Spinal Cord (2021) 59:159–166
https://doi.org/10.1038/s41393-020-0516-5

ISCoS The International Spinal Cord Society

ARTICLE

The effects of light touch on gait and dynamic balance during normal and tandem walking in individuals with an incomplete spinal cord injury

Alison R. Oates [1] · Tarun Arora [2] · Joel L. Lanovaz [1] · Kristin E. Musselman [3,4]

Received: 17 September 2019 / Revised: 29 June 2020 / Accepted: 30 June 2020 / Published online: 9 July 2020

Abstract

Study design Prospective cross-sectional study

Objectives To investigate the effect of adding haptic input during walking in individuals with incomplete spinal cord injury (iSCI).

Setting Research laboratory.

Methods Participants with iSCI and age- and sex-matched able-bodied (AB) individuals walked normally (SCI $n = 18$, AB $n = 17$) and in tandem (SCI $n = 12$, AB $n = 17$). Haptic input was added through light touch on a railing. Step parameters, and mediolateral and anterior–posterior margins of stability (means and standard deviations) were calculated. Surface electromyography data were collected bilaterally from the tibialis anterior (TA), soleus (SOL), and gluteus medius (GMED) and integrated over a stride. Repeated measures ANOVAs examined within- and between-group differences ($\alpha = 0.05$). Cutaneous and proprioceptive sensation of individuals with iSCI were correlated to changes in outcome measures that were affected by haptic input.

Results When walking normally, adding haptic input decreased stride velocity, step width, stride length, MOS_{ML}, MOS_{ML}_SD, MOS_{AP}, and MOS_{AP}_SD, and increased GMED activity on the limb opposite the railing. During tandem walking, haptic input had no effect; however, individuals with iSCI had a larger step width SD and MOS_{ML}_SD compared with the AB group. Sensory abilities of individuals with iSCI were not correlated to any of the outcome measures that significantly changed with added haptic input.

Conclusions Added haptic input improved balance control during normal but not in tandem walking. Sensory abilities did not impact the use of added haptic input during walking.

Introduction

An incomplete spinal cord injury (iSCI) can have a significant impact on an individual's ability for independent locomotion. While many individuals with iSCI relearn to walk [1], there is a high rate of falls in this population that can lead to further injuries and hospitalization, and lowers their confidence and community participation [2–4]. Improving balance control can decrease fall risk [5, 6] and so is important for fall-prevention efforts. Balance control requires the integration of sensory input and motor output. Sensorimotor impairments following an iSCI can compromise the ability to maintain good balance control during walking.

In individuals with sensory impairments, augmenting sensory input using external sources helps in improving their balance control [7, 8]. One such way of augmenting

Supplementary information The online version of this article (https://doi.org/10.1038/s41393-020-0516-5) contains supplementary material, which is available to authorized users.

✉ Alison R. Oates
alison.oates@usask.ca

1 College of Kinesiology, University of Saskatchewan, Saskatoon, SK, Canada

2 Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic Foundation, Cleveland, OH, USA

3 Lyndhurst Centre, Toronto Rehabilitation Institute, University of Toronto, Toronto, ON, Canada

4 Department of Physical Therapy, University of Toronto, Toronto, ON, Canada

sensory input is providing haptic information in the form of lightly touching an external stable surface [9–11]. The stable surface is used to provide spatial information about the body position through changes in arm proprioception and cutaneous feedback [9]. Added haptic input has been shown to improve balance control during standing and/or walking in young [11] and older able-bodied (AB) individuals [12], as well as individuals with neurological impairments such as stroke [8] and Parkinson's disease [7, 13]. During standing, improvement in balance control has been shown through a moderation in tone of the postural muscles and/or reduction in postural sway characteristics such as sway amplitude, velocity, and variability [12, 13]. During walking, improvement in balance control has been shown by an increase in activation of weight-bearing muscles, or a decrease in variability of margin of stability (MOS), step width, or center of mass in the mediolateral (ML) and/or anterior–posterior (AP) directions [8, 11].

