strike timing relative to the metronome during steady synchronization and using phase shifts to measure correction of the synchronization of heel strike with the auditory pulse. Participants produced small but equal step lengths, with no significant time difference between interlimb heel strikes. Although clinically the stroke group demonstrated lateralized motor control impairments on the Fugl-Meyer, which were sufficient to limit gross motor function, step length and heel strike appeared symmetrical during treadmill

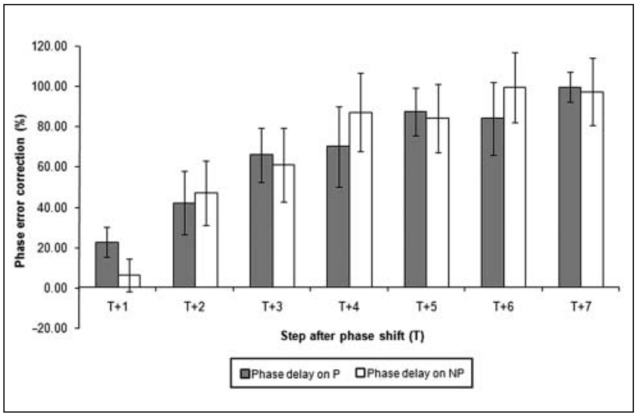

Figure 2. Bar chart illustrating the group mean (n = 7) percentage error correction for subsequent steps after phase shift (*T*) as a function of the side of perturbation: the shaded columns represent steps after the phase shift occurred on the paretic side, and the white columns show steps after phase shift on the nonparetic side. [= 100 - (((relative asynchrony at T - relative asynchrony at T+1, 2, 3…) / relative asynchrony at T) *100).] The error bars indicate the standard error for the sample

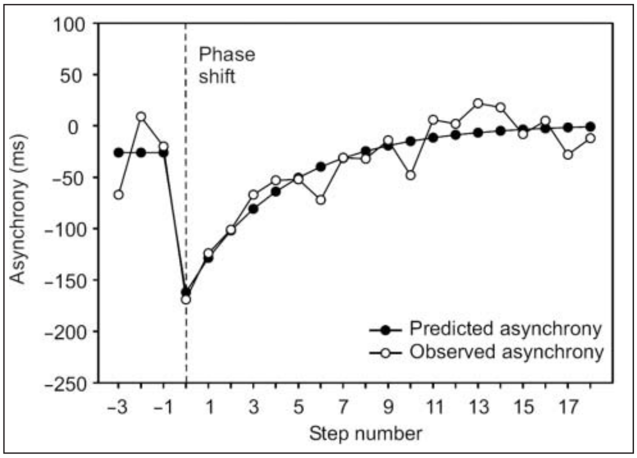

Figure 3. Illustration of observed and predicted correction for metronome phase shift: raw data from 1 participant trial with phase shift timed with nonparetic heel strike

walking. Similarly, a recent study showed minimal temporal and spatial asymmetries in a large proportion of high-level community-ambulating stroke participants while walking on a pressure-sensitive mat.[16]

Chronic stroke participants showed ability to synchronize heel strike responses with the metronome during treadmill walking. Overall, asynchrony was negative, indicating that the performer responded in advance of the metronome. This is a common observation in synchronization studies, referred to as "anticipation tendency," and its presence after stroke suggests that the predictive capabilities associated with the cerebellum were relatively intact in this patient group.[17,18]

Table 3. Correction Following Phase Shift[a]

Group Mean (n = 7) With Shift on Individual Limbs	Paretic	Nonparetic
Correction parameter p (ms/step)	4.0 (1.8)	6.5[b] (3.7)
Relative asymptote r (ms)	64.0 (68.0)	36.4 (138.9)

[a] The time constant p is defined as a function of $Y = r + q^{e-t/p}$, where Y is the estimated asynchrony, q is the point at which the curve intercepts the vertical axis, and r is the final asymptote at which the compensation function settles. The standard deviation across participants is given in parentheses.

