Investigating User Volitional Influence on Step Length in Powered Exoskeleton Designed for Users with SCI

Xiruo Cheng, Justin Fong, Ying Tan and Denny Oetomo

Abstract— **Volitional movement from users of assistive lower limb exoskeletons may be exploited to increase the controlled variability in the movements of a human-exoskeleton system. This may in turn allow these devices to handle the variability encountered in the terrain of everyday life. This study aimed to investigate the degree to which users can volitionally influence step length, when using an assistive exoskeleton designed for users with spinal cord injury (SCI) running a fixed robotic exoskeleton trajectory. An experiment was conducted to investigate the accessible range of step lengths when five able-bodied participants and one participant with SCI piloted a user-balanced exoskeleton. Participants were asked to take steps as large as possible ("large") and as small as possible ("small"), with the able-bodied individuals asked to minimise use of their leg muscles, with step length of each step measured. Surface electromyography (sEMG) data were collected on major leg muscles of the able-bodied subjects to monitor their muscle activities with a novel processing method introduced to facilitate discussion in the context of users with SCI. The results demonstrate that a user can intentionally manipulate the resulting step length, with every participant having significantly different large and small step sizes ($p < 0.05$). However, large variations were observed between individuals in terms of absolute step lengths and difference between large and small steps. Moreover, the range of step length (normalised by the leg length) ranged from 0.237 to 0.375 for the able-bodied subjects and 0.245 for the individual with SCI. Although positive correlation was present between the sEMG data and resulting step lengths, the result was not statistically significant ($p > 0.05$).**

I. INTRODUCTION

Assistive lower limb exoskeletons have been designed for individuals who have lost lower limb locomotive abilities, such as those with spinal cord injury (SCI) [1]. The intention of using such devices is to augment the function of users' lower limbs, including not only mobility but standing up and other functionalities [2]. Moreover, it has been identified that users' body ownership and being able to take charge of 'artificial limbs' as much as possible is very important [3]. This is a significant point to consider when designing trajectories for exoskeletons users.

For the majority of existing lower limb exoskeletons, with given pre-defined (off-line) trajectories, the robotic motion is autonomous. Recently, there has some research demonstrating that online trajectory generation methods can compute joint trajectories during the step. For example, a biologically-inspired neuromuscular controller was proposed in [4] to generate torque as healthy-like gait requires for rehabilitation exoskeleton. Also in [5], trajectory of one step is divided into four stages and gait parameters could be changed for each stages based on the information of the depth camera and user brain computer inference to avoid obstacles for walking assistive exoskeleton. However, offline trajectories still dominate assistive exoskeletons because it is simple and more predictable for the users [6], [7]. Thus, for most assistive lower-limb exoskeletons, once an off-line trajectory is given, a decision is made at the start of each step and the user is rarely provided with explicit means to alter the motion command to the robot. However, the dynamics of the whole human-exoskeleton system can be affected by human users during each movement as pointed out in [8], [9]. Specifically, users are capable of influencing the overall dynamics by exerting forces against the ground using crutches or by shifting their centre of gravity [10].

This work postulates that the deliberate user influence on the resulting gait can possibly increase user involvement in lower limb exoskeletons, as well as improve the human-exoskeleton system's capability to deal with varied environments without extra sensors and complex control algorithms. This approach is novel in its approach of treating the range of resulting step length as the desirable feature, compared to many current studies [11]–[13], where this is often thought of as an uncertainty in the 'resulting robotic movements' that ought to be suppressed.

Therefore, this work presents an initial study into how the user may volitionally influence the dynamics of the human-exoskeleton system to achieve a given goal. In particular, this work aims at investigating the degree to which human users can influence the outcome of the lower limb exoskeleton movement. Such an ability may be useful when designing the robotic trajectories, as the users may be able to "fill in the gaps" between the nominal trajectories provided when using offline trajectory generation methods. This would result in the human-exoskeleton system being able to adapt to variations in desired gait, whilst simultaneously retaining the simplicity offered by these offline methods. Experimental methods were used to quantify the range of positions in which the exoskeleton user can place their foot, given a single robotic motion trajectory.

