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Speed Impacts Frontal-Plane Maneuver Stability of Individuals with Incomplete Spinal Cord Injury

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Abstract

Background: Following incomplete spinal cord injury, people often move slowly in an effort to maintain stability during walking maneuvers. Here we examine how maneuver speed impacts frontal-plane stability in people with incomplete spinal cord injury. We hypothesized that the challenge to control frontal-plane stability would increase with maneuver speed; specifically, the minimum lateral margin of stability would be smaller and the required coefficient of friction to avoid a slip would be greater during fast vs. preferred speed maneuvers.

Methods: We measured kinematics and ground reaction forces as 12 individuals with incomplete spinal cord injury performed side-step, lateral maneuvers at preferred and fast speeds. We examined four sequential steps: the Setup and Pushoff steps initiated the maneuver, and the Landing and Recovery steps arrested the maneuver.

Findings: Our hypotheses were partially supported. Maneuver time was shorter during fast vs. preferred speed maneuvers ($p=0.003$). Minimum lateral margin of stability was smaller during the Setup step of fast vs. preferred speed maneuvers ($p=0.026$). We found no differences in minimum lateral margin of stability between speeds for the Landing and Recovery steps ($p>0.05$). The required coefficient of friction was not different between fast and preferred speed maneuvers ($p=0.087$).

Interpretation: The greatest effect of increasing maneuver speed occurred during the Setup step; as speed increased, participants reduced their minimum lateral margin of stability ipsilateral to the maneuver direction. This action allowed maneuvers to be performed more quickly without requiring a greater lateral impulse during the Pushoff step. However, this strategy reduced passive stability.

Keywords

Gait; Stability; Spinal Cord Injury; Maneuver; Balance; Locomotion

1. Introduction

Following incomplete spinal cord injury (iSCI), dynamic balance, the ability to maintain equilibrium during movement, is often impaired. Up to 75% of ambulatory individuals with iSCI fall annually [1], with most falls occurring during walking [2]. In addition to fall risk, impaired balance adversely affects mobility [3], contributing to limited walking speeds [4], poor energetic efficiency [5], and avoidance of difficult walking situations [6]. Improving our understanding of the biomechanical mechanisms people with iSCI use to maintain dynamic balance will assist identification of fall risk situations and inform population-specific mobility interventions.

Walking maneuvers – changes in direction or speed [7] – are an essential component of community ambulation [6, 8] that can challenge dynamic balance for people with iSCI. Compared to straight walking, turning maneuvers are less locally stable [9] and require a complex generation of mediolateral impulses and compensatory actions [10]. Although we have a poor understanding of how individuals with iSCI control maneuvers, strategies observed in older adults provide insight. Older adults walk slowly [11], and take slower, shorter, and more steps to change direction than younger adults [12, 13]. Moving slowly may be advantageous for maintaining the center-of-mass (CoM) within a dynamic base of support (BoS) because slow maneuvers reduce linear and angular momentum [11]. In addition, slow maneuvers reduce the required coefficient of friction (RCoF) [14] necessary to avoid slipping.

However, in situations when walking speed is crucial (e.g. avoiding oncoming pedestrians), maneuvering slowly may not be a viable control strategy for maintaining balance. Indeed, foot placement errors increase during rapid maneuvers among older adults [12], and younger adults make less accurate maneuvers at preferred vs. slower speeds [15]. Our purpose is to examine the interaction between maneuver speed and frontal-plane stability in people with iSCI. We investigate frontal-plane stability because research suggests that during gait, the control of mediolateral stability requires greater active control than fore-aft stability [16–18].

Specifically, we observed a side-step “lane-change” walking maneuver. This maneuver involved sequential steps we termed “Setup,” “Pushoff,” “Landing,” and “Recovery” (Figure 1) [19]. Increasing maneuver speed creates frontal-plane stability challenges unique to each step. The Setup and Pushoff steps initiate the maneuver. During the Setup step, people lean into the maneuver to reduce CoM accelerations contralateral to the maneuver direction [20]. This is followed by the Pushoff step, when a lateral impulse is generated to redirect CoM motion [19]. Further increase of the anticipatory lean during the Setup step may help individuals with iSCI initiate faster maneuvers by reducing the magnitude of the lateral impulse that must be generated during the Pushoff step [19]. To gain insight into the body’s passive resistance to lateral maneuvers, we can calculate a lateral margin of stability (MoS) as the distance between the extrapolated CoM (XCOM) – a measure accounting for both CoM position and velocity — and the base of support (BOS) [21]. During walking, the impulse required to move the XCOM beyond the BOS will be proportional to the magnitude of the MoS [21]. Thus, we hypothesize that as maneuver speed increases, individuals with iSCI will decrease the minimum lateral MoS during the Setup step in order to reduce the