Among individuals with iSCI, added haptic input has been shown to improve standing balance by reducing postural sway similar to AB individuals with a larger effect in individuals with iSCI who have more upper extremity (UE) cutaneous sensation and less lower extremity (LE) proprioception [14]. Improving balance control with the addition of light touch may be a novel, effective way to prevent falls and fall-related injuries in people with an iSCI. The impact of added haptic input will also provide insight into the sensorimotor integration abilities of someone with an iSCI. The objective of this study was to investigate the effect of added haptic input on balance control of individuals with iSCI during normal walking and during the challenging condition of tandem (heel to toe) walking. Based on previous research [14], we hypothesized that individuals with iSCI would improve their balance control no differently than AB individuals with added haptic input during both walking conditions and that any changes seen with added haptic input would be correlated to the greater upper UE and reduced LE sensory abilities in someone with an iSCI [14].

Methods

Participants

Adults with iSCI (≥1 year post injury, motor incomplete injury) and age- and sex-matched AB individuals were eligible for this study. Participants were included if they were able to walk 10 m without the assistance of a person or supportive device other than braces. Exclusion criteria included any disease or injury (other than the iSCI) that could impair walking or balance control such as vestibular conditions, musculoskeletal problems, etc. The protocol was approved by the institutional ethics review board (Bio 13-184) and all participants provided informed consent.

Experimental procedure and analysis

Participants were asked to walk normally (NW) and in tandem (TW) along a 10 m walkway with their shoes on and at a self-selected speed. Participants were secured in a safety harness to prevent a fall in case they lost their balance. The safety harness was secured to a ceiling-mounted system permitting support along the line of progression for each walking trial. Tandem walking was included to examine the effect of added haptic input in a walking task more challenging than normal walking [15]. During TW trials, if participants lost their balance and took a wider and/or corrective step, they were instructed to continue walking in tandem upon regaining their balance. On 50% of the trials, participants were asked to lightly (<1 N) touch a railing using their dominant or less impaired index finger (self-reported) throughout the entire trial. For individuals with iSCI, the finger used was based on their ability to hold their finger lightly in place on the railing throughout the trial. The railing was free-standing, on one side of the walkway, parallel to the walking path and was set at standard building code height of 86 cm. The railing was instrumented with force sensors (Futek Advanced Sensor Technology, Inc., Irvine, CA, USA; range 0–5 N) to give online information about the amount of vertical force applied to the railing. If the touch force was greater than 1 N during any trial, participants were instructed to use less force in the subsequent trial. In total, participants walked under four conditions—(i) NW—no touch, (ii) NW—touch, (iii) TW—no touch, and (iv) TW—touch. Both, NW and TW trials were randomized. Touch and no touch trials were paired such that each no touch trial was followed by a touch trial of the same walking style. Three to five trials/condition were collected for each participant. The strides in the middle of the 10 m walkway were used to calculate a mean value (range 4–52 strides/condition/participant), which was used for analyses. All data were used for analyses (corrective steps included) to provide a realistic indication of performance.

Muscle activity was collected to provide insight into the neuromuscular control of walking. Surface electromyography (EMG; $fs = 2000$ Hz) data were collected bilaterally from the tibialis anterior (TA), soleus (SOL), and gluteus medius (GMED) muscles using a telemetered EMG system (2400GT2, Noraxon Inc., Scottsdale, AZ, USA). EMG data were band-pass filtered (20–500 Hz), full-wave rectified, and then low-pass filtered at 10 Hz to calculate a linear envelope. The average EMG values across each stride were calculated for each muscle. Integrated EMG values were then calculated across each stride and normalized to the maximum average EMG values of the walking trials within that walking style. An average integrated EMG of all strides was calculated for each trial. A decrease in muscle activity would reflect an improvement in neuromuscular control of walking [16].

Kinematic data (fs = 100 Hz) were collected from 63 markers (14 mm diameter; 22 calibration) placed over 12 body segments (head, trunk, and right and left upper arms, forearms, thighs, shanks, and feet) using an eight-camera motion capture system (Vicon, Denver, CO, USA). Kinematic data were low-pass filtered at 8 Hz and were used to obtain segmental and total body center of mass. Kinematic data from markers placed on the feet were used to calculate the boundaries of the base of support.

