

Unilateral stiffness modulation with a robotic hip exoskeleton elicits adaptation during gait

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Abstract—Wearable robotic exoskeletons show promise in their ability to provide gait assistance and rehabilitation in real-world contexts. However, a better understanding is needed of how exoskeletons contribute to neural adaptation in locomotion, a critical component of neurological gait rehabilitation. We tested whether unilateral perturbations elicit neural adaptation in healthy participants using a novel robotic hip exoskeleton, taking inspiration from asymmetry augmentation strategies used in split-belt treadmill training. We found that applying a virtual stiffness parallel to the hip joint on one side elicited changes in hip range of motion and step length, and that these changes were time varying, indicating an adaptation response. However, participants converged on asymmetric hip ranges of motion and step lengths both with and without applied stiffness from the exoskeleton. These results suggest that while adaptation appears to have occurred, it was not solely driven by the nervous system reducing gait asymmetry. Our findings indicate that applying mechanical impedance asymmetrically to the joints may be an effective gait training and rehabilitation approach, as well as a method to elicit a novel adaptation response to further study neuromotor control of locomotion.

I. INTRODUCTION

Stroke is the leading cause of long-term disability and leaves 80% of its survivors with some form of locomotor impairment [1]. Locomotion is one of the most important aspects of human mobility, and any form of locomotor dysfunction can greatly diminish one’s quality of life. As such, regaining locomotor function is the goal most often stated by stroke patients [2].

Locomotion post-stroke is often asymmetric due to hemiparesis or hemiplegia (slight or severe paralysis of one side of the body) [3]. Functionally, those with asymmetric gait post-stroke walk at slower speeds [4] and are at a higher risk of falling [5] compared to typical gait. Asymmetric gait can cause excessive, abnormal joint loading that can degrade musculoskeletal health in the long-term [6].

In recent years, split-belt treadmill training has emerged as a potential approach to correct gait asymmetry caused by a neurological injury and/or disorder [7]. In split-belt treadmill training, the belts of a dual-belt treadmill run at different

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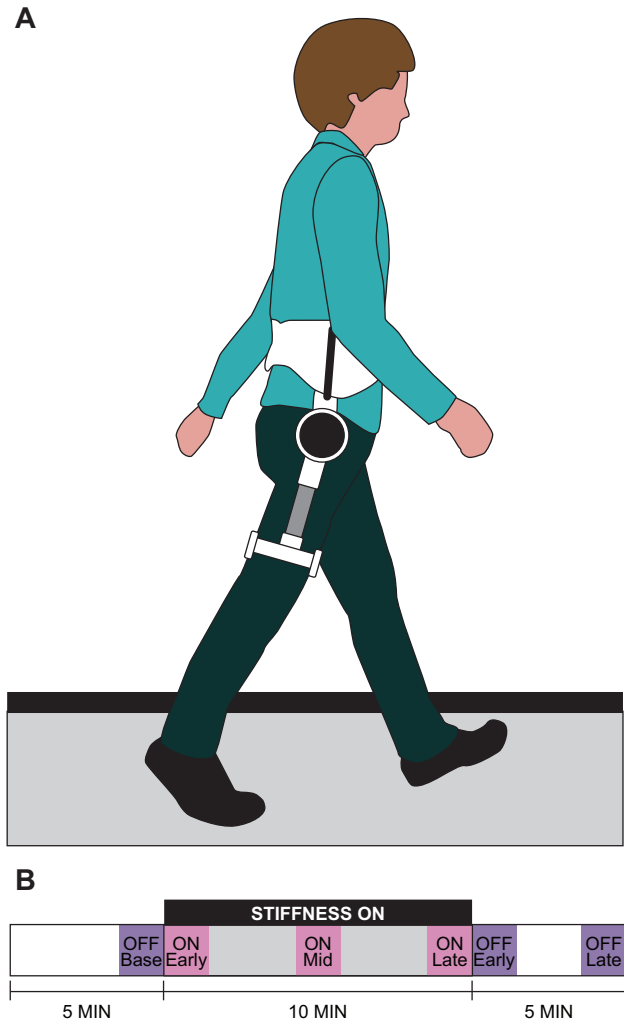


Fig. 1. **A** Experimental Setup. Participants were instructed to walk comfortably on treadmill while wearing a hip exoskeleton that emulated a virtual, rotation spring about the right hip joint. **B** Experimental protocol consisted of 5 minutes of baseline walking (stiffness controller OFF), 10 minutes of walking with stiffness ON, and 5 minutes of stiffness OFF.

speeds under each foot to exaggerate existing step length asymmetries and induce gait compensations. Research has shown that prolonged, repeated exaggeration of step length asymmetry can elicit neural adaptations to normalize step length asymmetries for steady-state walking [8]. This training results in more symmetrical step lengths for people with gait dysfunction caused by neurological disorders [9]. However, these aftereffects are short-lived and training effects produced on the treadmill show limited transfer to overground walking [10].