body's resistance to the impending maneuver, and increase the laterally-directed ground reaction force (GRF) during the Pushoff step in order to create a greater change in lateral momentum. The consequences of these actions, decreasing MoS and increasing shear GRF, could challenge gait stability by decreasing resistance to external perturbations during the Setup step and increasing the required coefficient of friction (RCoF) necessary to avoid a slip, respectively. Increases in RCoF have been observed previously as turning speed increases [22].

At the completion of the maneuver, lateral velocity must be arrested and forward walking safely reestablished [23]. During the Landing step of fast maneuvers, individuals with iSCI may have difficulty generating a sufficient lateral impulse to maintain a positive MoS. Attempts to generate a sufficient lateral impulse may also result in an increased RCoF to avoid a slip. Finally, during the Landing step, the nervous system acts to control the impending mediolateral foot placement of the trailing swing limb [24, 25]. Mediolateral foot placement of the swing limb will determine the BOS during the Recovery step. As maneuver speed increases, the lateral momentum of the swing limb will increase, an inability to accurately perceive or generate an appropriate muscular response to control this increased momentum could result in a narrow BOS during the Recovery step. We hypothesize that as maneuver speed increases, the challenges to frontal-plane stability associated with arresting the maneuver will increase, i.e. there will be a greater RCoF during the Landing step and reduced minimum lateral MOS during both the Landing and Recovery steps. These changes will increase the potential to slip during the Landing step and reduce the body's resistance to perturbations during the Landing and Recovery steps.

2. Methods

2.1 Participants

12 people provided written informed consent and participated in the study. Institutional Review Boards from both Northwestern University and the Edward Hines Jr. Veterans Administration Hospital approved the protocol. All participants were between 18 and 75 years of age. Inclusion criteria included: spinal cord injury level between C1-T10, American Spinal Injury Association Impairment Scale (AIS) C or D, > 1 year post-injury, range of motion within functional limits of ambulation, and ability to walk 10 meters without assistive devices. Participants were excluded for the following reasons: excessive lower limb spasticity of the quadriceps or hamstring muscle groups as measured by a score of > 3 on the Modified Ashworth Scale, inability to tolerate 10 minutes of standing, severe cardiovascular or pulmonary disease, recurrent fracture history, known lower extremity orthopedic problems, concomitant central or peripheral neurologic injury, or inability to provide informed consent due to cognitive impairments. Participants did not alter medications for the study.

2.2 Experimental setup

We measured ground reaction forces (GRF), and whole body kinematics as participants walked overground (Figure 1). Walking occurred within two parallel paths (7.5 x 0.51 meters) marked on a smooth, level floor. We measured GRF data at 1000 Hz from four force

plates (AMTI, Watertown, MA) mounted flush with the floor, arranged in a rectangular configuration (0.94 x 1.02 meters), and centered within the two parallel paths (Figure 1). We used a 12-camera motion capture system (Qualisys, Gothenburg Sweden), operating at 100 Hz, to record 3D coordinates of 67 reflective markers. We placed markers on the cervical vertebra 7 and bilaterally on: the 1st, 2nd, and 5th metatarsals; calcaneus, medial and lateral malleoli; medial and lateral epicondyles of the knee; greater trochanter; anterior superior iliac spine; posterior superior iliac spine; iliac crest; acromion process; medial and lateral epicondyles of the humerus; and the radial and ulnar styloid processes. In addition, to track segment motions, we affixed rigid 4-marker clusters bilaterally to the upper arm, forearm, thigh, and shank segments.

2.3 Protocol

First, to characterize participants' functional abilities, we collected demographic and clinical outcome measures. Clinical measures included the lower extremity motor score (LEMS) portion of the International Standards for Neurological Classification of Spinal Cord Injury (ISNCSCI) examination, manual muscle tests of the hip abductors, the Modified Ashworth Scale to test bilateral quadriceps and hamstring spasticity, the Berg Balance Scale (BBS), the 10 Meter Walk Test (TMWT) performed at preferred and maximum speed, the Timed Up and Go (TUG), and the Walking Index for Spinal Cord Injury II (WISCI).