Cutaneous sensation and proprioceptive abilities were measured in individuals with iSCI to determine individual sensory abilities and to examine the relative importance of both upper and lower proprioceptive and cutaneous sensations for sensorimotor integration of added haptic input during walking. For cutaneous sensation, Semmes-Weinstein monofilaments [17] of six different levels were touched to the plantar surface of the big toes (bilateral) and palmar areas (finger used to contact railing) of the skin, with eyes closed, starting from heaviest to lightest six times each. Proprioceptive ability was tested by moving the big toe and ankle of both lower extremities, and shoulder, elbow, wrist, and index finger of the touch side UE into flexion and extension with eyes closed [18]. Participants responded with a yes when they either felt the monofilament against their skin or felt their body moved by the researcher. A total score was calculated by summing the number of correct responses. The maximum possible cutaneous sensation scores were 72 (LE) and 36 (UE-touch side only) and the total possible proprioception scores were 24 (LE) and 48 (UE).

Customized MATLAB scripts were used for data analysis. Stride length was calculated as the distance between two subsequent heel strikes of the same foot. Step width was calculated as the mediolateral distance between heel strike of one foot to the subsequent heel strike of the other foot. An increase in stride length suggests improved walking function [19], whereas an increase in step width would suggest enhanced lateral balance control [20]. Stride velocity was obtained by calculating the first derivative of stride length relative to time, and was normalized to leg length with an increase in stride velocity suggesting improved control of walking [19]. The values for MOS in the ML and AP planes were calculated as the distance between an extrapolated center of mass, and medial and anterior boundaries of base of support, respectively, [21] over the duration of a stride. Increasing the average MOS suggests a reduction in the risk of balance loss [22], while a reduction in variability suggests improved control of the center of mass/base of support relationship [23]. A mean of all strides within a condition was calculated for all measures. Standard deviation for step width, MOS_{ML}, and MOS_{AP} was calculated from all the strides within a condition for each participant and was used as a measure of variability.

Statistical analysis

Means, standard deviation, and ranges were calculated for patient characteristics. The level of force contact was compared between groups using a Mann–Whitney U test due to the non-normal distribution. Kinematic and EMG data were checked for normality using the Shapiro–Wilk test and outliers of averages identified through SPSS (values outside of the 1.5 interquartile range) were removed. If data were then normally distributed, repeated measures ANOVAs were used separately for NW and TW conditions with touch/no touch as within-factors and iSCI/AB groups as between-subject factors to test for main effects with univariate analyses and planned comparisons to examine the direction of differences with Bonferroni corrections to account for the multiple comparisons ($\alpha = 0.05$). Step width in the tandem walking condition was not normally distributed after outliers were removed; therefore, all data were analyzed using a nonparametric Friedman's two-way analysis of variance by ranks. Sensory abilities were correlated with a change in the outcome measures (value during no touch conditions − value during touch conditions) [14] for the participants with iSCI and only for the outcome variables that showed a significant change with the added haptic input. All but one participant with iSCI had full proprioception of their UE and so only LE proprioceptive, and UE and LE cutaneous sensation were correlated to the change in outcome variables using Pearson's r or Spearman's rho correlations for normally and not normally distributed data accordingly. UE cutaneous sensation was correlated to the outcome variables for only the conditions with added haptic input. All analyses were completed using SPSS v25 (IBM Corp).

Results

Eighteen individuals with iSCI and 17 age- and sex-matched AB individuals participated in this study (see Table 1 for demographic information). All participants with iSCI walked for NW trials, but only 12 were able to walk for TW trials. All AB participants completed both NW and TW trials. For the average touch force, data for some participants (iSCI $n = 1$ for NW, $n = 6$ for TW: AB $n = 1$ for TW) were not available due to technical errors. For the data that were available, the mean (SD) level of force was 0.74 (0.39) N ($n = 17$) for the iSCI and 0.62 (0.48) N ($n = 17$)

Table 1 Demographic data for participants.

