Modulating PD-controller parameters allowing for differences in trajectory tracking error: LOKOMAT rehabilitation system.

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***Abstract*—The Lokomat gait rehabilitation system is a robotic system for gait training of patients after Spinal Cord Injury or stroke. The control settings of this gait rehabilitation system can allow either for strict tracking with small error-allowance or for more lenient tracking with increased path variability. In this paper, it is explored what the difference in trajectory tracking error is for different parameters of a Proportional-Derivative (PD) controller. The explored range of parameters is within those that achieve an adequate step-response, defined as less than 10% offset or overshoot. A PD-controller for a Simulink model of the LOKOMAT system is developed in Matlab, and the kp and kd values that generate a step response within 10% overshoot or offset are then tested in tracking an exemplary gait trajectory. The Root Mean Square Error (RMSE) for the trajectory tracking is determined per combination of kp and kd. The obtained maximum difference in RMSE within the explored range of PD-parameters is 0.321 degrees for knee flexion, 0.109 degrees for hip flexion and 0.0626 degrees for hip abduction. These findings indicate a difference between a more strict or more lenient tracking mode, however small. The difference should be increased by wider exploration of PD-parameters or through the addition of noise, after which the different PD-parameter settings can be used in further research on achieving path variability in gait rehabilitation robotics.**

I. Introduction

A. LOKOMAT gait rehabilitation system

Body weight supported treadmill training is a widely accepted treatment option to help patients with impaired gait due to Spinal Chord Injury (SCI) or stroke, retain or improve gait [1], [2]. Robotic devices aiding the legs during movement allow for less labor-intensive and thus cost-effective training without constant assistance of a physician [3].

One of these robotic gait rehabilitation systems is the LOKOMAT system [4]. This system used to allow for walking as a planar motion by facilitating knee and hip flexion or extension, just as most traditional robotic gait systems [5]. Recently, an addition to this system has been made by adding an extra Degree of Freedom (DOF) in the form of ad- and abduction of the hip joint, to achieve a more natural gait pattern [6]. Ad- and abduction of the hip-joint is used in natural gait for maintaining balance on uneven surface. With reduced balance of post-stroke and SCI patients being a cause for increased fall risk even after rehabilitation [7], taking this joint DOF into consideration during training can be of clinical relevance.

The hip joint in the improved LOKOMAT gait rehabilitation robot is driven by two linear drivers (one internal and one external) that will cause hip flexion when powered synchronously and can also cause ad- or abduction of the hip when powered non synchronously. Figure 1 shows a schematic of this mechanism. This added indirect control of the hip DOFs allows for more naturalistic gait, but also complicates the drive and control scheme of the Lokomat system.

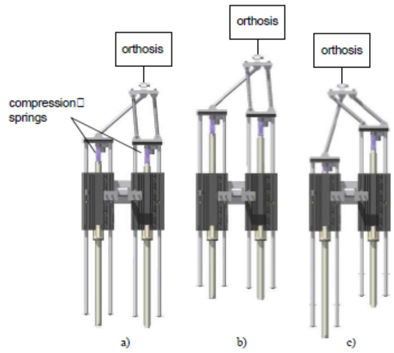


Fig. 1: The hip-actuators in the renewed LOKOMAT system. Taken from [6]. **(a)** original position, **(b)** synchronous activation for hip flexion, **(c)** asynchronous activation for hip ab- or adduction.

B. Control in Rehabilitation Robotics

An important part of designing rehabilitation robotics, is the control scheme. The conventional and first gait training orthoses use Proportional-Derivative (PD) controlled trajectory tracking [5]. With this type of control, a set trajectory is determined for the motion. An error between the current and desired position and the derivative of that error, are used to calculate the necessary joint torques to keep the (robotic) limb on track.

One common way of tuning a PD-controller is to start with a step-response [8]. When taking a step amplitude of the maximum expected instantaneous change in joint angle in one time instant, one can see if the robotic system can follow this step adequately. That is: not too much offset or overshoot. If the step response has a too large offset, tracking of most trajectories will be inadequate. If the step response contains too large overshoots, sudden changes in trajectories can cause uncomfortable or dangerous situations when muscles or joints get overstretched. In this paper, an adequate step response is defined as an overshoot or offset of less than 10%, chosen based on observations of this value in literature as upper bound for PD-tuning [9].