Next, participants performed overground walking trials consisting of straight-ahead walking and side-step "lane change" maneuvers. We examined this particular maneuver because it is common during community ambulation (e.g. obstacle avoidance), is quantifiable using established methods, and builds on our previous work [19, 26]. Participants walked at both preferred and fast speeds. During fast trials, we instructed participants to walk and maneuver as quickly as possible while maintaining safety.

During straight-walking trials, participants' feet remained within a single path. During maneuver trials, participants began walking in 1 of 2 parallel paths selected specifically to create a maneuver to the left or right. We instructed participants to initiate the maneuver to the parallel walking path when the limb contralateral to the maneuver direction first contacted the force plates, and defined this step as the Pushoff step. This movement resulted in a "side-step" maneuver. Participants continued walking forward after maneuvering to the new path.

There were six total overground walking conditions. The following conditions were performed at comfortable preferred and fast walking speeds: straight-ahead walking, maneuvers to the left, and maneuvers to the right. Condition order was block-randomized (all trials of a condition were completed before progressing to the next condition). Participants performed 2-5 practice trials for each condition. We adjusted the starting distance during the practice trials to minimize any observable step length adjustments required for the Pushoff step to land entirely within the force plates. After practice, we recorded five successful overground walking trials for each condition. Participants rested two minutes between conditions and as needed between trials. For safety, participants wore a gait belt and a physical therapist guarded the participant using stand-by assist (non-contact).

2.4 Data analysis

Kinematic marker data was processed using Visual3D (C-Motion, Germantown, MD) and custom MATLAB (Mathworks, Natick, MA) scripts. Marker data were gap-filled and low-pass filtered (Butterworth, 6 Hz cut-off frequency). GRFs were low-pass filtered (Butterworth, cut-off frequency 20 Hz). Time of initial foot contact (IC) and toe-off (TO) were identified using fore-aft positions of the calcaneus and 5th metatarsal markers. All gait events were visually inspected and, when available, checked against GRF data to ensure accuracy. During each straight-walking trial, we analyzed the two consecutive steps occurring as the participant crossed the force plates. During each maneuver trial, we analyzed four sequential steps: Setup, Pushoff, Landing, and Recovery (Figure 1). The Pushoff and Landing steps occurred as the participant crossed the force plates. The Setup and Landing steps began at IC of the limb ipsilateral to the maneuver direction, and the Pushoff and Recovery steps began at IC of the limb contralateral to the maneuver direction.

To quantify maneuver performance, we calculated maneuver time and the total body CoM path length during the period starting at IC of the Pushoff step and ending at IC of the Recovery step. We calculated the CoM path length as the path integral of the total body CoM position during this period. Total body CoM position was estimated using segmental analysis, with processed marker data to calculate joint centers and segment CoM positions. We then calculated a total body CoM position using a weighted average based on the segment mass fractions and segment CoM positions.

To quantify changes in frontal-plane stability, we calculated step width, minimum lateral MoS, and RCoF. For each participant, we used ten total steps at each speed to calculate average values for these three stability metrics during the Straight, Setup, Pushoff, Landing, and Recovery steps. We pooled data from the left and right limbs for these calculations.

Step width was calculated at IC for each step as the medio-lateral distance between the 5th metatarsal markers on each foot.

To quantify passive lateral stability, we calculated participants' lateral XCoM position [21]:

$$XCoM = CoM + CoMV * \sqrt{l/g}$$

CoM = lateral center of mass position

CoMV = lateral center of mass velocity

l = leg length

g = gravitational constant

We calculated continuous CoMV as the derivative of CoM position. We calculated continuous leg length as the distance between the greater trochanter and lateral malleolus marker. Next, we used XCoM to identify the minimum lateral MoS during stance phase of each step. We calculated lateral MoS as:

$$MoS = BoS_{lat} - XCoM$$

Lateral base of support (BoS_{lat}) was approximated from the lateral position of the stance limb's 5th metatarsal. MoS was positive when the XCoM was medial to the BoS_{lat}.