	Individuals with iSCI		Able-bodied matches ($n=17$)
	Normal walking ($n=18$)	Tandem walking ($n=12$)	
Age in years (mean (SD))	60.6 (18.0)	55.3 (17.0)	62.5 (17.2)
Sex (female:male)	5:13	3:9	4:13
Estimated AIS[a]	All D	All D	–
Level of injury (number (range) cervical:thoracic: lumbar)	10 (C3–5):4 (T4–11):4 (L1–4)	6 (C4–5):3 (T8–11):3 (L1–4)	–
Paraplegia:tetraplegia	8:10	6:6	–
Traumatic:nontraumatic	13:5	8:4	–
Etiology of nontraumatic SCI	Unknown, ossification of the posterior longitudinal ligament, spinal stenosis, transverse myelitis, staph infection		
Time since injury in years (mean (SD))	8.4 (10.7)	5.6 (4.5)	–
Number of participants who wore a brace	3	1	
Upper extremity cutaneous sensation (median (range)/max)	24 (1–34)/36		–
Lower extremity cutaneous sensation (median (range)/max)	33 (0–52)/72		–
Upper extremity proprioception (median (range)/max)	24 (22–24)/24		–
Lower extremity proprioception (median (range)/max)	21 (13–24)/24		–

[a]Based on available medical records.

for the AB participants during NW trials; and 1.30 (0.76) N ($n=12$) for the iSCI and 0.76 (0.78) N ($n=16$) for the AB participants during TW trials. There was no difference between groups for the force applied during NW ($p=0.231$); however, individuals with iSCI applied more force to the railing than AB individuals during TW ($p=0.003$). Average values for groups and conditions are available in a Supplementary File (Supplementary Table 1).

For normal walking, a significant main effect of touch was found ($F_{(14,\ 11)}=5.38$, $p=0.004$, partial $\eta^2=0.873$, observed power $=0.978$). There was no main effect for group ($F_{(14,\ 11)}=0.939$, $p=0.552$, partial $\eta^2=0.544$, observed power $=0.284$) nor any significant interaction ($F_{(14,\ 11)}=0.453$, $p=0.918$, partial $\eta^2=0.366$, observed power $=0.146$) for the kinematic and EMG data. On univariate analysis, stride velocity ($p<0.001$), step width ($p=0.003$), stride length ($p<0.001$), the MOS_{ML} ($p<0.001$), MOS_{ML}_SD ($p<0.001$), MOS_{AP} ($p<0.001$), MOS_{AP}_SD ($p<0.001$) all decreased with touch (Table 2). GMED EMG on the side furthest form the railing increased with touch ($p=0.019$). None of the outcome variables that significantly changed with added haptic input were correlated to UE or LE sensation.

For tandem walking, there was a significant main effect for group ($F_{(13,\ 12)}=4.056$, $p=0.010$, partial $\eta^2=0.815$, observed power $=0.931$) but no main effect for touch ($F_{(13,\ 12)}=2.645$, $p=0.051$, partial $\eta^2=0.741$, observed power $=0.761$) and no significant interaction ($F_{(13,\ 12)}=1.363$, $p=0.299$, partial $\eta^2=0.596$, observed power $=0.429$). On univariate analysis, SW_SD ($p=0.016$) and MOS_{ML}_SD ($p=0.007$) were larger in participants with iSCI compared with the AB group (Table 3). There was no significant effect of group or touch for step width ($p=0.475$).

Discussion

This study investigated the effects of added haptic input in the form of light touch on walking balance control in individuals with iSCI. Despite significant differences in the level of vertical force applied during TW, the level of force was still well below that which could be considered mechanical support [24] and shows that individuals with iSCI and their AB matches can walk normally and in tandem while adding haptic input with a low level of contact on a railing. The results also indicate that adding haptic input impacts walking balance control. Similar to AB individuals, individuals with iSCI show a decrease in their stride velocity, step width, stride length, MOS_{ML}, MOS_{ML}_SD, MOS_{AP}, MOS_{AP}_SD, and an increase in their GMED activity on the limb opposite to the railing for NW trials. In contrast, participants with iSCI walked with a wider step and larger MOS_{ML}_SD compared with AB participants for TW. Sensory abilities were not correlated to

Table 2 Normal walking data comparing main effect of added haptic input.