II. METHOD

The purpose of this experiment was to investigate the degree to which volitional adjustment to step length the user can make when using a user-balanced exoskeleton. In the experiment, participants were asked to make two types of steps — "as large as possible" and "as small as possible", whilst the lower limb exoskeleton was driven with the

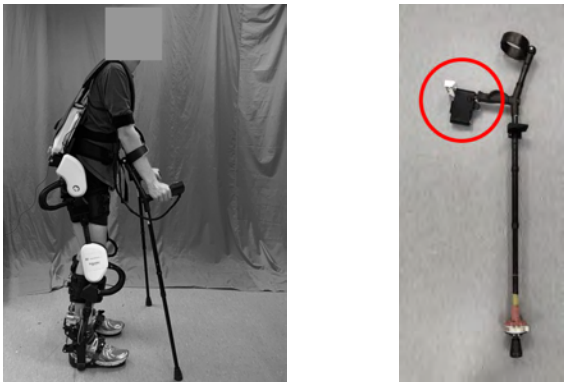
---

The authors are with the University of Melbourne and Fourier Intelligence Joint Laboratory, Melbourne Australia.

Correspondence: `xiruoc@student.unimelb.edu.au`

TABLE I: Participants characteristics

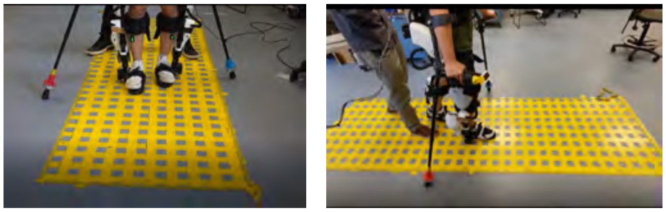
Subject	Gender	Age	Leg length of Exoskeleton/m
AB1	Male	28	1
AB2	Male	25	0.95
AB3	Male	26	0.96
AB4	Female	29	0.95
AB5	Male	23	1.02
SCI1	Male	46	1.02
Mean ± SD		29.5 ± 8.36	0.983 ± 0.031



(a) Able-bodied user wearing user-balanced exoskeleton ExoMotus X2 utilised in the study with active hip and knee joints

(b) Controller (circled) on the right crutch with buttons to select movement and one trigger to initiate movement

Fig. 1: ExoMotus X2 and its user interface



(a) Front camera view

(b) Side camera view

Fig. 2: Camera views used for foot placement localisation. Note the ground grid, which is marked by yellow tape in the figure, provided a fixed 0.05m grid size.

same nominal trajectory. It was assumed that any step sizes between the largest and smallest steps were achievable, and therefore that the results would give the range of achievable volitional adjustments for the given individual and the given exoskeleton trajectory. In addition, the able-bodied participants were asked to try to not activate their leg muscles. Surface electromyography (sEMG) were used to measure this activation. The aim was that this would approximate the condition of a person with no volitional leg control such as people with SCI. However, it is acknowledged that this approximated behaviour does not necessarily reflect the realities of all individuals with SCI, as muscle activation of individuals with SCI may not always be zero [14]. For example, individuals with incomplete spinal cord injuries may have some volitional control of the muscles, and even those with complete injuries may have high muscle tone or spasticity [15] [16], which would be measured by sEMG.

A. *Participants*

Five able-bodied participants (AB1-AB5) with no known underlying conditions affecting their motor control, and one participant with SCI (SCI1, T10, ASIA A) participated in this study, with lower limb lengths ranging from 0.95 m to 1.02 m. The detailed information of participants is displayed in Table. I. All participants gave informed consent, which was approved by the University of Melbourne Human Research Ethics Committee under application 20528.

B. *Materials*

1) Exoskeleton: A modified assistive user-balanced lower limb exoskeleton ExoMotus X2 (Fourier Intelligence, Shanghai, China) was used in this study. The exoskeleton had four active degrees of freedom (hip and knee flexion/extension) as shown in Fig. 1a. A user interface is mounted on the right crutch (in Fig. 1b). In the experiment, a customised software, which is developed using CANOpen Robot Controller (CORC) [17], is used.