We calculated RCoF (μ) as the ratio of shear to normal GRFs using the following equation [22]:

$$RCoF = \frac{F_{shear}}{F_{normal}} = \frac{\sqrt{F_x^2 + F_y^2}}{F_z}$$

F_x = lateral ground reaction force

F_y = anterior – posterior ground reaction force

F_z = vertical ground reaction force

The configuration of the force plates only allowed us to capture GRFs and calculate RCoF for the Straight, Pushoff, and Landing steps. Each step was time-normalized to stance duration (IC to ipsilateral TO). To avoid calculating RCoF values in regions when both shear and normal forces approached zero, we analyzed only the region between 10 and 90% of stance phase [27]. We then identified the peak RCoF for each step to estimate the greatest challenge to slipping. The Pushoff and Landing steps for two participants regularly overlapped spatiotemporally on a single force plate. We were not able to analyze individual limb RCoF data for these two participants.

2.5 Statistical Analysis

To compare the effect of maneuver speed on performance metrics of maneuver time and CoM path length, we used Wilcoxon Sign-Rank tests (since t-test assumptions were violated).

To compare the effects of step (Straight and each of the four maneuver steps) and walking speed (preferred and fast) on step width and minimum lateral MoS, we used repeated measures ANOVAs. We also used repeated measures ANOVAs to compare the effects of step (Straight, Pushoff, and Landing) and walking speed (preferred and fast) on RCoF. If sphericity was violated, we used the Greenhouse-Geisser (G-G) F-statistic and p-value to test for main and interaction effects. When we found a significant main effect of step, we performed Bonferroni-corrected pairwise comparisons to identify differences between steps. We conducted a simple effects analysis when we found a significant interaction of step and speed (i.e. a t-test compared speed at each step).

We set significance at the $p < 0.05$ level for the Wilcoxon Sign-Rank tests, repeated measures ANOVAs, pairwise comparisons, and simple effects analysis.

3. Results

3.1 Participants

Twelve ambulatory individuals with chronic motor iSCI (all AIS D) participated. Due to fatigue, one participant did not perform fast maneuver trials; this participant's data was

excluded from the group analysis. The eleven participants (9 male/2 female) completing the full protocol were a mean age of 53 years (SD 10), mean height of 1.73 m (SD 0.10), and mean body mass of 76.6 kg (SD 16). One participant wore a rigid ankle-foot orthosis on their right limb. No other participants used assistive devices. See Table 1 for complete participant demographics and clinical outcome measures.

3.2 Maneuver Performance

During straight-walking the XCoM path oscillated mediolaterally, remaining inside the BoS. During the maneuver trials, the lateral motion of the XCoM path occurred predominantly during the Pushoff and Landing steps and often moved outside the BoS during the Setup step of fast trials (Figure 2). Lateral motion was faster and the CoM path length was shorter during fast maneuvers vs. preferred speed maneuvers. Maneuver time was also significantly less during fast vs. preferred speed maneuvers (Wilcoxon's $z = 2.934$, $p = 0.003$) (Figure 3a). CoM path length was significantly shorter during fast vs. preferred speed maneuvers (Wilcoxon's $z = 2.045$, $p = 0.041$) (Figure 3b).

3.3 Step Width

The repeated measures ANOVA found a significant main effect of step ($p < 0.0005$), no significant main effect of speed ($p = 0.863$), and no significant interaction between step and speed ($p = 0.776$) on Step Width (Figure 4a). The Pushoff and Recovery steps were both narrower than straight-walking steps (both Bonferroni-corrected pairwise $p < 0.0005$). The Landing step was wider than straight-walking steps (Bonferroni-corrected pairwise $p < 0.0005$).

3.4 Minimum Lateral Margin of Stability

The repeated measures ANOVA found a significant main effect of step ($p < 0.0005$) and significant interaction between step and speed ($p = 0.004$) on minimum lateral MoS (Figure 4b). The Setup step had a significantly smaller minimum lateral MoS than every other step (all Bonferroni-corrected pairwise $p < 0.002$). The minimum lateral MoS of the Setup step was significantly smaller during the fast vs. preferred speed maneuvers ($p = 0.026$). The minimum lateral MoS of the Pushoff step was greater during the fast vs. preferred speed maneuvers ($p = 0.007$).

3.5 Required Coefficient of Friction

There was a trend for greater RCoF values during fast vs. preferred speed (Figure 5). However, none of the repeated measures ANOVA main or interaction effects were significant (step: $p = 0.295$, speed: $p = 0.087$, step x speed: $p = 0.668$) for RCoF.