Outcome variable	Without added haptic input (mean (95% CI))	With added haptic input (mean (95% CI))	Significance of difference (*p* value)
Normalized stride velocity (a.u.)	0.30 (0.26–0.33)	0.25 (0.22–0.29)	**<0.001**
Step width (mm)	82.6 (67.9–97.2)	70.6 (56.1–85.1)	**0.003**
Step width SD (mm)	23.1 (21.7–24.5)	24.4 (21.0–27.7)	0.419
Stride length (mm)	1164.8 (1077.7–1251.82)	1073.5 (984.3–1162.8)	**<0.001**
MOS_{ML} (mm)	111.1 (104.4–117.8)	104.0 (97.7–10.3)	**<0.001**
MOS_{ML}_SD (mm)	23.2 (21.1–25.3)	20.6 (18.5–22.6)	**<0.001**
MOS_{AP} (mm)	522.3 (482.3–562.2)	477.7 (437.1–518.3)	**<0.001**
MOS_{AP}_SD (mm)	166.0 (154.4–177.5)	152.6 (140.9–164.3)	**<0.001**
TA (non-touch side) (a.u.)	1.03 (0.946–1.11)	1.07 (0.98–1.17)	0.222
TA (touch side) (a.u.)	0.981 (0.89–1.07)	1.03 (0.94–1.13)	0.510
SOL (non-touch side) (a.u.)	1.08 (1.01–1.16)	1.17 (1.07–1.26)	0.368
SOL (touch side) (a.u.)	1.13 (1.06–1.21)	1.18 (1.09–1.28)	0.092
GMED (non-touch side) (a.u.)	1.03 (0.92–1.14)	1.11 (0.99–1.23)	**0.019**
GMED (touch side) (a.u.)	1.08 (0.97–1.19)	1.11 (0.97–1.23)	0.446

Data are collapsed across groups as there was no main effect of group. Integrated EMG values were calculated within a stride and an average of all strides was calculated for each trial.

a.u. arbitrary units, *SD* standard deviation, *MOS* margin of stability, *ML* mediolateral, *AP* anterior–posterior, *TA* tibialis anterior, *SOL* soleus, *GMED* gluteus medius.

Bold values represent statistical significance between walking with and without added haptic input.

any of the outcome variables that changed with added haptic input.

A reduction in variability has been reported as the most consistent effect of haptic input on gait in previous studies [25]. In our study, adding haptic input through light touch lead to a reduction in MOS_{ML}_SD and MOS_{AP}_SD. Stride-to-stride variability in gait parameters has been shown to be an independent predictor of falls in older adults [26], therefore a decrease in variability suggests enhanced balance control. The reduced variability may also be an attempt to lessen the motor control challenge while performing an additional, potentially attention-demanding task (i.e., adding haptic input) during walking [27].

A previous study using the same experimental setup in younger healthy adults showed an 11% decrease in normal walking velocity with light touch on a railing [11]. In our study, the velocity during normal walking decreased by 12% and 16% (iSCI and AB, respectively) when lightly touching the railing. In addition to walking slower, there was also a significant reduction in step width and stride length, which may suggest a more cautious gait. The reduction in forward velocity and step parameters can also be due to an increase in attentional demands that maybe associated with maintenance of light touch on a railing [25, 27, 28].

This study also found a significant decrease in MOS_{ML} and MOS_{AP} with added haptic input during normal walking, which is contrary to what we expected. We expected the MOS in both directions to increase resulting in a decreased risk for balance loss. A narrower step width may have led to a narrower base of support, thereby reducing MOS_{ML}. Slower walking velocity and shorter steps may have moved the extrapolated COM away from the anterior boundary of the base of support and closer to the rear boundary, thereby reducing MOS_{AP}.

Changes in LE muscle activation in previous studies have found conflicting results, with no change to a decrease or increase in muscle activity with added haptic input [25]. In our study, the only significant difference was an increase in GMED activity on the side opposite to the railing. The GMED muscle plays an important role in mediolateral stability during walking, thus an increase in activity of this muscle may indicate greater effort of maintaining balance control in the mediolateral direction during normal walking with added haptic input. It is important to note that the relevance of the magnitude of change of the GMED is not known and so, paired with a lack of change in the other LE muscles measured, the increase in GMED activity should be interpreted with caution.