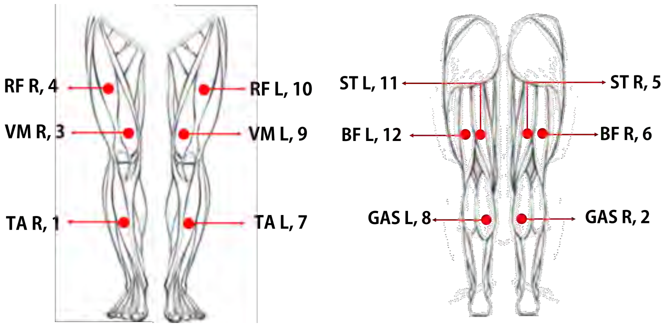
Exoskeleton movements were selected and initiated by the user through pressing buttons presented on the user interface of instrumented crutch (as Fig. 1b). A single nominal gait trajectory was used for all steps in this experiment, with a nominal step length normalised by leg length (as per [18]) of 0.324, and a preprogrammed step duration of 2 seconds.

2) Ground grid: Step sizes were measured using a simple visual approach, which is relatively convenient to set up within the limited confines of the laboratory. A flat ground marked with a grid (shown in Fig. 2) is used when participants were walking in the exoskeleton. The overall size of the grid measured 3.05 m× 0.9 m with a grid size of 0.05 m. The experiment was recorded by three cameras — two at the left and right side of the grid, and one at the front of the grid. The placement of each foot was identified by a researcher by comparing the position of the foot tip against the grid in the video recording. The accuracy has been tested by randomly selecting 20 points on the grid, the averaged measurement errors among these points is around ±0.012 m, which is a reasonable accuracy for this experiment.

3) sEMG: During the experiment, able-bodied participants were asked to try not use their lower limb muscles. To measure the muscle activities in the able-bodied participants, twelve surface electromyography (sEMG) sensors (Delsys Trigno, Delsys, USA) were used. sEMG sensors were placed on the major muscles in the leg, which primarily drive dorsiflexion, plantarflexion, knee extension and knee flexion motions [19], as shown in Fig. 3. Such sEMG measurements were used to evaluate the degree of leg muscle activation of able-bodied participants during the experiment.

C. *Experiment Protocol*

1) Preparation: Before data collection began, participants were given instructions on how to operate the exoskeleton, with the exoskeleton suitably adjusted and donned for each



(a) Front view (b) Back view

Fig. 3: Location of Surface Electromyography (sEMG) sensors placed on the legs of the able-bodied participants: Tibialis anterior (TA), Gastrocnemius (GAS), Vastus medialis(VM), Rectus femoris (RF), Semitendinosus (ST) and Biceps femoris (BF), with L and R representing left and right legs respectively. TA mainly dorsiflex the foot and GAS is in charge of plantar flexion. VM and RF function as knee extensors meanwhile ST and BF work as knee flexors [19].

participant. Participants were then asked to familiarise themselves with operating and walking with the exoskeleton over a ten metre walk training area. Participants were given as much time as required to practise using the exoskeleton, to the point where the participant confirmed they were comfortable with its operation, and were able to complete five consecutive steps without external assistance.

Only for the able-bodied participants, once the familiarisation session was complete, sEMG sensors were placed on the designated muscles after their skin were sanitized and dry. Then a standard Maximum Voluntary Contraction (MVC) assessment was completed, as identified in [20], [21]. sEMG data was not collected for the participant with SCI.

2) Data collection and preprocessing: The experiment was organised in several runs. Each run consisted of a single pass of the ground grid (Fig. 2) where the participant was instructed to either take steps as large as possible or take steps as small as possible. Runs were repeated until the participant had completed a minimum of 10 steps of each step type, with the step type for the first run randomised for each participant. Able-bodied participants were instructed to try not use their lower limb muscles during the experiment, however, this was not enforced.

For the able-bodied participants, sEMG data was collected using EMGworks Acquisition (Delsys Incorporated, USA) with sampling rate of 2148 Hz. Video data was simultaneously captured from three cameras.

