ADAPTATION OF WALKING PATTERNS TO BILATERAL LOADED WALKING

Michael Herzog[1], Norman Riedel[2], Melina Beyerlein[1], Jonas Weber[1], Barbara Deml[2], and Thorsten Stein[1]

BioMotion Center, Institute of Sports and Sports Science (IfSS), Karlsruhe Institute of Technology (KIT), Karlsruhe, Germany[1]
Institute of Human and Industrial Engineering (ifab), Karlsruhe Institute of Technology (KIT), Karlsruhe, Germany[2]

Motor adaptation is the process of adjusting motor commands to perform a movement task under altered conditions. The ability and time needed to adapt to an exoskeleton may be crucial to their acceptance and effectiveness. The mechanical changes with exoskeletons are multi-faceted. Therefore, this study focused only on the additional mass and aimed to investigate motor adaptation to bilaterally loaded walking. Six females and six males (24.6±3.8y; 173.3±9.4cm; 66.4±10.1kg) walked with and without weight cuffs attached to their thighs and shanks on a treadmill. Spatio-temporal parameters and lower-limb sagittal angles did not show the hypothesized adaptation progressions and after-effects. Yet, angle-angle diagrams exhibited patterns that may reflect adaptation. Adaptation may occur in the interplay of the lower limb segments, reflecting underlying motor control processes.

KEYWORDS: locomotion, gait, motor learning, exoskeletons, weight cuffs.

INTRODUCTION: Lower-limb exoskeletons are increasingly used in rehabilitation and industry. However, their mechanical properties change the user’s leg mechanics (Qiu et al., 2023) and probably require the user to adapt the motor commands to achieve and maintain stable walking (Poggensee & Collins, 2021). In motor neuroscience, this process is known as motor adaptation. The motor system responds to perturbations to regain a former performance level in the new setting. The adaptation process typically shows a monotonic increase in performance, which is initially rapid and then converges to an asymptote close to the original performance level (exponential course). When the perturbation is removed, participants show “after-effects”, deviations in the opposite direction to those when the perturbation was first introduced (Krakauer et al., 2019). Motor adaptation has been studied extensively in upper limb movements but much less in locomotion (Severini & Zych, 2022). The ability and time needed to adapt walking may be crucial to exoskeleton use, as difficulties or prolonged adaptation time may reduce their acceptance and effectiveness. Considering that older people often have problems maintaining a stable gait (Verghese et al., 2006), the adaptation phase, which comprises the steps after putting on the exoskeleton, and the de-adaptation phase, which comprises the steps after taking off the exoskeleton, may be critical, as gait during these phases likely exhibits instabilities. However, there has been barely any research into motor adaptation to walking with exoskeletons (Poggensee & Collins, 2021), especially as various factors influence the gait pattern with exoskeleton assistance (Qiu et al., 2023). To approach the multi-faceted influences, we simulated the mass of a lower limb exoskeleton under development. This study aimed to investigate motor adaptation to bilaterally loaded walking in young participants to establish baseline information helpful to developing exoskeletons. Based on the adaptation literature (Krakauer et al., 2019), we hypothesize that (H1) people adapt to loaded walking, i.e., reduce the deviation in the parameters’ values induced by the weight cuffs with ongoing strides, and that (H2) people show after-effects when walking after the weight cuffs are taken off. This research is also relevant for training in sports and rehabilitation, as walking with additional loads is a common practice (Washabaugh et al., 2020).