The findings for TW did not reveal any significant changes when haptic input was added, which is contrary to our expectations. We expected equal or even greater improvement in balance control during the more challenging task of tandem walking. In young healthy adults, added haptic input has been shown to significantly reduce the stride velocity and MOS_{ML} during tandem walking [11, 27]. The reason for different results between studies could be more cautious behavior among individuals with

Table 3 Tandem walking data comparing main effect of group.

Outcome variable	Participants with iSCI (mean (95% CI))	AB participants (mean (95% CI))	Significance of difference (p value)
Normalized stride velocity (a.u.)	0.12 (0.09–0.14)	0.12 (0.10–0.14)	0.612
Step width (mm)	28.5 (19.5–37.6)	27.0 (16.4–37.6)	0.303[a]
Step width SD (mm)	18.8 (14.8–22.9)	12.1 (8.7–15.6)	**0.016**
Stride length (mm)	675.1 (616.3–733.9)	638.5 (588.1–688.8)	0.338
MOS_{ML} (mm)	66.7 (61.8–71.6)	67.4 (63.2–71.6)	0.820
MOS_{ML}_SD (mm)	16.4 (14.1–18.7)	12.1 (10.1–14.1)	**0.007**
MOS_{AP} (mm)	325.4 (294.5–356.4)	326.0 (299.5–352.5)	0.977
MOS_{AP}_SD (mm)	103.2 (95.3–111.1)	94.9 (88.2–101.6)	0.111
TA (non-touch side) (a.u.)	1.43 (1.26–1.59)	1.28 (1.13–1.44)	0.189
TA (touch side) (a.u.)	1.24 (1.11–1.37)	1.20 (1.08–1.32)	0.459
SOL (non-touch side) (a.u.)	1.36 (1.16–1.56)	1.47 (1.28–1.65)	0.907
SOL (touch side) (a.u.)	1.35 (1.21–1.49)	1.49 (1.36–1.62)	0.877
GMED (non-touch side) (a.u.)	1.56 (1.34–1.77)	1.43 (1.23–1.63)	0.605
GMED (touch side) (a.u.)	1.42 (1.18–1.66)	1.45 (1.23–1.67)	0.440

Data are collapsed across conditions as there was no main effect of touch.

a.u. arbitrary units, *SD* standard deviation, *MOS* margin of stability, *ML* mediolateral, *AP* anterior–posterior, *TA* tibialis anterior, *SOL* soleus, *GMED* gluteus medius.

[a]Nonparametric test used.

Bold values represent statistical significance between walking with and without added haptic input.

iSCI [29] and comparatively older AB individuals between studies (mean ages = 24.5 years [27]; 25.8 years [11]; and 62.5 years for this study (AB participants only)). Participants in this study may have already adapted a more cautious gait to enhance stability in anticipation of a potential balance perturbation, which leaves little room for improvement in stability with haptic input. Attentional demands created by the act of maintaining light touch on the railing may also have impacted walking behavior [27]. Measures of walking balance control during the TW trials were significantly different between individuals with iSCI and AB individuals such that AB individuals walked with significantly reduced variability in their step width and MOS_{ML}. A reduced variability suggests that AB individuals had a better balance control during tandem walking, irrespective of added haptic input.

The lack of correlations between UE and LE sensations does not support our hypotheses. The differences between the current results and a similar protocol examining standing balance control [14] could be due to the dual motor task paradigm used and, as previously mentioned, the associated attentional demands of adding haptic input through light touch on a railing [27]. The increased attentional demands of walking compared with previous research done in standing [14] may have attenuated the benefits of added haptic input. In addition, while UE and LE sensation was tested, a more comprehensive sensory testing protocol could have provided insight about sensory abilities of other locations on the body (e.g., trunk), which may be important for the successful integration of added haptic input for walking balance control.

There was no group by condition interactions noted in either normal or tandem walking. The lack of interaction suggests that there were fundamental differences in either the condition of adding haptic input (normal walking) or in the ability of the groups to complete the task (tandem walking). Despite the varying sensory capabilities of the individuals with iSCI, lightly touching the railing impacted their normal walking behavior similar to the AB group suggesting that adding haptic input may be a feasible compensatory strategy in a rehabilitation context. Tandem walking, however, seemed more challenging for the individuals with iSCI and adding sensory information through light touch on the railing did not positively impact balance control in this difficult walking task. These results suggest that individuals with iSCI could benefit from added haptic input during normal walking but not for more challenging walking tasks.