Often an adequate step response is achieved with a large range of PD-parameters. Within this range, PD-controller gains can be modulated to either increase or decrease tracking error of more dynamic trajectories such as those of joint angles in human gait. Increasing PD-controller gains can decrease tracking error and path variability by more strict tracking. However, a too high proportional gain values without increasing the derivative gain can lead to dangerous overshoots in the step response. And increasing both the proportional and derivative gains is limited by the actuators' maximal force. Decreasing PD-controller gains could achieve higher path variability by allowing some error in the path tracking. At the same time, too low PD-gains can also result in inadequate tracking of the step response, making it harder to ensure adequate tracking of any trajectory.

PD-controllers for rehabilitation robotics are often optimized for perfecting tracking of trajectories, by using high PD-controller gains [10]–[12]. Current state of the art systems obtain low tracking error, with average or maximum errors reported between 0.1 and 0.5 degrees [12]–[14]. However, this precise tracking also has its downsides. It could give too much guidance, decreasing the motor learning of the patient due to decreased effort [15]. Allowing more tracking errors will cause more variability of the motion, which is suggested to induce increased patient effort and engagement [16].

C. Research goal

Ideally, multiple variations in tracking error and the resulting patient engagement can be explored and researched for more optimal usage of the rehabilitation device. Therefore it is relevant to have an overview of what PD-parameters make for a decent step response and what the possible differences in trajectory tracking errors are for those PD-parameters, for any newly implemented control system.

To obtain this overview for the LOKOMAT system, the following question will be answered in this research paper: 'What is the maximum difference in trajectory tracking error is that can be observed for different PD-control parameters for the LOKOMAT system, within the range of parameters that ensure an adequate step response?' As mentioned an adequate step response is defined as one within 10% overshoot or offset. The error between desired and actual trajectory will be expressed with the Root Mean Square Error (RMSE).

II. Methods

A. Model design

For implementation and evaluation of the PD-controller, a model of the Lokomat system was used. This model was previously created in Simulink by Manzari [17] and used with consent of the creators. The model consists of two main components. Firstly the Forward Dynamics from rotary knee actuator torque or linear hip actuator force to the angle or position of these actuators repectively. Secondly he kinematics, both forward and inverse, between the internal/external linear actuator positions and the ab-/adduction and flexion angles of the hip joint.

B. Controller design

For the control of the Lokomat, a PD-controller was developed using Matlab and Simulink software. The PD-controller adheres to the standard PD-control rule, as depicted in equation 1.

$$u(t) = kp * e(t) + kd * \frac{\mathrm{d}e(t)}{\mathrm{d}t} \tag{1}$$

For the directly actuated knee joint, $e(t)$ (rad) is the difference in desired and actual knee angle. For the indirectly actuated hip joint, $e(t)$ (m) follows from the inverse kinematic relations between actuator position and hip angles. Variable $u(t)$ is the actuator output in either $\mathrm{N\,m}$ or N. The variable kp is the proportional gain in $\frac{\mathrm{N\cdot m}}{\mathrm{rad}}$ or $\frac{\mathrm{N}}{\mathrm{m}}$ and kd is the derivative gain in $\frac{\mathrm{N\cdot m}}{\mathrm{rad\cdot s^{-1}}}$ or $\frac{\mathrm{N}}{\mathrm{m\cdot s^{-1}}}$. The closed loop control scheme of the Lokomat system with the PD-controller is given in figure 2.

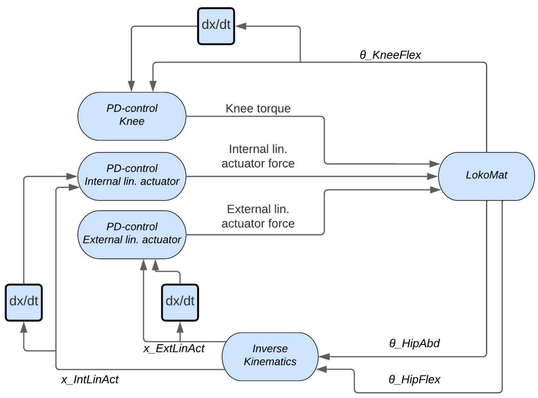


Fig. 2: Closed loop control scheme of the PD-controller and Lokomat system

C. Determining the step response

The tested step input had an amplitude of 1 degree for the knee and 0.5 degrees for both hip abduction and flexion. For every tested kd, kp was increased until an overshoot of more than 10% was reached.