METHODS: Twelve healthy volunteers (six females and six males; 24.6 ± 3.8 years; 173.3 ± 9.4 cm; 66.4 ± 10.1 kg) walked on a treadmill (h/p/cosmos Saturn; Nussdorf-Traunstein, Germany; Figure 1) with and without weight cuffs attached to their thighs and shanks (4 × 2.25

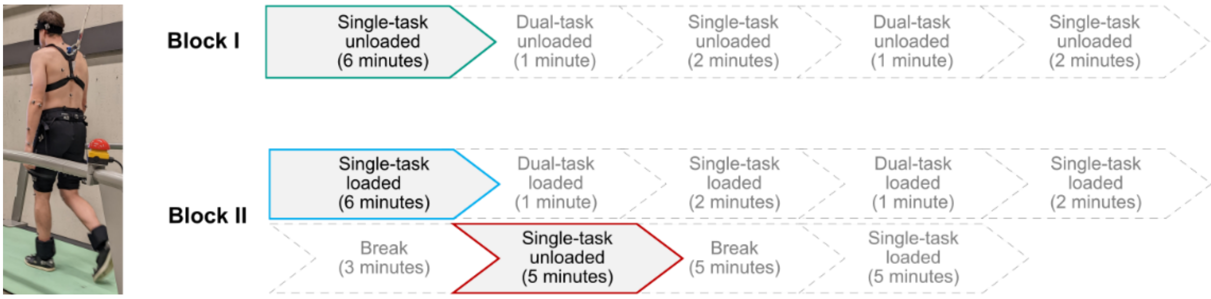

Figure 1: Experimental setup (left) and protocol (right). The analyzed phases are highlighted. The red phase will be referred to as "washout". Figure modified from Riedel et al. (2023).

kg = 9 kg), representing a typical weight of lower-limb, gait-assisting exoskeletons (Bortole et al., 2013). A custom-made hip belt with Velcro straps ensured proper weight cuff positioning. The lower edge of the thigh's weight cuff was fixed 10 cm above the knee joint. Sixteen cameras (200 Hz; Vicon, Oxford, UK) captured full-body movements using a modified Master-Motor-Map marker setup with 56 markers (Mandery et al., 2016). The participants walked without weight cuffs on the treadmill for six minutes to familiarize themselves; then, their individual preferred gait speed was measured (4.0 ± 0.3 m/s). The order of blocks I and II, as shown in Figure 1, was counterbalanced across the participants. The marker trajectories were post-processed and filtered (6 Hz low-pass, 4^{th}-order Butterworth) with VICON Nexus (v2.14.0, Vicon, Oxford, UK) and MATLAB (R2023a, The MathWorks, Natick, MA, USA). Contrary to well-known adaptation paradigms, a metric reflecting motor adaptation to bilateral loaded walking has not yet been described. Therefore, step width, stride length, stance phase, stride time, and cadence were evaluated. Sagittal hip, knee, and ankle angles were derived with OpenSim's Inverse Kinematics Tool (OpenSim v4.4) using the model by Arnold et al. (2010). For gait coordination analysis (Wheat & Glazier, 2006), hip-knee and knee-ankle diagrams ("coupling angles") were calculated. To test differences between unloaded walking (five strides in the middle of the unloaded phase) and the start and end of loaded walking (first five and last five strides; H1), repeated-measures ANOVA and *post-hoc* Bonferroni-corrected dependent t-tests were used ($p \approx 0.016$). Dependent t-tests were used to test for differences between unloaded walking and the start of the washout (first five strides; H2). The significance level was set a priori at two-sided $p = 0.05$.

RESULTS: The parameters deviated partially from unassisted to the start of loaded walking (Table 1: $p_{U\leftrightarrow S}$) and did not exponentially converge back toward the end of loaded walking at the same time (H1 rejected; Table 1: $p_{U\leftrightarrow E}$, $p_{S\leftrightarrow E}$; Figure 2A). Furthermore, the parameters did not show after-effects, i.e., mirror-inverted values during washout (H2 rejected; Table 1: $p_{U\leftrightarrow W}$). However, the coupling angles showed changes, especially during the swing phase (Figure 2B), in line with H1 (indicated by arrows).