[b] Indicates a statistically significant ($P < .05$) difference between paretic and nonparetic limbs.

Overall variability in synchronization was high relative to stepping on the spot by healthy adults.[6] This could partly be a result of the hemiparesis and partly a result of the larger IPI used for this group. Variability in synchronization may reflect impaired motor execution or phase correction.[19] We had anticipated that stroke participants might have impairments in both functions and for this reason assessed their ability to respond to perturbation of metronome phase. Despite obvious impairments, 7 of the 8 stroke participants showed correction for metronome phase shifts. Overall, the group demonstrated a progressive reduction in synchronization error following phase shifts, which is consistent with linear first-order correction of phase.[19] These results extend qualitative evidence[7] that stroke participants respond to a positive phase shift with a slowed step response. By comparing the correction parameter for phase shifts on either side, we identified a previously unreported, but significant, difference in the recovery of synchrony between the paretic and nonparetic sides. What is important is that the ability to correct for the phase shift was slower when it occurred on the nonparetic side. The slowed correction could have been a result of impaired motor execution or timing adjustment of the paretic limb following phase shift on the nonparetic side. Nonetheless, the ability to correct for the metronome perturbation demonstrates the capacity of auditory pacing to produce temporal changes in hemiparetic gait.

Period timing control exhibited appreciable variability, and contrary to expectations, neither the accuracy nor the variability was reliably altered by the presence of the metronome. Based on immediate improvements in poststroke gait symmetry with acoustic-paced treadmill walking,[5] it was expected that the metronome would facilitate timing control by providing feedback for error correction. Despite an overall reduction in mean proportional error of the paretic limb with acoustic pacing, we found no significant difference between the timing of paretic and nonparetic limb response intervals with or without the metronome. The finding is comparable with a previously published study,[7] which also failed to demonstrate immediate improvements in spatiotemporal symmetry with the metronome. However, the absence of positive effects on gait symmetry could be explained by the high level of gait symmetry observed during treadmill walking, which provided little room for improvements with acoustic pacing.

Recent work[16] emphasizes the importance of evaluating gait asymmetry in hemiparetic stroke. Although asymmetry is not detected in all community-ambulating stroke participants during steady-state walking, results suggest that patients with mild impairments may have increased difficulty responding to perturbations with the paretic side. We found a lateralized problem of recovery after phase shift, with paretic limb correction latencies on the step immediately after perturbation. This finding may be important for obstacle avoidance, walking on uneven surfaces, and negotiating rapid changes in direction. Previous research[20] also reported impaired ability to modify the stepping pattern following unilateral cortical stroke. In contrast to the present study, however, no interlimb differences were found in response to a physical obstacle, and it was proposed that the impairment was caused by a generalized problem with gait coordination. Although lateralized postural control impairments have been reported in chronic stroke participants in response to platform perturbations,[21] further investigation is required to determine the underlying cause for the slowed recovery when the nonparetic step is perturbed.

Future studies with this participant group might include changing the speed of the treadmill and the intensity of training. The same paradigm could also incorporate kinetic measures to determine the moments produced by lower-limb joints to execute the motor correction. Further research measuring ground reaction force changes in response to phase correction could provide new insight into gait timing control in hemiparesis, which may facilitate optimal gait rehabilitation and adaptation to environmental changes. By applying this paradigm to different patient groups, more specific rehabilitation programs for gait may be developed.

In summary, gait compensation for metronome phase shift is slowed on the hemiparetic side, so that restoration of phase is slower when the shift occurs on the nonparetic side. Nonetheless, phase correction was observed in 7 of 8 participants, and this ability may be considered an important factor in the benefit of metronome cueing of gait in rehabilitation.

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Declaration of Conflicting Interests

The authors declared no conflicts of interest with respect to the authorship and/or publication of this article.

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