The values of kd and kp need to be limited in the model, since infinite increase of these gains in reality leads to force actuator saturation. For the knee joint actuator, literature on comparable actuators and their kd-values [10] suggested a max. kd of 10 $\frac{\mathrm{N\cdot m}}{\mathrm{rad\cdot s^{-1}}}$. The range of 0 to 10 was split into 20 steps of 0.5 $\frac{\mathrm{N\cdot m}}{\mathrm{rad\cdot s^{-1}}}$. At every kd, kp was increased with steps of 20 $\frac{\mathrm{N\cdot m}}{\mathrm{rad}}$, to judge the resulting step response.

The step input of the linear actuator controller in meters was around a factor 200 smaller than the knee controller input in radians. Therefore the kd range (0 to 2000 $\frac{\mathrm{N}}{\mathrm{m\cdot s^{-1}}}$) and the kd and kp step sizes (100 $\frac{\mathrm{N}}{\mathrm{m\cdot s^{-1}}}$ and 4000 $\frac{\mathrm{N}}{\mathrm{m}}$ resp.) were 200 times higher to achieve similar tuning.

These are large assumptions regarding the range of parameters that will result in physically feasible forces. To validate these assumptions, the maximum actuator force (N) / torque ($\mathrm{N\,m}$) at the point of maximum controller gains within this range will be compared to the physical maximum actuator force. For the physical knee actuator this is $500\,\mathrm{N\,m}$ and for the linear hip actuators this is $2000\,\mathrm{N}$.

D. Determining tracking error

Once the parameter region for adequate step response was determined, the difference in trajectory tracking within this parameter region was judged. For every combination of kd and kp, a predefined exemplary trajectory of joint angles created by Manzari [17] was tracked using the PD-controller. The Root Mean Square Error (RMSE) in degrees between the desired and actual trajectory was taken as a measure for tracking error, as is often done in comparable literature [14], [18] and thus leads to possible comparison with other state of the art research to confirm plausibility of the results.

III. Results

A. Step Response

For the step response testing, it was generally observed that too low kp resulted in offsets larger than 10%, while increasing the kp decreased the final offset, but eventually increased overshoot. At larger kd, the kp could be increased further before 10% overshoot was reached. Figure 3 shows the different kp for the PD-controller that resulted in adequate tracking of the step response (within 10% overshoot or offset), for the predetermined range of kd. Figure 3 A shows the knee joint actuator with the kd-range of 0 to 10 $\frac{\mathrm{N \cdot m}}{\mathrm{rad \cdot s^{-1}}}$ and Figure 3 B shows the internal and external linear actuator with the kd-range of 0 to 2000 $\frac{\mathrm{N}}{\mathrm{m \cdot s^{-1}}}$. The maximum torque at the highest kp- and kd-value for the knee joint actuator was 203.56 N m. The maximum forces at the highest kp- and kd-value were 2733.7 N and 9609.8 N for the internal and external actuator, respectively.