The recent rise of autonomously powered, wearable robots offer a potential solution to the short-comings of split-belt treadmill training [11]. Unlike treadmill-based interventions, wearable robots can directly (re)train gait in real-world contexts, even during activities of daily living. In addition, wearable robots are more accessible due to their small size and potentially lower cost. Before wearable robots can live up to their promise of enhancing gait recovery, however, a better understanding of how human gait behavior responds to robotic exoskeleton interventions, and most importantly what drives those behaviors, is needed.

The purpose of the present study was to (1) test if unilateral application of a virtual stiffness using a wearable hip exoskeleton elicits neural adaptation and (2) if so, assess whether or not that adaptation is driven by the nervous system reducing gait asymmetry. Addressing this open question is important as it may point to new approaches for correct gait asymmetry in neurological patients. Previous work found that applying a bilateral virtual stiffness using a hip exoskeleton resulted in immediate changes in the gait patterns, and no aftereffects were observed after the removal of the intervention [12]. Such immediate changes in the gait pattern suggested that the intervention did not evoke neural adaptation nor persistent changes in the neuromotor control. However, the application of a bilateral virtual stiffness does not affect symmetry between the two legs, whereas the application of a unilateral virtual stiffness does. Thus, we predicted that the application of a unilateral virtual stiffness with a hip exoskeleton would result in time-varying behavioral changes that could be attributed to neural adaptation (Fig. 1A).

The remainder of the paper is organized as follows. Section II describes the experimental design and methods used to evaluate the effect of unilateral modulation of hip joint stiffness on gait kinematics and asymmetry. Section III presents results which are discussed in Section IV. Concluding remarks follow in Section V.

II. METHODS

A. Participants

Five healthy, young adults (sex: 2 female, 3 male; age: 20.8 ± 1.3 years; height: 172.6 ± 13.5 cm; mass: 72.7 ± 16.8 kg) participated in this study. None had previously worn a hip exoskeleton nor partook in a similar experiment. All subjects gave informed written consent before the experiment. The experimental protocol was reviewed and approved by the Institutional Review Board of the University of Massachusetts Amherst.

B. HRSL Hip Exoskeleton

The hip exoskeleton used in this study was developed by Human Robot Systems Laboratory (HRSL) at the University of Massachusetts Amherst (Fig. 2). This lightweight robotic exoskeleton is modular and can be configured to apply torque either unilaterally or bilaterally about the hip joints in the sagittal plane. The weight of the exoskeleton is 1.75 kg. Actuator torque is transmitted to the human through

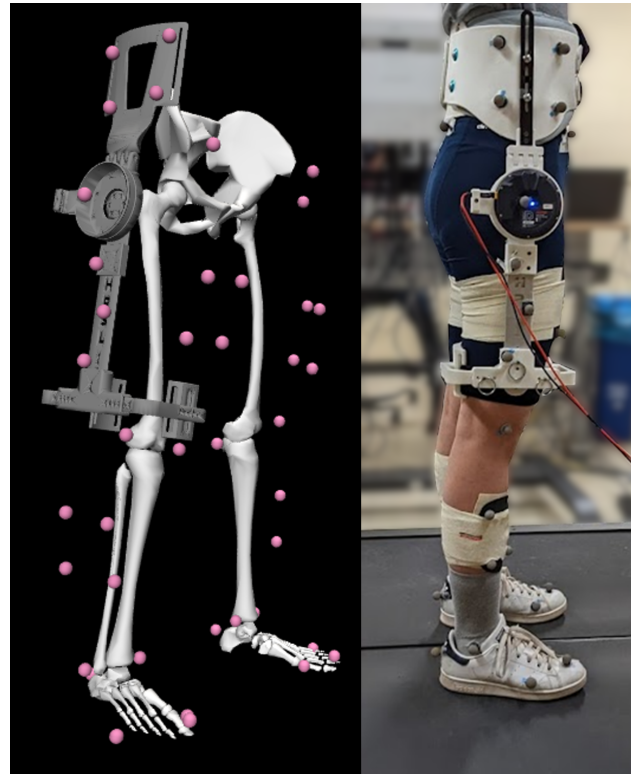


Fig. 2. OpenSim models and corresponding marker placements were generated to match experimental participants. Models were used to recreate kinematics for the biological joints as well as the exoskeleton.