DISCUSSION: This study analyzed motor adaptation to bilaterally loaded walking on a treadmill with spatio-temporal parameters and sagittal lower-limb joint angles. Contrary to our hypotheses and the literature (Krakauer et al., 2019), people demonstrated neither a classical exponential adaptation progression during loaded walking (H1) nor after-effects (H2) based on the evaluated parameters. Yet, the coupling angles revealed patterns, especially regarding the change from unloaded to loaded and during loaded walking, possibly indicating underlying adaptation processes. In general, adding masses leads to kinematic differences at the ankle and knee during early swing and at the hip throughout the gait cycle (Fang et al., 2022). We can now add that these differences change over time during loaded walking and, thus, may reflect adaptation. Possibly, adaptation occurring in the coordination of the lower limb segments is not reflected by the spatio-temporal parameters. Instead, people might be able to switch their walking patterns immediately to achieve the task goal - they can immediately walk at the imposed speed with weight cuffs attached. But still, adaptation may happen in the underlying control processes, e.g., optimizing the interplay of the lower limb segments to minimize energetic cost successively (Selinger et al., 2015). A possible contrasting explanation

Table 1: Statistical results to evaluate adaptation during loaded walking (H1, top) and after-effects (H2, bottom). Statistically significant differences are in bold numbers.

Adaptation	Unloaded	Start loaded	End loaded	p_{ANOVA}	$p_{U\leftrightarrow S}$	$p_{U\leftrightarrow E}$	$p_{S\leftrightarrow E}$
Stride time [s]	1.16 ± 0.07	1.19 ± 0.08	1.21 ± 0.08	**<0.01**	0.035	**<0.01**	0.106
Stride length [m]	1.30 ± 0.09	1.34 ± 0.07	1.36 ± 0.08	**<0.01**	0.035	**<0.01**	0.091
Stance phase [%]	66.47 ± 0.83	64.68 ± 1.10	64.98 ± 0.83	**<0.01**	**<0.01**	**<0.01**	0.125
Step width [m]	0.08 ± 0.03	0.11 ± 0.03	0.10 ± 0.03	**<0.01**	**<0.01**	**<0.01**	0.142
Cadence [steps/min]	104.15 ± 5.85	101.32 ± 6.97	99.58 ± 6.55	**<0.01**	0.039	**<0.01**	0.089
Sag. hip RoM [°]	41.44 ± 4.87	42.22 ± 4.84	43.64 ± 5.99	0.132			
Sag. knee RoM [°]	64.13 ± 4.97	57.36 ± 5.30	61.40 ± 5.40	**<0.01**	**<0.01**	**<0.01**	**<0.01**
Sag. ankle RoM [°]	29.07 ± 6.73	24.16 ± 4.22	26.99 ± 5.33	**<0.01**	**<0.01**	0.035	**<0.01**

Washout	Unloaded	Washout	$p_{U\leftrightarrow W}$
Stride time [s]	1.16 ± 0.07	1.15 ± 0.07	0.473
Stride length [m]	1.30 ± 0.09	1.29 ± 0.08	0.211
Stance phase [%]	66.47 ± 0.83	66.37 ± 1.10	0.470
Step width [m]	0.08 ± 0.03	0.07 ± 0.03	0.224
Cadence [steps/min]	104.15 ± 5.85	104.56 ± 6.72	0.508
Sag. hip RoM [°]	41.44 ± 4.87	42.17 ± 3.90	0.255
Sag. knee RoM [°]	64.13 ± 4.97	63.69 ± 4.76	0.534
Sag. ankle RoM [°]	29.07 ± 6.73	27.20 ± 5.98	**0.041**