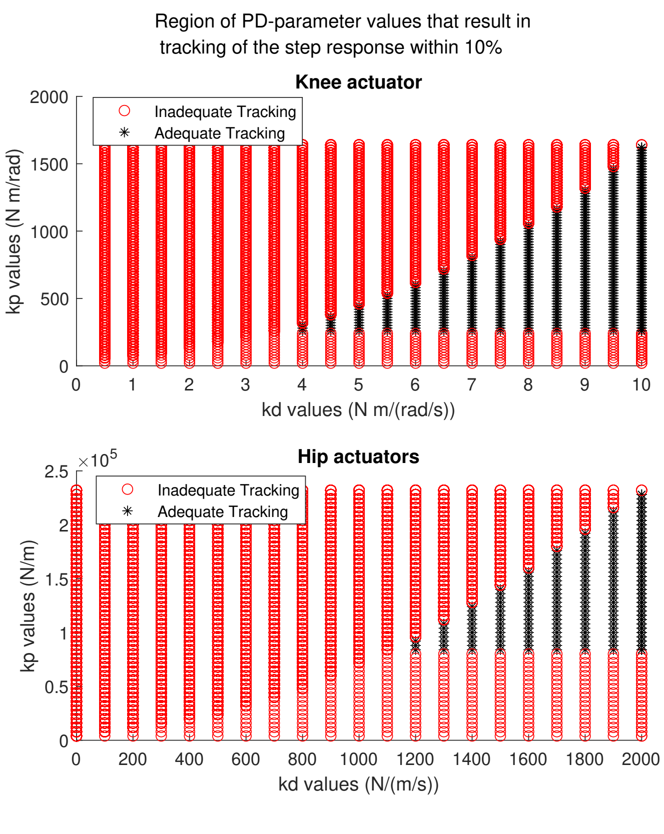


Fig. 3: PD-parameters that ensure a step response within 10% offset or overshoot, for (a) the knee and (b) the hip actuators.

B. Trajectory tracking

For trajectory tracking, it was generally observed that larger kp-values, allowed by also larger kd-values, resulted in a smaller RMSE. Figure 4 shows the RMSE for the three different joint angles at the different PD-controller parameters. (A) shows the RMSE for the knee flexion angle at different kp and kd of the knee actuator controller. (B) and (C) show the RMSE for the hip flexion and hip ab/adduction respectively, for the set of kp and kd values of the controller for both the internal and external linear actuator. Table I shows the maximum and minimum RMSE for all three angles, the corresponding kp and kd values and the maximum achieved force/torque of the responsible actuators.

TABLE I: Max. and min. found RMSE, with corresponding kp/kd values and maximum instantaneous force (F) / torque (T).

Joint Angle	Maximum RMSE		
	RMSE (deg)	PD-par. (kp/kd)	Max. F / T
Knee flex.	0.550	260 $\frac{\mathrm{N \cdot m}}{\mathrm{rad}}$ / 4 $\frac{\mathrm{N \cdot m}}{\mathrm{rad \cdot s^{-1}}}$	7.9 N m
Hip flex.	0.172	84000 $\frac{\mathrm{N}}{\mathrm{m}}$/ 1300 $\frac{\mathrm{N}}{\mathrm{m \cdot s^{-1}}}$	1340 N
Hip abd.	0.0974	84000 $\frac{\mathrm{N}}{\mathrm{m}}$ / 1300 $\frac{\mathrm{N}}{\mathrm{m \cdot s^{-1}}}$	1340 N
Joint Angle	Minimum RMSE		
	RMSE (deg)	PD-par. (kp/kd)	Max. F / T
Knee flex.	0.229	1620 $\frac{\mathrm{N \cdot m}}{\mathrm{rad}}$ /10 $\frac{\mathrm{N \cdot m}}{\mathrm{rad \cdot s^{-1}}}$	27.8 N m
Hip flex.	0.0630	228000 $\frac{\mathrm{N}}{\mathrm{m}}$/2000 $\frac{\mathrm{N}}{\mathrm{m \cdot s^{-1}}}$	2177 N
Hip abd.	0.0348	228000 $\frac{\mathrm{N}}{\mathrm{m}}$/2000 $\frac{\mathrm{N}}{\mathrm{m \cdot s^{-1}}}$	2177 N

The difference between the maximum RMSE and minimum RMSE within the explored set of PD-parameters as determined in secion III.A, was 0.321 degrees for knee flexion, 0.109 degrees for hip flexion and 0.0626 degrees for hip ab-/adduction.