thigh frames designed specifically for the application of mechanical impedance at the hip joints. Passive hinges allow for hip abduction and adduction in the frontal plane.

Each actuator contains a brushless DC motor with a 6:1 gearhead and an absolute encoder, along with additional sensors and electronics (ActPack 4.1, Dephy, Maynard, MA, USA). Output torque from the actuator is estimated and controlled by sensing electrical current in the motor. All electronics and actuators are located onboard the hip exoskeleton. Because the current experiment was performed on a treadmill, the power source was offboard. Placing an onboard power source is feasible for autonomous operation.

C. Stiffness Controller

For the current experiment, the hip exoskeleton was configured in unilateral mode with the actuator on the right hip joint. The actuator emulated a virtual, rotational spring on the right leg using the following control law: $\tau_R = k\theta_R$, where τ_R is the motor torque applied to the right hip joint, θ_R is right hip angle defined relative to an upright standing position as measured by the encoder in the actuator, and k is the stiffness of the spring, which was set to 7.0 Nm/rad for this experiment. The stiffness controller was turned off by setting $\tau_R = 0$.

D. Experimental Procedure

The experiment took place in a single session for each participant. Each session was began with an exoskeleton fitting and adjustment period to ensure a snug, comfortable fit. This was followed by a short recording of quiet standing

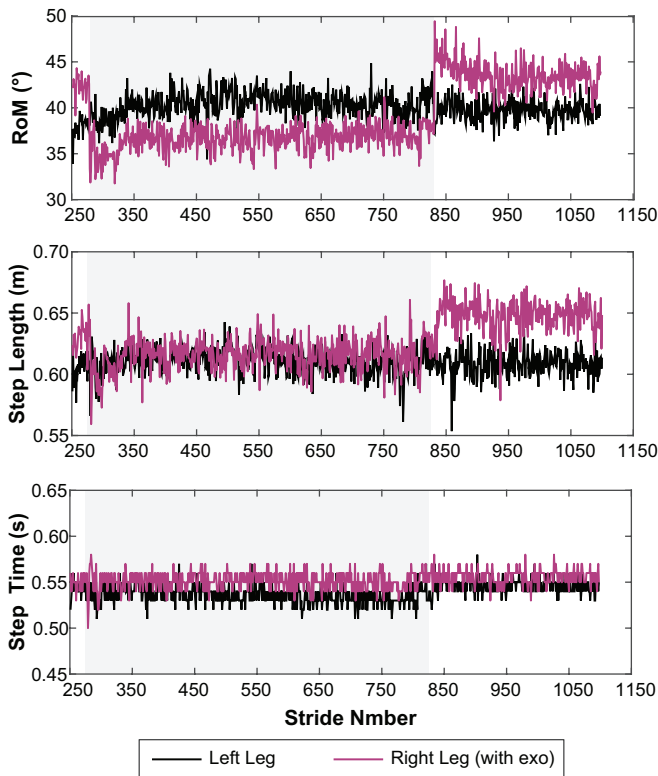


Fig. 3. Dependent measures of a representative participant. Shaded regions represent when the stiffness controller was on.

while wearing the powered-off exoskeleton to ensure that all markers were visible in the calibrated capture volume and to provide a static reference for data analysis. Each participant then performed one trial in which they walked at 1.30 m/s for a total of 20 minutes on a split-belt treadmill (Bertec Corporation, Columbus, OH, USA). The stiffness controller was turned off during the first five minutes, turned on for ten minutes, and turned off again for the remaining five minutes.

E. Data Collection

Reflective markers were placed on each participant for model scaling and motion tracking and their positions were recorded at 100Hz with an eight-camera motion capture system (Qualisys, Inc., Gothenburg, Sweden). The motion capture system was calibrated immediately before each experiment per manufacturer specifications. A total of 45 markers were placed on each participant, locating the pelvis, thighs, shanks, feet, and exoskeleton in 3D space.