The video data and sEMG were synchronised offline, and segmented into 2-second blocks representing the time that the exoskeleton trajectory was moving in each step. Steps in which the participant required physical assistance to balance were discarded.

D. Metrics

This study focused on the resulting step length for all subjects, and the muscle activation for the able-bodied subjects. It is reiterated for clarity that participants were using a lower limb exoskeleton with the same nominal robotic trajectory, suggesting that variations in step length were due to the specific actions of that individual during that step.

1) Normalised step length: The step length was defined as the anterior-posterior distance between the front of the stance foot and the front of the swing foot at heel strikes.

The step length was measured through a simple approach using three cameras (as described in Section II-B.2), with visual comparison of the foot landing position to the grid. The multiple camera angles associated with each video were cross-referenced. The accuracy of this method were tested, showing an acceptable accuracy.

The measured step length of the j^{th} subject at the i^{th} step was denoted as $s_{j,i}$. Where relevant, a step length in which the participant was asked to make a largest step, was denoted as $s^l_{j,i}$, and for a smallest step as $s^s_{j,i}$. We use $s^r_{j,i}, r = s,l$ to represent the step length of i^{th} step of the j^{th} subject.

To allow comparisons between participants, each step was normalised by the length of the leg, as measured from the acetabulum (right above hip joint) to the base of the heel. Specifically, the length of the i^{th} step from the j^{th} subject ($s_{j,i}$) was normalised by subject leg length L_j, calculated as:

$$\bar{s}^r_{j,i} = \frac{s^r_{j,i}}{L_j}, \tag{1}$$

for $r = s,l$.

2) Muscle activation: The aim of measuring the muscle activation was to investigate whether (a) the able-bodied individual were able to 'emulate' an individual with SCI, who does not have any volitional control of his/her muscles and (b) whether muscle activation had any correlation with resulting step sizes. Thus, for the j^{th} able-bodied subject, this work proposes a novel metric of 'Activation level' E_j — a basic metric which represents the total muscle activity of the measured muscles in the legs, as a percentage of MVC. Specifically, this metric was calculated for each step as follows:

1) For the j^{th} participant, the i^{th} step, and the k^{th} sEMG sensor, the raw sEMG data $m^r_{j,i,k}(t)$ was filtered by a digital finite impulse response (FIR) bandpass filter with passband from 15 Hz to 400 Hz, then rectified, offset removed and normalised to the measured MVC of each muscle, to produce a rectified sEMG signal $m_{j,k,i}(t)$.
2) The resulting signal $m_{i,j,k}(t)$ was integrated and normalised by the time taken for each step, using a trapezoidal approximation of the sampled data. That is, for the i^{th} step of the j^{th} participant, with the starting time $t_1^{j,i}$ and the finishing time $t_N^{j,i}$ within each step, it was assumed that $N^{j,i}$ samples are obtained for the signal $m_{j,k,i}(t_l), l = 1,\ldots N^{i,j}$. Consequently, the time integral of $m_{j,k,i}(\cdot)$, denoted as $M_{j,k,i}$, was computationally ap-

proximated as:

$$
\begin{aligned}
M_{j,k,i} &= \int_{t_1^{j,i}}^{t_N^{j,i}} m_{j,k,i}(t)dt \\
&\approx \frac{t_N^{j,i} - t_1^{j,i}}{2N^{j,i}} \sum_{n=1}^{N^{j,i}-1} \left(m_{j,k,i}(t_n^{j,i}) + m_{j,k,i}(t_{n+1}^{j,i})\right). \qquad (2)
\end{aligned}
$$

3) Finally, the activation level for the j^{th} participant (E_j) was defined as the average $M_{i,j,k}$ over all steps and all muscles:

$$E_j = \frac{1}{S_j}\frac{1}{12}\sum_{i=1}^{S_j}\sum_{k=1}^{12} M_{i,j,k}, \qquad (3)$$

where S_j was the total number of steps performed by the j^{th} subject ($j = 1, \ldots, 6$).