Limitations

Participants included individuals with a nontraumatic SCI. As such, there may have been impairments/comorbidities that would have increased variability in the sample and may have affected the ability to integrate the added haptic input during walking. In theory, participants with injuries below the cervical region should have intact UE sensation; however, some of our participants with below-cervical iSCI had reduced UE sensation. There is a chance that there were unknown comorbidities affecting sensation. In addition, details about the level and severity of injury came from medical records and reports from participants. An International Standards for Neurological Classification of Spinal Cord Injury exam could potentially have provided more up to date and accurate reporting of injury characteristics. The participants were not given specific instructions on focusing their attention on the task of walking or maintaining light touch on the railing. There is evidence that adding haptic input through light touch on a railing requires additional attentional resources [27]; therefore, future work could investigate the amount of attention required and whether training with added haptic input could reduce attentional demands in people with iSCI. The sensors on the railing were limited in that they were only able to detect vertical forces applied. Capturing the horizontal forces applied to the railing may give a more comprehensive picture of how haptic input was generated. The safety harness prevented participants from large lateral displacements and may have constrained recovery steps if someone lost their balance in a lateral direction. The harness was not instrumented to detect any tension applied through the connecting cable. Using a fall-prevention system that allows freedom of movement in the full horizontal plane and an instrumented harness would permit all directions of movement and information about use of the safety harness, respectively.

Conclusion

Haptic input in form of light touch can improve balance control during normal walking in individuals with iSCI similar to that seen in AB individuals. Added haptic input can modulate speed, step parameters, and variability of stability margins when walking normally but not during the more challenging task of walking in tandem. UE and LE sensory abilities did not seem to impact the ability to integrate and use added haptic input for walking balance control. Future research should examine the different ways in which haptic input can be incorporated into rehabilitation efforts to benefit individuals with iSCI and prevent falls.

Data availability

The datasets generated and/or analyzed during the current study are available from the corresponding author on reasonable request.

Acknowledgements The authors would like thank all of their participants involved in this study. The authors would like to acknowledge the funding support from the Saskatchewan Health Research Foundation awarded to ARO and KEM.

Author contributions ARO was involved with protocol design; data collection, analysis and interpretation; and drafting and finalizing the manuscript. TA was involved with data collection, analysis and interpretation; and revising the manuscript. JL was involved with protocol design; analysis and interpretation; and revising the manuscript. KEM was involved with protocol design; data interpretation; and revising the manuscript.

Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

Ethics statement We certify that all applicable institutional and governmental regulations concerning the ethical use of human volunteers were followed during the course of this research.

References

1. Burns SP, Golding DG, Rolle WA, Graziani V, Ditunno JF. Recovery of ambulation in motor-incomplete tetraplegia. Arch Phys Med Rehabil. 1997;78:1169–72.
2. Musselman KE, Arnold C, Pujol C, Lynd K, Oosman S. Falls, mobility, and physical activity after spinal cord injury: an exploratory study using photo-elicitation interviewing. Spinal Cord Ser Cases. 2018;4:1–10.
3. Unger J, Oates AR, Arora T, Musselman KE. Prospective monitoring of falls after spinal cord injury: circumstances, causes and consequences. J Spinal Cord Med. 2017;40:849.
4. Khan A, Pujol C, Laylor M, Unic N, Pakosh M, Dawe J, et al. Falls after spinal cord injury: a systematic review and meta-analysis of incidence proportion and contributing factors. Spinal Cord. 2019;57:526–39.
5. Shumway-Cook A, Brauer S, Woollacott M. Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. Phys Ther. 2000;80:896–903.
6. Maki BE, Sibley KM, Jaglal SB, Bayley M, Brooks D, Fernie GR, et al. Reducing fall risk by improving balance control: development, evaluation and knowledge-translation of new approaches. J Saf Res. 2011;42:473–85.
7. Rabin E, Demin A, Pirrotta S, Chen J, Patel H, Bhambri A, et al. Parkinsonian gait ameliorated with a moving handrail, not with a banister. Arch Phys Med Rehabil. 2015;96:735–41.
8. Boonsinsukh R, Panichareon L, Phansuwan-Pujito P. Light touch cue through a cane improves pelvic stability during walking in stroke. Arch Phys Med Rehabil. 2009;90:919–26.
9. Rabin E, Bortolami SB, DiZio P, Lackner JR. Haptic stabilization of posture: changes in arm proprioception and cutaneous feedback for different arm orientations. J Neurophysiol. 1999;82:3541–9.
10. Jeka JJ, Lackner JR. The role of haptic cues from rough and slippery surfaces in human postural control. Exp Brain Res. 1995;103:267–76.
11. Oates A, Unger J, Arnold C, Fung J, Lanovaz J. The effect of light touch on balance control during overground walking in healthy young adults. Heliyon. 2017;3:e00484.
12. Baccini M, Rinaldi LA, Federighi G, Vannucchi L, Paci M, Masotti G. Effectiveness of fingertip light contact in reducing postural sway in older people. Age Ageing. 2007;36:30–5.