F. Data Processing

3D models of the human-exoskeleton system were created in OpenSim 4.3 [13] for each participant based on the "Gait2354" model included with the software (Fig. 2). The model includes a three-segment model of the exoskeleton which includes two internal revolute joints: the ab/adduction hinge connecting the motor to the waist harness, and the flexion/extension motor angle. The exoskeleton has a six degree-of-freedom joint defined between the pelvis and the waist harness segment. This method is based on a similar approach in [14], which quantified prosthesis socket motion

relative to the residual tibia, but does not restrict the degrees-of-freedom of the device-human interface due to the ability to locate visible marker clusters on all segments. This model can be downloaded at <https://simtk.org/projects/gait-hip-exo>.

Recorded marker positions were low-pass filtered (6 Hz) with a fourth-order zero-lag Butterworth filter using the `filtfilt` function in MATLAB (The Mathworks, Natick, MA) to remove high frequency noise. Quiet standing motion capture data was used to scale the generic model and place virtual markers for each participant. Participant body and exoskeleton kinematics were calculated by using a global least-squares optimization which minimizes weighted model marker position errors relative to experimental marker positions, subject to model joint constraints [15].

G. Dependent Measures

The gait cycle (%) was defined using heel marker position data. 0% of the gait cycle was defined as the peak forward translation of the heel marker of the right-side leg. The body kinematics from each trial were segmented to calculate the following dependent measures for each stride using a custom MATLAB script. The spatial aspects of gait were quantified by the angular range of motion (RoM) of the left and right hip joints, and the step lengths of the left and right legs. The temporal aspect of each stride was characterized by step time for each limb. Step time was defined as the time between heel-strike and the following heel-strike of the opposite limb.

The asymmetry between left and right sides was quantified by calculating a ratio defined as

$$\text{Asymmetry}_\alpha = \frac{\alpha_R - \alpha_L}{\alpha_R + \alpha_L}$$

where α represents the dependent measure for the right (R) and left (L) leg.

H. Statistical Analysis

For each participant, the mean of each dependent measure was calculated during each of the following conditions: the terminal 10 strides in the baseline phase with the stiffness controller off (OFF-Base), the initial (ON-Early), midpoint (ON-Mid), and terminal (ON-Late) 10 strides in the exposure phase with the stiffness controller on, and the initial (OFF-Early) and terminal (OFF-Late) 10 strides in the post-exposure phase with the stiffness controller off. These conditions are summarized in Fig. 1B.

One-way repeated measures analysis of variances (ANOVAs) were conducted to assess the effect of condition (OFF-Base, ON-Early, ON-Mid, ON-Late, OFF-Early, OFF-Late) on each of the dependent measures. The Greenhouse-Geisser correction factor was applied to the within-subject effect of condition. A significant effect of condition was followed up with planned comparisons in the form of pairwise *t*-tests between the following conditions: (1) OFF-Base vs. ON-Early, (2) ON-Early vs. ON-Mid, (3) ON-Mid vs. ON-Late, (4) ON-Late vs. OFF-Early, (5) OFF-Early vs. OFF-Late, and (6) OFF-Base vs. OFF-Late. These planned comparisons were conducted to assess behavioral signatures of neural adaptation [16].