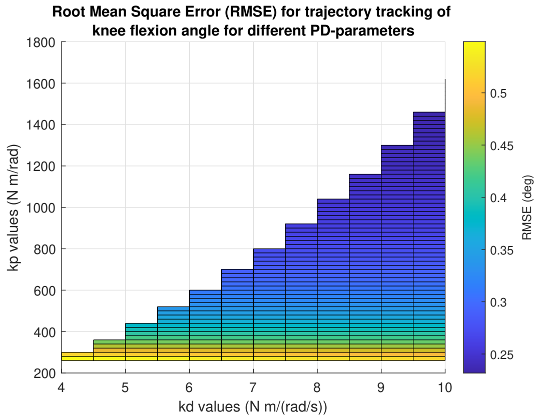
IV. Discussion

A. Obtained results

The research question of this report was what the maximum difference in RMSE was that could be achieved by alternating PD-control parameters within a range of parameters that achieve an adequate step-response. The results show that this difference is different per tracked joint angle, with 0.320, 0.109 and 0.0626 degrees for knee flexion, hip flexion and hip abduction respectively.

The results show two types of parameter settings, illustrating the difference in operating modes for the PD-controller of the LOKOMAT system. A choice can be made either for a very precise tracker with high kp- and kd-values to decrease the RMSE in trajectory tracking, but with increasing overshoot in the step response. To increase kp further without increasing overshoot, the damping value kd needs to be increased as well, up until the point where actuator torque/force limits are reached. On the other hand, a more lenient PD-controller can be used, with low kp-values allowing for a larger RMSE, but eventually leading to large offsets in the step response.

However, the results also show that within the explored range of parameters, the difference between these two modes when expressed by the RMSE is only minimal. The primary



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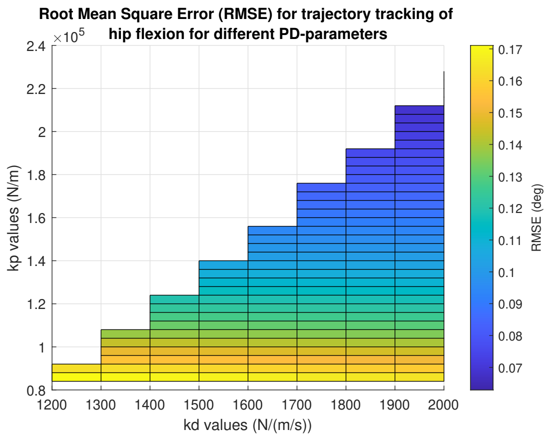

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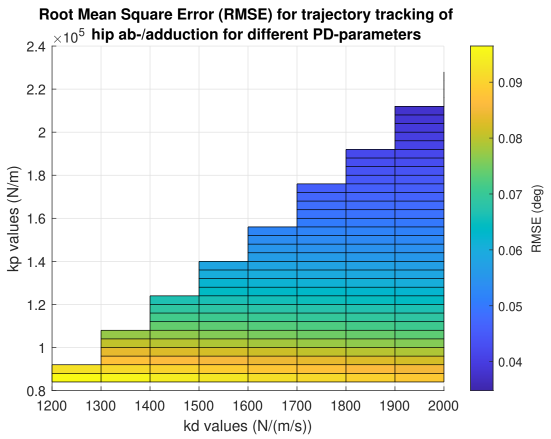

C

Fig. 4: Root Mean Square Error for the examplary trajectory tracking for the knee flexion angle (A), the hip flexion angle (B) and the hip ab-/adduction angle (C).

rationale behind the small absolute difference, is the overall small RMSE. As in other research [12], [13], the RMSE found in this paper stays below 1 degree and thus high percentual increases yield minimal absolute change. For a clinically relevant difference in trajectory tracking error of the different operating modes, it seems that more parameter or controller options need to be explored. For this, multiple options can be considered in line with other performed research on this topic.

B. Further exploration

First, it can be observed that trying to limit the explored kd and kp values to those that were expected to not lead to nonphysical actuator forces/torques was not successful. The maximum force of the knee actuator is never reached, while for the hip actuators it is largely exceeded at some points. Further research should limit the actuator force by programming a saturation on the actuator and not limiting the kd and kp values explored, so that a larger difference in RMSE can be found.

Another aspect that was not explored in this paper is the possibility of adding an integral gain ki to the controller. This could decrease the steady state offset and allow for lower kp- and kd-values giving an adequate step response, with higher RMSE in trajectory tracking. In an article by Shen et al. it can be observed that with added integral gain, much lower kp than in this research are used and an RMSE larger than 1 degree is observed [14]. However, no comparison is explicitly mentioned of the controller before and after the addition of ki and it remains a topic for further research.