13. Franzen E, Paquette C, Gurfinkel V, Horak F. Light and heavy touch reduces postural sway and modifies axial tone in Parkinson's disease. Neurorehabil Neural Repair. 2012;26:1007–14.
14. Arora T, Musselman KE, Lanovaz J, Oates A. Effect of haptic input on standing balance among individuals with incomplete spinal cord injury. Neurosci Lett. 2017;642:91–6.
15. Hedayat I, Moraes R, Lanovaz JL, Oates AR. Different haptic tools reduce trunk velocity in the frontal plane during walking, but haptic anchors have advantages over lightly touching a railing. Exp Brain Res. 2017;235:1731–9.
16. Forero J, Misiaszek JE. The amplitude of interlimb cutaneous reflexes in the leg is influenced by fingertip touch and vision during treadmill locomotion. Exp Brain Res. 2015;233:1773–82.
17. Mueller MJ. Identifying patients with diabetes mellitus who are at risk for lower-extremity complications: use of Semmes-Weinstein monofilaments. Phys Ther. 1996;76:68–71.
18. Gilman S. Joint position sense and vibration sense: anatomical organisation and assessment. J Neurol Neurosurg Psychiatry. 2002; 73:473–7.
19. Thrasher TA, Flett HM, Popovic MR. Gait training regimen for incomplete spinal cord injury using functional electrical stimulation. Spinal Cord 2006;44:357.
20. Bauby CE, Kuo AD. Active control of lateral balance in human walking. J Biomech. 2000;33:1433.
21. Hof AL, Gazendam MGJ, Sinke WE. The condition for dynamic stability. J Biomech. 2005;38:1–8.
22. McAndrew Young PM, Wilken JM, Dingwell JB. Dynamic margins of stability during human walking in destabilizing environments. J Biomech. 2012;45:1053–9.
23. Hollman JH, Kovash FM, Kubik JJ, Linbo RA. Age-related differences in spatiotemporal markers of gait stability during dual task walking. Gait Posture. 2007;26:113–9.
24. Tremblay F, Mireault A-C, Dessureault L, Manning H, Sveistrup H. Postural stabilization from fingertip contact: I. Variations in sway attenuation, perceived stability and contact forces with aging. Exp Brain Res. 2004;157:275–85.
25. Oates AR, Hauck L, Moraes R, Sibley KM. The effects of haptic input on biomechanical and neurophysiological parameters of walking: a scoping review. Gait Posture. 2017;58:232–9.
26. Maki BE. Gait changes in older adults: predictors of falls or indicators of fear? J Am Geriatr Soc. 1997;45:313–20.
27. Awdhan AP, Bone MD, Lanovaz JL, Moraes R, Oates AR. Are there attentional demands associated with haptic modalities while walking in young, healthy adults? Hum Mov Sci. 2019; 66:399.
28. Grabiner MD, Troy KL. Attention demanding tasks during treadmill walking reduce step width variability in young adults. J Neuroeng Rehabil. 2005;2:25.
29. Arora T, Musselman KE, Lanovaz JL, Linassi G, Arnold C, Milosavljevic S, et al. Walking stability during normal walking and its association with slip intensity among individuals with incomplete spinal cord injury. PM&R. 2019;11:270–7.