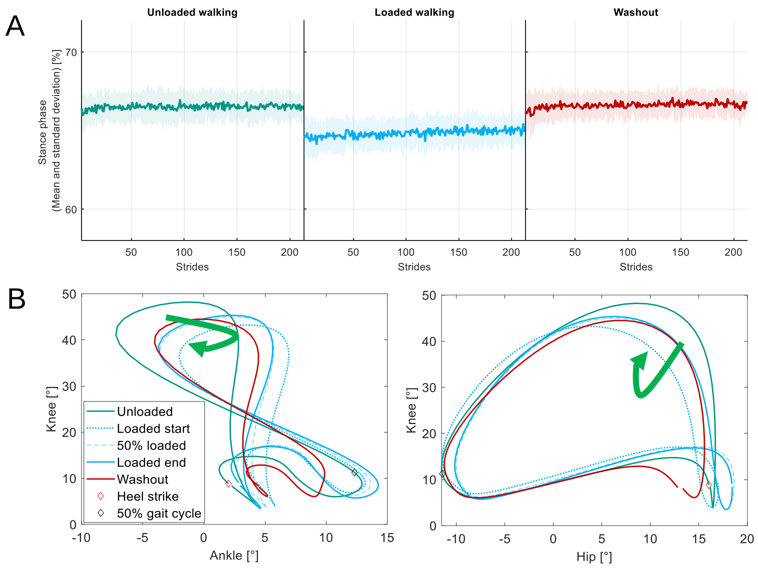

Figure 2: A: Change of stance phase duration over time in the three phases as a representative of the spatio-temporal parameters. B: Coupling angles of hip, knee, and ankle sagittal angles. The green arrows illustrate the change (first inward at the start of loaded walking and then back toward unloaded walking with increasing strides).

for why typical adaptation patterns are not displayed might be that walking with additional loads is a frequent task, e.g., wearing hiking boots. They might, thus, have formed a permanent walking pattern approximating the new movement (our setup) and be able to switch their gait pattern rather than needing to adapt over multiple strides (Reisman et al., 2010). Several limitations may have influenced the obtained results. Walking speed was fixed, so participants could not alter it, which could have been a possible adaptation mechanism. Yet, Fang et al. (2022) observed that neither thigh nor shank loading changed the walking speed significantly. As coupling angles illustrate the coordination between lower limb joints and exhibit a pattern that might reflect motor adaptation, a future investigation could investigate possible adaptations at the joint angle level more holistically. We would not expect the CoM movement to change significantly, but the level below (e.g., joint angles) generating the CoM movement (Möhler et al., 2021). Also, exoskeletons not only alter the limbs' masses but often have a mismatch of the human–robot joint rotation centers and provide support (Qiu et al., 2023). These additional factors can be addressed in the future to more thoroughly assess possible adaptation to the multi-faceted changes induced by an exoskeleton.

CONCLUSION: Motor adaptation is a process crucial for the performance and acceptance of exoskeletons, which may be used in training or rehabilitation. The first results based on spatio-temporal and joint-kinematic variables suggest that young participants do not exhibit an exponential stride-by-stride adaptation pattern. Yet, angle-angle diagrams provide a starting point for gait coordination analysis, possibly revealing underlying motor adaptation processes. The biomechanical changes induced by exoskeletons are complex. Therefore, it is necessary to disentangle various aspects, such as the additional loading. In the future, the methodology presented here may help to investigate motor adaptation to walking with actuated, i.e., actively gait-supporting exoskeletons.

REFERENCES

Arnold, E. M., Ward, S. R., Lieber, R. L., & Delp, S. L. (2010). A Model of the Lower Limb for Analysis of Human Movement. *Annals of Biomedical Engineering*, *38*(2), 269–279. https://doi.org/10.1007/s10439-009-9852-5

Bortole, M., del Ama, A., Rocon, E., Moreno, J. C., Brunetti, F., & Pons, J. L. (2013). A robotic exoskeleton for overground gait rehabilitation. *2013 IEEE International Conference on Robotics and Automation*, 3356–3361. https://doi.org/10.1109/ICRA.2013.6631045

Fang, S., Vijayan, V., Reissman, M. E., Kinney, A. L., & Reissman, T. (2022). How Do Joint Kinematics and Kinetics Change When Walking Overground with Added Mass on the Lower Body? *Sensors*, *22*(23). Scopus. https://doi.org/10.3390/s22239177