4. Discussion

When asked to perform lateral “lane-change” maneuvers rapidly, individuals with iSCI moved significantly faster than during preferred speed maneuvers, decreasing their maneuver time by ~20%. Our hypothesis that increasing maneuver speed would create step-specific challenges to lateral stability, consisting of either a decrease in minimum lateral MoS or increase in RCoF, were partially supported. Compared to straight-ahead walking, we

observed significant decreases in the minimum lateral MoS during the Setup, Landing, and Recovery steps of lateral maneuvers. Of these steps, increasing maneuver speed only resulted in further reductions in minimum lateral MoS during the Setup step. There were no significant differences in RCoF between preferred and fast maneuvers.

In support of our hypothesis, we observed that individuals with iSCI made greater reductions in their minimum lateral MoS during the Setup step during fast vs. preferred speed maneuvers. Our observation is consistent with past research that found that during walking maneuvers, non-impaired people make anticipatory adjustments during the Setup step by leaning [28] and reducing minimum lateral MoS ipsilateral to the maneuver direction [19, 26]. Decreasing MoS ipsilateral to the maneuver direction should assist in facilitating maneuver initiation by reducing the impulse required to redirect the CoM trajectory laterally [19]. For individuals with iSCI, this anticipatory adjustment, which decreases the body's resistance to the impending change of direction, may be critical for increasing maneuver speed if motor deficits limit their ability to generate a larger lateral impulse. During the Setup step of many fast maneuvers trials, minimum lateral MoS was negative, indicating that the XCoM was outside the base of support. A negative MoS takes advantage of gravity to assist with the intended lateral motion. However, a negative MoS during fast maneuvers may increase vulnerability to external perturbations by placing a person in a less passively stable position [21], and increases the importance of placement of the following step to recapture the CoM. It is possible that individuals use other mechanisms to offset this decrease in passive stability such as increasing cognitive focus [29], which could allow a faster or more appropriate reactive response should an unintended perturbation occur. Interestingly, a similar strategy has been observed during finger dexterity tasks [30, 31] where people also make anticipatory reductions in finger stability in preparation for rapid movements.

We hypothesized that increasing maneuver speed during the Pushoff step would challenge frontal-plane stability due to a greater RCoF necessary to avoid a slip. Although there was a trend for a greater RCoF during fast maneuvers, we did not find a significant difference between speeds. This finding contrasts previous research that found a significantly larger RCoF during fast vs. normal speed walking turns [22] in non-impaired populations. If a true peak RCoF occurred during the first or last 10% of stance phase, our analysis would not detect these values. However, the peak RCoF values we report are representative of the phase in the gait cycle when slips are most likely to occur. Peak RCoF occurred on average between 65-75% of stance phase for the Pushoff and Landing steps. This timing coincided with peak shear GRF values, which are thought to be predictive of slip potential during gait activities [32]. Nonetheless, across all participants the average peak RCoF during the Pushoff step of fast maneuvers was $\mu=0.28$, and for any single participant the largest average peak RCoF we observed was $\mu=0.39$. These values are well below the Occupational Safety and Health Administration recommended static coefficient of friction of $\mu=0.50$ for walking surfaces [33], suggesting that under normal walking conditions the risk of slipping during the Pushoff step is small in the iSCI population. In contrast, non-impaired individuals were found to have peak RCoF during normal and fast speed turns of $\mu=0.45$ and $\mu=0.54$ respectively [22]. The smaller RCoF we observed were likely a result of the relatively slow walking speeds used by participants with iSCI. It is unclear if the walking speeds selected during fast maneuvers were limited because participants were physically unable to move

faster or because participants limited walking speed to prioritize for safety given that previous research suggests that individuals with iSCI will restrict mobility activities to avoid difficult tasks that challenge balance [6].

To terminate lateral momentum and reestablish forward walking following the maneuver, we hypothesized that increasing maneuver speed would decrease minimum lateral MoS during the Landing and Recovery steps and increase RCoF during the Landing step. These hypotheses were not supported. The minimum lateral MoS was significantly less during the Landing step when compared to straight-walking, but there were no significant differences in MoS between maneuver speeds. During the Landing step, participants' step width was 90% wider than during straight-walking steps. The large BoS that resulted from taking a very wide step may have aided in maintaining a consistent minimum lateral MoS between maneuver speeds by creating a large gravitational moment about the ankle [34, 35], which would act to slow the lateral velocity of the CoM. Participants may have also modulated their hip abduction moment during the Landing step as needed to offset differences in lateral velocity; this strategy has been observed among older adults who generate a substantially larger peak hip abduction moment vs. younger adults during the Landing step of lateral maneuvers [23].