3) Statistical analysis: An Anderson-Darling test was used to confirm if the results of normalised step lengths $s^r_{i,j}$ or muscle activation level E_j followed a normal distribution. For each participant, a Student's t-test was used to compare the normalised step lengths captured for the smallest steps (*i.e.* the set of all $\bar{s}^s_{j,i}$) against those captured for the largest steps (that of $\bar{s}^l_{j,i}$). A significant difference between the two indicates that the individual is able to volitionally modulate their step size.

The correlation between muscle activation level and the mean of normalised step lengths for each subject was assessed by Pearson correlation coefficient, to assess whether differences in resulting step lengths were driven by muscle activation in the able-bodied subjects.

All calculations were performed using the MATLAB Statistics and Machine Learning Toolbox Release 2019b (MathWorks Inc., United States).

III. RESULTS

Five able-bodied participants and one participant with SCI completed the protocol, with the resulting number of the small or the large steps. The total number of small steps completed by each participant, $S^s_j, j = 1, \ldots, 6$, was with the range of $[13, 20]$. Similarly, the number of completed large steps for each subject, $S^l_j, j = 1, \ldots, 6$ was within $[10, 24]$.

Fig. 4 illustrates the normalised step lengths $\bar{s}^s_{j,i}$ and $\bar{s}^l_{j,i}$ for each participant. All participants had shown that their largest steps are significantly different from their smallest steps $p < 0.05$.

The range of step length (difference between longest steps and shortest steps) for AB1-AB5 ranged between 0.237 and 0.325, and was 0.245 for SCI1. The standard deviation of large steps for AB1-AB5 ranged between 0.052 and 0.0732 and was 0.045 for SCI. The standard deviation of small steps for AB1-AB5 were between 0.041 and 0.057 and was 0.06 for SCI1.

In terms of muscle activation, Fig. 5 shows the activation level (E_i defined in (3)) for each subject against that subject's mean small and large steps. Activation levels ranged between 0.0318 and 0.1257. The correlation between activation level and the mean step length of large steps and smalls steps were computed. The Pearson's linear correlation coefficient ρ between activation level and the mean of normalised large steps was 0.8401 with $p = 0.0749$. ρ between activation level and the mean of normalised small steps is 0.8261 with $p = 0.0848$. Although correlation coefficient $\rho > 0.8$, p-values for testing the hypothesis of no correlation between two vectors are larger than 5%, indicating that the correlation result is statistically insignificant.

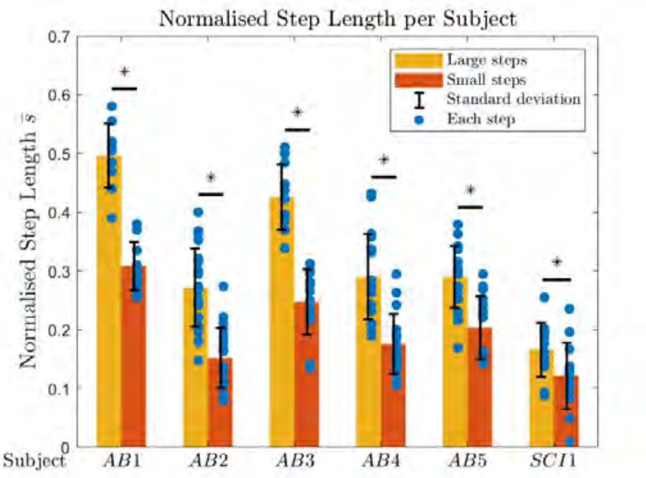


Fig. 4: Normalised step lengths of each subject. Circles represent each normalised step length (i.e. all data points in the experiment). Bar represents the mean of large or small step types for each subject with the standard deviation. Significant difference between large steps and small steps was observed for participants AB1-AB5 ($p < 0.001$) and SCI1 ($p < 0.05$)