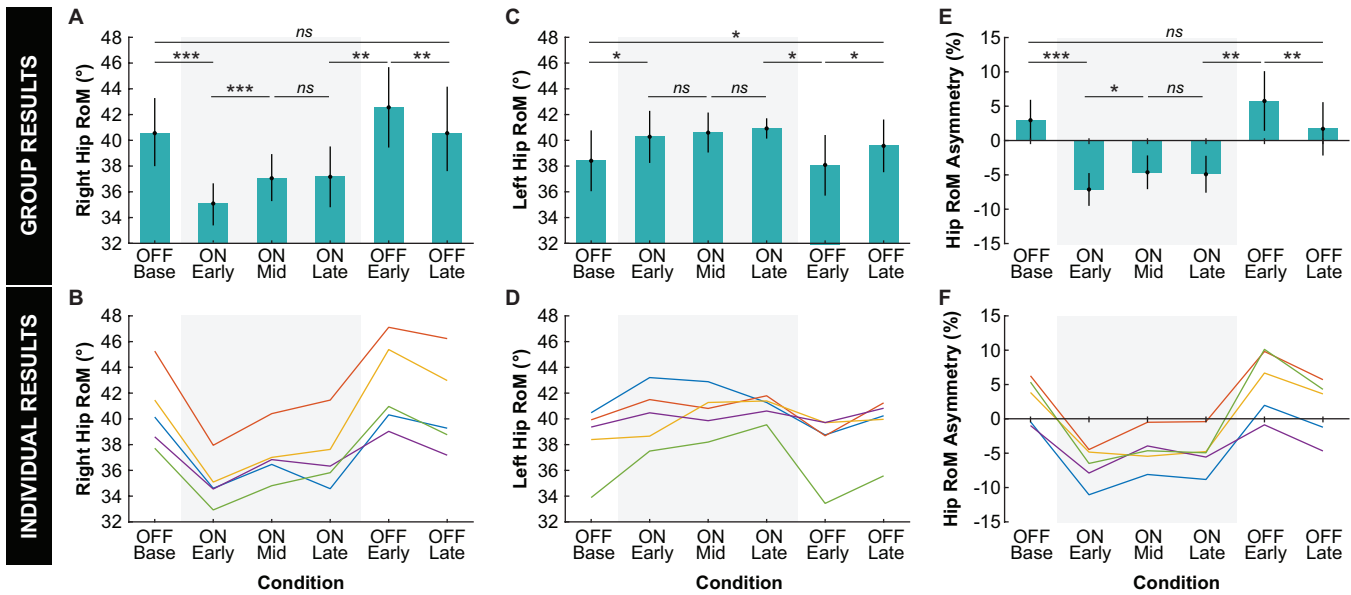


Fig. 4. Hip RoM Results. **A:** Group average and **B:** individual results for Right Hip RoM. **C:** Group average and **D:** individual results for Left Hip RoM. **E:** Group average and **F:** individual results for Hip RoM Asymmetry. **A, C, E:** Error bars indicate one standard deviation. Error bars represent two standard errors of the mean. Shaded regions represent when the stiffness controller was on. The ANOVAs revealed a statistically significant effect of condition on Right Hip RoM, Left Hip RoM, and Hip RoM Asymmetry. *, **, ***, and **** indicate that the planned comparison between conditions was statistically significant with $p < 0.05$, $p < 0.01$, $p < 0.005$, $p < 0.001$, respectively; *ns* indicates no statistical difference. **B, D, F:** Color indicates the different individual subjects.

The statistical analyses were performed using SPSS Statistics for Windows, Version 25.0 (IBM Corporation, Armonk, NY). For all statistical tests, the significance level was set to $\alpha = 0.05$.

III. RESULTS

A. Hip RoM

An ANOVA revealed that the effect of condition on Right Hip RoM, Left Hip RoM, and Hip RoM Asymmetry was statistically significant ($F_{1.56,6.22} = 28.99, p < 0.001$; $F_{1.29,5.15} = 34.34, p = 0.0015$; $F_{2.40,9.59} = 5.41, p = 0.023$, respectively). Planned comparisons found that the effect was typical of that suggesting neural adaptation and are summarized in Fig. 4A, C, E.

Hip RoM Asymmetry was dominated by the behavior of the Right Hip RoM. As shown in the OFF-Base condition, the passive dynamics of the exoskeleton induced asymmetric Hip RoM ($M = 2.8\%$). Turning on the stiffness controller significantly decreased Right Hip RoM and increased Left Hip RoM, such that the asymmetry was in the opposite direction in the OFF-Base condition ($M = -6.9\%$). When the stiffness controller was on, the Right Hip RoM increased to reduce the magnitude of asymmetry at ON-Mid ($M = -4.5\%$) but remained unchanged from ON-Mid to ON-Late.

Turning off the stiffness controller increased Right Hip RoM and decreased Left Hip RoM from ON-Late to OFF-Early, bringing the asymmetry ratio back to positive ($M = 5.5\%$). These measures reverted back to baseline values by OFF-Late, with the exception of Left Hip RoM, which was significantly greater.

B. Step Length

An ANOVA revealed that the effect of condition on Right Step Length and Step Length Asymmetry was statistically significant ($F_{2.13,8.51} = 28.99, p = 0.01$, $F_{2.44,9.75} = 2.78, p = 0.0043$, respectively), but that the effect of condition on Left Step Length was not statistically significant ($F_{1.81,7.23} = 2.78, p = 0.13$). Planned comparisons are summarized in Fig. 5A, C, E).