Our hypothesis that the RCoF during the Landing step would be greater during fast vs. preferred speed maneuvers was not supported. Similar to our observations during the Pushoff step, there was a trend for a greater RCoF during fast vs. preferred speed maneuvers. Again, it is possible that the difference in speed between fast and preferred speed maneuvers was not large enough to result in a significant difference in RCoF. We believe it is likely that individuals with iSCI may have selected a strategy to distribute the negative work performed on the CoM over multiple steps (as opposed to arresting lateral velocity completely within a single step). Such a strategy would be beneficial if the individual has motor impairments that impacts their ability to generate a sufficient hip abduction moment during the Landing step. A limitation of this study was that we were only able to analyze the RCoF during the Pushoff and Landing steps.

Research on non-impaired individuals has found that, in response to perturbations, frontal-plane hip muscles modulate activity during swing phase to control mediolateral foot placement [24, 25]. During the Landing step of the current experiment the nervous system should act similarly to control the impending mediolateral foot placement of the trailing swing limb to determine the BoS during the Recovery step. As maneuver speed increases, the lateral momentum of the swing limb will increase and will require greater negative work to be performed about the hip in the frontal-plane to maintain a minimum lateral MoS consistent with preferred speed maneuvers. As such, we hypothesized that as maneuver speed increased, individuals with iSCI would have a smaller minimum lateral MoS during the recovery step of fast versus preferred speed maneuvers. We found that the minimum lateral MoS was significantly less during the Recovery step when compared to straight-walking. However, minimum lateral MoS during the Recovery step did not change with maneuver speed, suggesting that individuals with iSCI were able to generate a sufficient frontal-plane hip abduction moment to control mediolateral foot placement during the preceding swing phase.

Old age [36, 37] and body mass associated with obesity [38] are both associated with reductions in stability. To offset this instability, older adults [11–13] and obese individuals [38] select conservative gait patterns. In the current study, our participants spanned a large range of ages (18–75 years) and body weights (49–96 kg). This variability may have influenced the individual gait patterns selected to maintain balance during maneuvers. We performed a supplemental analysis to examine if there was a relationship between passive stability during the maneuvers and age or body weight. We calculated a Pearson's correlation coefficient relating minimum lateral MOS during the Setup step (the step demonstrating the greatest change in MOS with gait speed) at fast and preferred walking speeds to age and body weight. None of the correlations were significant ($p > 0.05$), suggesting that in individuals with iSCI these additional risk factors did not result in consistent changes in passive stability.

Overall, our results found that the greatest effect of increasing the speed of lateral side-step maneuvers on frontal-plane stability in individuals with iSCI occurred during the Setup step. Interestingly, our analysis suggests that individuals were in a less biomechanically stable position, and potentially at greater risk for falls, during the steps initiating the maneuver than during the steps used to arrest the maneuver, and that speed exacerbated this effect. As maneuver speed increased, in anticipation of performing a lateral maneuver, individuals reduced their minimum lateral MoS ipsilateral to the maneuver direction. This reduction in lateral MoS may have allowed individuals to increase the rate at which they could perform the maneuver without having to generate a greater lateral impulse during the Pushoff step. However, this strategy also reduces passive stability in the maneuver direction, increasing susceptibility to external perturbations. During the steps arresting the maneuver, although lateral MoS was less than during straight walking, we did not observe additional reductions in MoS as maneuver speed increased. As maneuver speed increased, we believe individuals distributed the additional negative work required to terminate lateral motion over multiple steps. Such a strategy would limit changes in MoS or RCoF that might create additional frontal plane stability challenges during any single step but could create issues in situations requiring an abrupt termination of lateral motion. These results suggest that individuals with iSCI compromise stability for speed when asked to perform rapid maneuvers. This trade-off might translate to an interaction between community ambulation ability, walking speed, and fall risk.