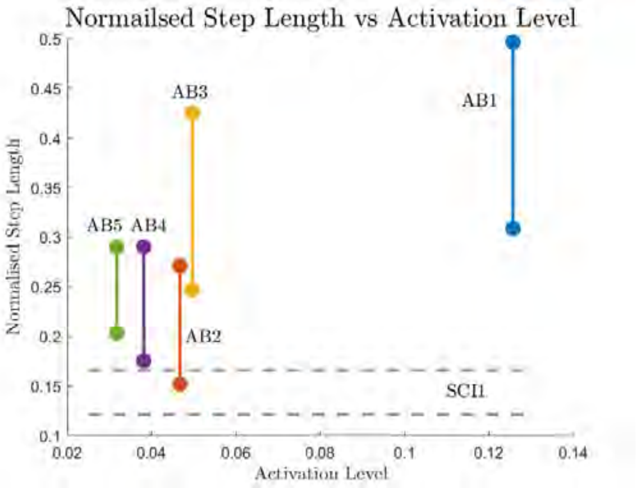


Fig. 5: Activation level versus mean normalised step lengths for both large and small step types.

IV. DISCUSSION

A. Voluntary Adjustments to Step Length

The step length results suggest that the user does have the ability to affect step length, with the normalised step lengths ranging from 0.001 to 0.580 among all participants in this study. It is noted that the nominal normalised step length — the trajectory as was designed based on the exoskeleton's kinematics — was 0.3235.

The results also suggest that all participants were able to make both large and small steps, as proven by the significant difference between large steps and small steps for each individual ($p < 0.05$). This again validates the hypothesis that the exoskeleton user is able to actively modulate their step length even if a single exoskeleton trajectory is used. However, the ranges achieved by each individual vary greatly — with the mean normalised large step varying from 0.50 to 0.27 for the able-bodied participants, with the equivalent for the small steps ranging from 0.31 to 0.15. SCI1 hada mean large step size of 0.17 and mean small step size of 0.12 — which are far smaller than all able-bodied participants. Additionally, in general the participants with larger s^l_j have larger s^s_j.

It is also noted that the range of normalised step lengths for each subject, ranged from 0.237 to 0.375 for the able-bodied participants and 0.245 for the participant with SCI, which corresponds to absolute values of 0.225 m to 0.36 m. This reveals the degree to which user can influence step length of the whole system. Therefore, each participant's capability to affect step length was different, which may be important when designing exoskeleton trajectories. It also suggests the designed trajectories of lower-limb exoskeleton should have the ability to adapt for each individual. Moreover, additional training may be required to enable consistent results.

The average standard deviation across individuals for all large steps was 0.058 and for all small steps was 0.051. There are two interesting points to note about this. First, the fact that this standard deviation is relatively consistent across individuals suggests that the variance might be inherent in the experimental setup, such as the mechanical design of the exoskeleton produced in this study. Secondly, this is a relatively large standard deviation — the leg length is approximately 1 m, thus this standard deviation represents approximately 0.05 m. Whilst this may be related to the ability of the participants to precisely (repeatedly) place their foot in a given position, it may also be a reflection of the instructions given to the participants. A further study with a targeted foot position is needed to more thoroughly understand the accuracy and precision of the user's ability to place their foot.

B. Muscle Activation during Experiment

The activation level shown in Fig. 5 for each subject provides an indication of the able-bodied participants' abilities to minimise the use of their leg muscles. Whilst it is clear that the exoskeleton itself can be used by individuals with no control over their leg muscles (as evidenced by SCI1's results), the discrepancy between the step lengths of SCI1 when compared to the able-bodied participants' suggests a significant difference in either ability or use. Nevertheless, the activation level for AB2 to AB5 are relatively small compared to that of AB1, which means that their leg muscles are less activated during the experiment. This also implies that the step length results of AB1, which has larger step length might be influenced by active lower limb movement. This may motivate further investigations into the use of able-bodied individuals as proxies for assistive devices designed for individuals with SCI.

C. Limitations and Future Work

This work was motivated by the idea that via quantifying the volitional movement possible by an exoskeleton user, a control strategy for exoskeletons may be developed which allows the pilot to step in a continuous range of places with a limited number of predetermined trajectories. However, there are a number of potential limitations to be overcome in order to achieve this goal.