As with Right Hip RoM, turning on the stiffness controller decreased Right Step Length from OFF-Base to ON-Early, resulting in Step Length Asymmetry decreasing from $M = 2.2\%$ to $M = -0.3\%$. When the stiffness controller was on, Right Step Length increased from ON-Early to ON-Mid, but the change in Step Length asymmetry was not statistically significant. Comparisons between the remaining condition pairs were not statistically significant, although the mean Step Length Asymmetry for each condition increased progressively from ON-Early ($M = -0.3\%$) to OFF-Early ($M = 2.4\%$).

C. Step Time

ANOVAs revealed that the effect of condition was not statistically significant on any of the Step Time dependent measures (Right Leg Step Time: $F_{2.37,9.49} = 1.22, p = 0.35$, Fig. 6A-B; Left Leg Step Time: $F_{2.14,8.57} = 1.94, p = 0.20$, Fig. 6C-D; Step Time Asymmetry: $F_{1.75,7.01} = 1.76, p = 0.24$, Fig. 6E-F).

IV. DISCUSSION

A. Observation of an adaptation response

The first aim of this study was to test if unilateral application of a virtual stiffness using a wearable hip exoskeleton elicits neural adaptation.

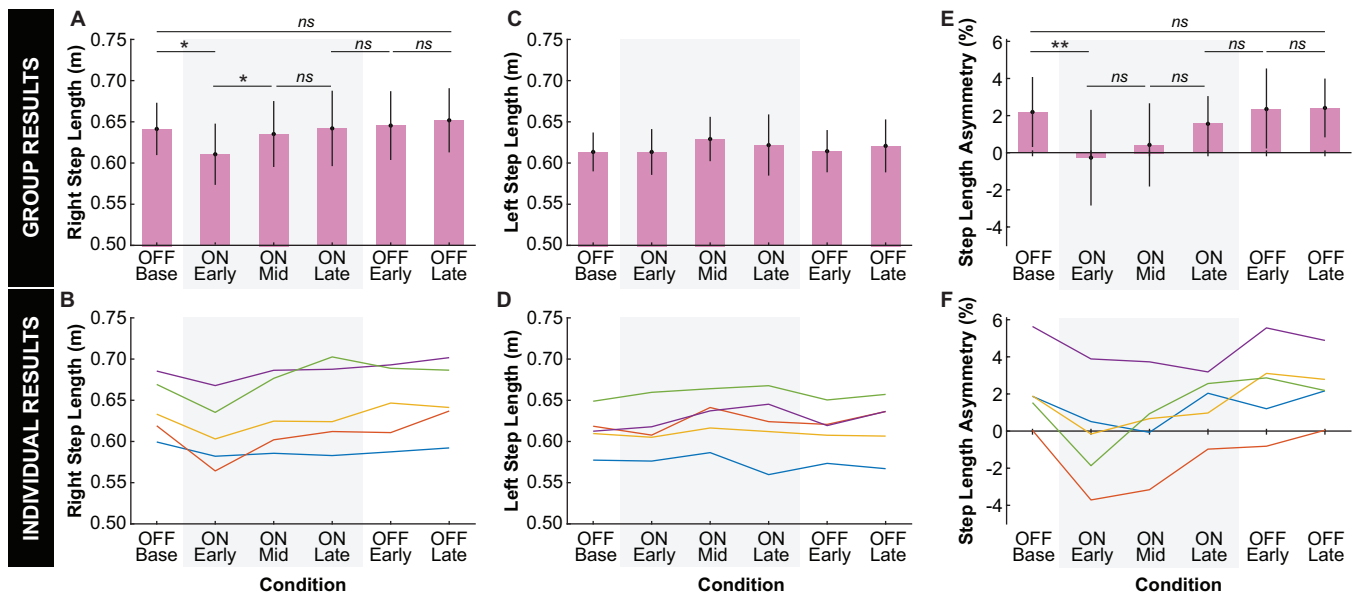


Fig. 5. Step Length Results. **A:** Group average and **B:** individual results for Right Step Length. **C:** Group average and **D:** individual results for Left Step Length. **E:** Group average and **F:** individual results for Step Length Asymmetry. **A, C, E:** Error bars indicate one standard deviation. Error bars represent two standard errors of the mean. Shaded regions represent when the stiffness controller was on. The ANOVAs found a statistically significant effect of condition on Right Step Length and Step Length Asymmetry. The effect of condition on Left Step Time was not statistically significant. Thus, planned comparisons between conditions were only made for Right Step Length and Step Length Asymmetry. **B, D, F:** Color indicates the different individual subjects.