Krakauer, J. W., Hadjiosif, A. M., Xu, J., Wong, A. L., & Haith, A. M. (2019). Motor Learning. *Comprehensive Physiology*, *9*(2), 613–663. https://doi.org/10.1002/cphy

Mandery, C., Terlemez, O., Do, M., Vahrenkamp, N., & Asfour, T. (2016). Unifying Representations and Large-Scale Whole-Body Motion Databases for Studying Human Motion. *IEEE Transactions on Robotics*, *32*(4), 796–809. https://doi.org/10.1109/TRO.2016.2572685

Möhler, F., Stetter, B., Müller, H., & Stein, T. (2021). Stride-to-Stride Variability of the Center of Mass in Male Trained Runners After an Exhaustive Run: A Three Dimensional Movement Variability Analysis With a Subject-Specific Anthropometric Model. *Frontiers in Sports and Active Living*, *3*. https://doi.org/10.3389/fspor.2021.665500

Poggensee, K. L., & Collins, S. H. (2021). How adaptation, training, and customization contribute to benefits from exoskeleton assistance. *Science Robotics*, *6*(58), eabf1078. https://doi.org/10.1126/scirobotics.abf1078

Qiu, S., Pei, Z., Wang, C., & Tang, Z. (2023). Systematic Review on Wearable Lower Extremity Robotic Exoskeletons for Assisted Locomotion. *Journal of Bionic Engineering*, *20*(2), 436–469. https://doi.org/10.1007/s42235-022-00289-8

Reisman, D. S., Bastian, A. J., & Morton, S. M. (2010). Neurophysiologic and Rehabilitation Insights From the Split-Belt and Other Locomotor Adaptation Paradigms. *Physical Therapy*, *90*(2), 187–195. https://doi.org/10.2522/ptj.20090073

Riedel, N., Herzog, M., Stein, T., & Deml, B. (2023). Effects of Modified Leg Mechanics on Cognitive Performance and Workload during Dual-Task Walking. In D. de Waard, V. Hagemann, L. Onnasch, A. Toffetti, D. Coelho, A. Botzer, M. de Angelis, K. Brookhuis, & S. Fairclough, *Proceedings of the Human Factors and Ergonomics Society Europe Chapter 2023 Annual Conference*, 197-211. ISSN 2333-4959.

Selinger, J. C., O'Connor, S. M., Wong, J. D., & Donelan, J. M. (2015). Humans Can Continuously Optimize Energetic Cost during Walking. *Current Biology*, *25*(18), 2452–2456. https://doi.org/10.1016/j.cub.2015.08.016

Severini, G., & Zych, M. (2022). Locomotor adaptations: Paradigms, principles and perspectives. *Progress in Biomedical Engineering*, *4*(4), 042003. https://doi.org/10.1088/2516-1091/ac91b6

Verghese, J., LeValley, A., Hall, C. B., Katz, M. J., Ambrose, A. F., & Lipton, R. B. (2006). Epidemiology of gait disorders in community-residing older adults. *Journal of the American Geriatrics Society*, *54*(2), 255–261. https://doi.org/10.1111/j.1532-5415.2005.00580.x

Washabaugh, E. P., Augenstein, T. E., & Krishnan, C. (2020). Functional resistance training during walking: Mode of application differentially affects gait biomechanics and muscle activation patterns. *Gait & Posture*, *75*, 129–136. https://doi.org/10.1016/j.gaitpost.2019.10.024

Wheat, J. S., & Glazier, P. S. (2006). Measuring Coordination and Variability in Coordination. In K. Davids, S. Bennett, & K. Newell (Hrsg.), *Movement System Variability* (1. Aufl., S. 167–182). Human Kinetics. https://doi.org/10.5040/9781492596851

ACKNOWLEDGEMENTS: This research was supported by the JuBot project, funded by the Carl-Zeiss-Foundation.