First, this result is highly exoskeleton-specific, which means that different exoskeleton designs, structures, rigidity and control structures would have a big effect on the results. This is consistent with works in the literature. For example, [22] suggested that step length can be increased by reducing trunk flexion via improving shoulder and chest strapping design. The degree to which user can affect the step length could be smaller when using control strategies which are intended for precisely tracking the desired trajectory against uncertain exoskeleton dynamics, like the control method proposed in [23]. As such, whilst the results have relevance for development of control strategies for the specific ExoMotus X2 Exoskeleton used in this work, the results are unlikely to be transferable to other devices. However, it is also noted that the results are also user-specific, and thus it is likely that some sort of calibration procedure will be required to apply such a strategy to any user on any device.

Secondly, the methods used to measure step length are subject to human errors. This measured step length may be affected by the estimated error from parallax and orientation changes as a result of the stepping motion. Although the participants were instructed to only walk in a forward direction, it is inevitable that the human-exoskeleton system will experience small amounts of twist and swivel during walking.

Finally, the results of the present study suggest that there is possibly a discrepancy between the performance of an able-bodied individual whilst using an exoskeleton, and an individual with SCI. Although this is not necessarily a surprising or novel conclusion, it does raise the question of to what degree the result of testing with such able-bodied individuals is meaningful. Using able-bodied individuals in tests has clear advantages in safety and availability of testers. However, the early results presented in this work (and additional analysis not presented here) suggest that an able-bodied individual's natural, reflexive actions could be different from an individual with SCI. This may indicate that testing control strategies with able-bodied individuals might not provide useful insights, even if the individuals are instructed to 'act' like an individual who has SCI. Though simultaneously acknowledging that the every SCI is different, the authors believe the work exploring that the suitability of using able-bodied individuals for testing assistive devices may provide clarity around when such arrangements are appropriate, and significantly accelerate and improve the quality of the outcomes in the field.

V. CONCLUSION

This study investigated how much volitional influence users have on step length when they are wearing a lower-limb exoskeleton to walk. The experimental results demonstrate that the user can intentionally manipulate step length. The range of volitional adjustment found in this study is human-dependent and exoskeleton-specific, and thus suggests that customisation or training will be required to implement a control scheme, which can take advantage of our observations found in walking with an exoskeleton.

ACKNOWLEDGEMENT

This work was funded through an Australian Research Council Linkage Project Grant (LP180101074).

REFERENCES

[1] H. Lee, P. W. Ferguson, and J. Rosen, “Lower limb exoskeleton systems—overview,” in *Wearable Robotics*. Elsevier, 2020, pp. 207–229.

[2] B. Chen, H. Ma, L.-Y. Qin, F. Gao, K.-M. Chan, S.-W. Law, L. Qin, and W.-H. Liao, “Recent developments and challenges of lower extremity exoskeletons,” *Journal of Orthopaedic Translation*, vol. 5, pp. 26–37, 2016.

[3] M. Pazzaglia and M. Molinari, “The embodiment of assistive devices—from wheelchair to exoskeleton,” *Physics of life reviews*, vol. 16, pp. 163–175, 2016.

[4] A. R. Wu, F. Dzeladini, T. J. Brug, F. Tamburella, N. L. Tagliamonte, E. H. Van Asseldonk, H. Van Der Kooij, and A. J. Ijspeert, “An adaptive neuromuscular controller for assistive lower-limb exoskeletons: A preliminary study on subjects with spinal cord injury,” *Frontiers in neurorobotics*, vol. 11, p. 30, 2017.

[5] H. Ren, W. Shang, N. Li, and X. Wu, “A fast parameterized gait planning method for a lower-limb exoskeleton robot,” *International Journal of Advanced Robotic Systems*, vol. 17, no. 1, p. 1729881419893221, 2020.

[6] D. Shi, W. Zhang, W. Zhang, and X. Ding, “A review on lower limb rehabilitation exoskeleton robots,” *Chinese Journal of Mechanical Engineering*, vol. 32, no. 1, pp. 1–11, 2019.

[7] T. Yan, M. Cempini, C. M. Oddo, and N. Vitiello, “Review of assistive strategies in powered lower-limb orthoses and exoskeletons,” *Robotics and Autonomous Systems*, vol. 64, pp. 120–136, 2015.