Results showed that all Hip RoM dependent measures exhibit behavioral signatures of neural adaptation. For example, unilateral application of right hip joint stiffness initially decreased Right Hip RoM by $\sim 5^\circ$. Over time, Right Hip RoM gradually increased back towards the value at baseline, but never actually reached baseline value, even after ~ 550 strides (Fig. 4A). Removal of the added hip stiffness resulted in an aftereffect, in which the the Right Hip RoM increased past the baseline value, but eventually settles back down to the baseline value. This trend was observed, albeit in different directions, for left Hip RoM and Hip RoM Asymmetry.

The effect of condition on the Step Length dependent measures was more subtle. Mirroring the effect on Right Hip RoM, the unilateral application of right hip joint stiffness initially decreased Right Step Length, which increased to baseline behavior. However, no aftereffects were observed. Left Step Length remained constant during the application and subsequent removal of unilateral hip stiffness. Thus, Step Length Asymmetry followed the trend of Right Step Length. As will be discussed in Section IV.C, strong evidence of neural adaptation in the RoM may have been masked in the Step Length measures due to the speed constraint imposed by the treadmill.

The application and subsequent removal of unilateral hip stiffness did not significantly affect any of the temporal dependent measures. Mean values changed less than the typical variability observed in normal walking (3% of mean stride duration [17]). This is consistent with previous work investigating bilateral stiffness perturbations [12].

B. Asymmetry persists through all conditions

The second aim of this study was to test if the evidence of neural adaptation could be explained by the nervous system reducing gait asymmetry.

The dependent measures in the OFF-Baseline condition suggest that passive dynamics of the hip exoskeleton induced asymmetric gait (larger hip RoM and longer steps on the side with the exoskeleton) in some participants. The exact cause of this asymmetry is unknown, but there are several possibilities. For instance, these participants could have been compensating for the additional mass of the actuator. Another possibility is that the fit introduced a kinematic constraint between the thigh and the pelvis for these participants.

The spatial measures indicate a preference for asymmetry while wearing the exoskeleton, suggesting that reducing asymmetry was not the main driver of the adaptation observed. Overall, participants appeared to converge to a gait strategy that results in longer steps on the exoskeleton-side, and a reduced exoskeleton-side hip RoM while increased stiffness is applied. This apparent contradiction may be caused by compensation elsewhere in the body, such as rotating the pelvis relative to the exoskeleton waist harness. Regardless, the persistence of asymmetry indicates that the nervous system may not recognize asymmetry as an error that must be corrected, or that a symmetry objective is in competition with other objectives such as energy minimization. Further evidence for this idea has been observed in recent studies which indicate that taking longer steps on the fast belt of a split-belt treadmill is energetically advantageous, and that human participants self-select a gait pattern with positive step-length asymmetry when exposed to the energy cost landscape [18] or allowed to adapt for a sufficient