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Figure 1. Experimental Setup:

Participants performed overground walking over four force plates. During maneuver trials, participants began walking in an initial path selected to create maneuvers to the left or right (shown). During each straight-walking trial, we analyzed the two consecutive steps occurring as the participant crossed the force plates. During each maneuver trial, we analyzed four sequential steps: 1) Setup, 2) Pushoff, 3) Landing, and 4) Recovery. Only the Pushoff and Landing steps occurred as the participant crossed the force plates.

Figure 2. Foot Placement and Extrapolated Center of Mass (XCOM) data from an Example Representative Subject:

Data shown are from a) Straight-walking at comfortable preferred speed, b) a maneuver to the right at preferred speed, and c) a maneuver to the right at fast speed. The effect of maneuver speed was greatest during the Setup step. The minimum lateral margin of stability was smaller during fast vs. preferred speed maneuvers.

Figure 3. Maneuver Performance:

a) We measured maneuver time starting at initial contact (IC) of the Pushoff step and ending at IC of the Recovery step. Maneuver Time was significantly shorter during fast vs. preferred speed maneuvers. b) We calculated the maneuver path length as the path integral of the CoM during the period used to calculate maneuver time. CoM path length was significantly shorter during fast vs. preferred speed maneuvers.

Figure 4. Step Width and minimum lateral Margin of Stability:

a) The Pushoff and Recovery steps were both narrower than straight-walking steps, and the Landing step was wider than straight-walking steps. b) The minimum lateral Margin of Stability (MoS) was identified during stance phase of each step as the distance between the extrapolated center of mass (XCoM) and the lateral base of support. MOS was positive when the XCoM was medial of the BOS. The Setup step had a significantly smaller minimum lateral MoS than every other step. The minimum lateral MoS of the Setup Step was

significantly smaller during the fast vs. preferred speed maneuvers. The minimum lateral MoS of the Pushoff step was greater during the fast vs. preferred speed maneuvers.

Figure 5. Peak Required Coefficient of Friction:

We calculated Peak Required Coefficient of Friction (RCoF) as the ratio of shear to normal ground reaction force during straight-walking and during the Pushoff and Landing steps of the maneuver trials. There was a trend towards higher values of RCoF for the fast speed maneuvers, but none of the differences were statistically significant.

Table 1.

Participant Demographic and Clinical Data

Age (yrs)	Gender	Height (m)	Weight (kg)	SCI level	LEMS	MMT-Hip		MAS (Quad/Ham)		Berg	TMWT (s)		TUG (s)	WISCII
						Left	Right	Left	Right		Pref.	Max.		
53	M	1.83	79.5	C5-7	48	4	3	1/1+	2/2+	47	6.6	4.64	16.94	20
43	M	1.83	90.9	L 3-4	48	5	4	0/0	0/0	51	6.3	4.35	17.84	20
51	M	1.85	95.5	C5-6	35	5	5	0/0	0/0	45	7	5.36	16.47	20
74	M	1.58	60.5	C4-6	47	4	4	0/0	0/0	55	5.89	4.02	9.56	20
54	M	1.68	93.2	C6-7	42	3	3	1/2	1/2	39	12.13	8.66	21.34	20
56	F	1.65	53.2	T8-9	42	2	4	0/1	0/0	53	6.85	5.74	11.88	20
67	M	1.85	93.2	C4-6	35	2	1	0/1	0/0	36	47.96	36.56	62.15	19
49	M	1.85	84.5	C7	48	4	5	0/0	0/0	56	5.2	3.97	10.19	20
61	M	1.78	75.0	C5-7	41	3	5	0/1	0/0	49	7.29	5.06	11.16	20
32	M	1.73	77.3	C-7	45	4	4	0/0	0/0	56	4.27	3.61	8.09	20
50	M	1.65	84.1	C 3-6	45	3+	5	0/0	0/0	55	4.73	3.59	8.87	20
55	F	1.65	48.6	C 4-6	43	3	4	1/1+	0/0	47	8.01	6.17	14.1	20

LEMS: Lower extremity motor score portion of the American Spinal Injury Association Impairment Scale.

MMT-Hip: manual muscle test scores of the hip abductors

MAS: Modified Ashworth Scale to test quadriceps and hamstring spasticity

TMWT: 10 Meter Walk Test (10 MWT) time in seconds performed at preferred and maximum speed

BBS: Berg Balance Scale

TUG: Timed Up and Go time in seconds

WISCII: Walking Index for Spinal Cord Injury II