[8] S. Ahmed, H. Wang, and Y. Tian, “Robust adaptive fractional-order terminal sliding mode control for lower-limb exoskeleton,” *Asian Journal of Control*, vol. 21, no. 1, pp. 473–482, 2019.

[9] J. C. Moreno, F. Brunetti, E. Navarro, A. Forner-Cordero, and J. L. Pons, “Analysis of the human interaction with a wearable lower-limb exoskeleton,” *Applied Bionics and Biomechanics*, vol. 6, no. 2, pp. 245–256, 2009.

[10] F. Xu, J. Qiu, W. Yuan, and H. Cheng, “A novel balance control strategy based on enhanced stability pyramid index and dynamic movement primitives for a lower limb human-exoskeleton system,” *Frontiers in neurorobotics*, vol. 15, 2021.

[11] D. L. Castro, C.-H. Zhong, F. Braghin, and W.-H. Liao, “Lower limb exoskeleton control via linear quadratic regulator and disturbance observer,” in *2018 IEEE International Conference on Robotics and Biomimetics (ROBIO)*. IEEE, 2018, pp. 1743–1748.

[12] S. G. Khan, M. Tufail, S. H. Shah, and I. Ullah, “Reinforcement learning based compliance control of a robotic walk assist device,” *Advanced Robotics*, vol. 33, no. 24, pp. 1281–1292, 2019.

[13] M. S. Amiri, R. Ramli, and M. F. Ibrahim, “Hybrid design of pid controller for four dof lower limb exoskeleton,” *Applied Mathematical Modelling*, vol. 72, pp. 17–27, 2019.

[14] F. Sylos-Labini, V. La Scaleia, A. d’Avella, I. Pisotta, F. Tamburella, G. Scivoletto, M. Molinari, S. Wang, L. Wang, E. van Asseldonk *et al.*, “Emg patterns during assisted walking in the exoskeleton,” *Frontiers in human neuroscience*, vol. 8, p. 423, 2014.

[15] L.-P. Hiersemenzel, A. Curt, and V. Dietz, “From spinal shock to spasticity: neuronal adaptations to a spinal cord injury,” *Neurology*, vol. 54, no. 8, pp. 1574–1582, 2000.

[16] C. K. Thomas, M. Dididze, A. Martinez, and R. Morris, “Identification and classification of involuntary leg muscle contractions in electromyographic records from individuals with spinal cord injury,” *Journal of Electromyography and Kinesiology*, vol. 24, no. 5, pp. 747–754, 2014.

[17] J. Fong, E. B. Küçüktabak, V. Crocher, Y. Tan, K. M. Lynch, J. L. Pons, and D. Oetomo, “CANopen robot controller (CORC): An open software stack for human robot interaction development,” in *International Symposium on Wearable Robotics*. Springer, 2020, pp. 287–292.

[18] A. L. Hof, “Scaling gait data to body size,” *Gait & posture*, vol. 3, no. 4, pp. 222–223, 1996.

[19] R. R. Neptune, I. C. Wright, and A. J. Van Den Bogert, “Muscle coordination and function during cutting movements.” *Medicine and science in sports and exercise*, vol. 31, no. 2, pp. 294–302, 1999.

[20] D. J. Rutherford, C. L. Hubley-Kozey, and W. D. Stanish, “Maximal voluntary isometric contraction exercises: a methodological investigation in moderate knee osteoarthritis,” *Journal of Electromyography and Kinesiology*, vol. 21, no. 1, pp. 154–160, 2011.

[21] A. Ghazwan, S. M. Forrest, C. A. Holt, and G. M. Whatling, “Can activities of daily living contribute to emg normalization for gait analysis?” *PLoS One*, vol. 12, no. 4, p. e0174670, 2017.

[22] A. J. J. Smith, “Modeling human dynamics for powered exoskeleton control,” Ph.D. dissertation, Université d’Ottawa/University of Ottawa, 2019.

[23] S. Ahmed, H. Wang, and Y. Tian, “Model-free control using time delay estimation and fractional-order nonsingular fast terminal sliding mode for uncertain lower-limb exoskeleton,” *Journal of Vibration and Control*, vol. 24, no. 22, pp. 5273–5290, 2018.