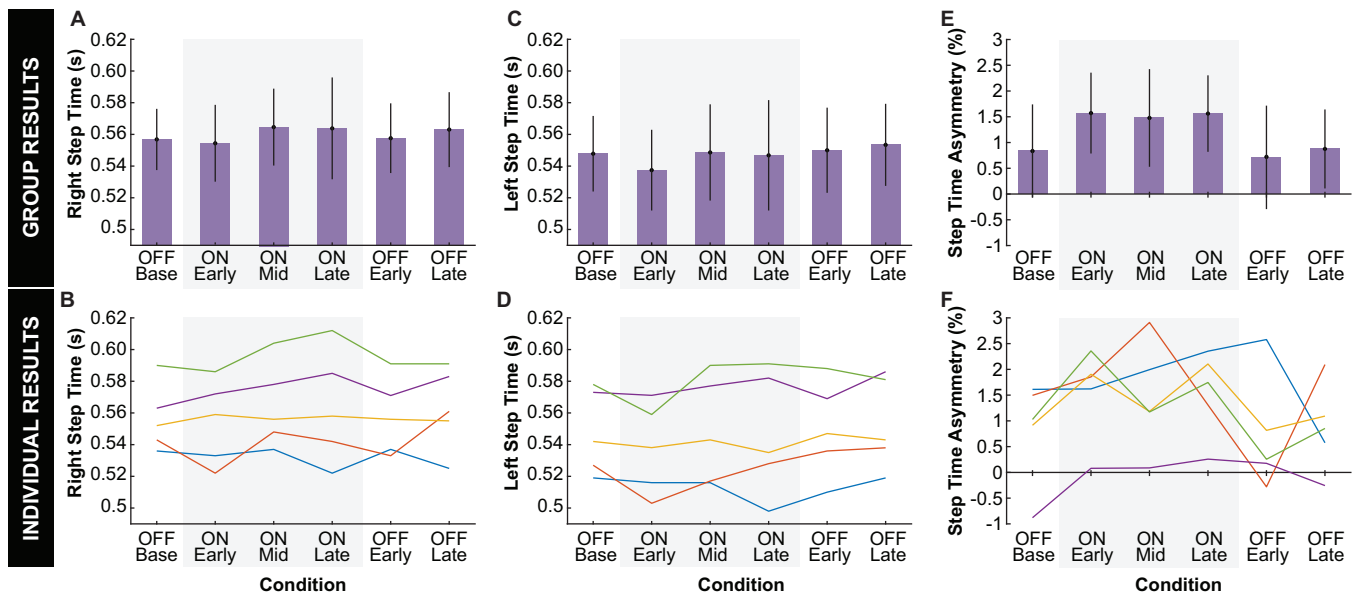


Fig. 6. Step Time Results. **A:** Group average and **B:** individual results for Right Step Time. **C:** Group average and **D:** individual results for Left Step Time. **E:** Group average and **F:** individual results for Step Time Asymmetry. **A, C, E:** Error bars indicate one standard deviation. Error bars represent two standard errors of the mean. Shaded regions represent when the stiffness controller was on. The ANOVAs found no statistically significant effect of condition on Right Step Time, Left Step Time, and Step Time Asymmetry. Thus, no planned comparisons between conditions were made. **B, D, F:** Color indicates the different individual subjects.

interval [19].

C. Limitations

The experiment was conducted on a treadmill, which imposes a speed constraint on the human, allowing for minimal adjustment to their walking speed, thereby coupling their stride length and stride time. It is possible that the opposing changes in hip RoM and step length are compensations induced by the treadmill; participants must preserve a constant stride length if maintaining a constant stride time. This trend may be different overground (i.e. participants may slow down). This limitation will be mitigated with overground trials in a controlled environment in future work.

Like other hip exoskeletons, there was significant motion relative to the body despite deliberate attention spent custom-fitting the exoskeleton to each participant. A comparison of the exoskeleton motor angle (mean RoM during ON-Early: 20°) and the anatomical hip angle (mean RoM during ON-Early: 35°) shows that relative motion introduces an effective dead zone in which torque is not efficiently transmitted across the joint. The mean sagittal range of motion between the waist harness and the pelvis reached up to 11° during the ON conditions as reconstructed by our models. Further work is required to design solutions that reduce relative motion for a wide range of body types.

Finally, model-based reconstruction of kinematics from motion capture is subject to model scaling and marker placement errors. These errors could account for minor joint angle inaccuracies which may be significant given the relatively small changes in RoM observed. However, given that our conclusions are based on trends that are consistent across subjects, and that step length asymmetry is calculated directly from marker coordinates, minor modeling errors should not affect the conclusions of this experiment.

V. CONCLUSIONS

This study investigated the changes in spatio-temporal gait patterns induced by applying unilateral stiffness at the hip joint with the HRSL hip exoskeleton. We found that both stride time and hip joint range of motion were affected by the applied mechanical impedance, which was consistent across different subjects. In contrast to previous work with a bilateral exoskeleton, we observed a time-varying response to exposure to asymmetric hip joint stiffness, indicating that neural adaptation may be occurring. This response appeared to converge to an asymmetric gait pattern, suggesting that asymmetry is not treated as an error overriding other motor control objectives by the nervous system. These results suggest that applying asymmetrical mechanical impedance using lower-limb exoskeleton robots may have potential as a gait neurorehabilitation intervention.

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