Lastly, the addition of random noise on the force output of the actuators could increase the RMSE, especially for the lower kp- and kd-values and thus make the effect of the different parameter settings more pronounced. This was explored in a research article by Xu et a. [13] and resulted in a doubled error in trajectory tracking.

C. Clinical Implications

The used controller mode will have clinical implications. A higher RMSE in trajectory tracking illustrates more room for path variability and could thus increase patient engagement. However, in very early stages of rehabilitation patients could perhaps be most aided by exactly following the correct path, before gaining more freedom in the exploration of different trajectories. This paper contributes to exploring the differences in trajectory tracking lenience of different PD-control parameters, to aid further investigation of the clinical effects of these different control settings. Further research on the resulting path variability and clinical effects thereof when these PD-controller settings are implemented in the LOKOMAT system, will be a useful step forward in optimizing control settings for gait rehabilitation using this device.

V. Conclusion

The found difference in RMSE for the explored PD-parameters in this paper is present but small. This difference could be further increased with noise models or wider parameter exploration, before using the different parameter settings in further research on the clinical effects of strict or lenient trajectory tracking with a PD-controller for the LOKOMAT system. This research could be used for creating rehabilitation environments tuned to the needs of the patient, improving rehabilitation healthcare.

VI. Data Availibility Section

"The data analysis scripts/code that are used in this study are available in the corresponding GitHub repository [19].

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REFERENCES

[1] J. Mehrholz, M. Pohl, J. Kugler, and B. Elsner, "The improvement of walking ability following stroke: a systematic review and network meta-analysis of randomized controlled trials," *Deutsches Ärzteblatt International*, vol. 115, no. 39, p. 639, 2018.

[2] R. W. Bohannon, A. W. Andrews, and M. B. Smith, "Rehabilitation goals of patients with hemiplegia," *International Journal of Rehabilitation Research*, vol. 11, no. 2, pp. 181–184, 1988.

[3] S. Hussain, S. Q. Xie, and G. Liu, "Robot assisted treadmill training: mechanisms and training strategies," *Medical engineering & physics*, vol. 33, no. 5, pp. 527–533, 2011.

[4] R. Riener, L. Lünenburger, I. C. Maier, G. Colombo, and V. Dietz, "Locomotor training in subjects with sensori-motor deficits: an overview of the robotic gait orthosis lokomat," *Journal of Healthcare Engineering*, vol. 1, no. 2, pp. 197–216, 2010.

[5] P. K. Jamwal, S. Hussain, and M. H. Ghayesh, "Robotic orthoses for gait rehabilitation: An overview of mechanical design and control strategies," *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 234, no. 5, pp. 444–457, 2020.

[6] D. Wyss, "Enabling balance training in robot-assisted gait rehabilitation," Ph.D. dissertation, ETH Zurich, 2019.

[7] A. A. Divani, G. Vazquez, A. M. Barrett, M. Asadollahi, and A. R. Luft, "Risk factors associated with injury attributable to falling among elderly population with history of stroke," *Stroke*, vol. 40, no. 10, pp. 3286–3292, 2009.

[8] J. G. Ziegler and N. B. Nichols, "Optimum settings for automatic controllers," *Transactions of the American society of mechanical engineers*, vol. 64, no. 8, pp. 759–765, 1942.

[9] A. Ali and S. Majhi, "Pi/pid controller design based on imc and percentage overshoot specification to controller setpoint change," *ISA transactions*, vol. 48, no. 1, pp. 10–15, 2009.

[10] M. K. Joyo, Y. Raza, S. F. Ahmed, M. Billah, K. Kadir, K. Naidu, A. Ali, and Z. Mohd Yusof, "Optimized proportional-integral-derivative controller for upper limb rehabilitation robot," *Electronics*, vol. 8, no. 8, p. 826, 2019.

[11] A. Ali, S. F. Ahmed, K. A. Kadir, M. K. Joyo, and R. S. Yarooq, "Fuzzy pid controller for upper limb rehabilitation robotic system," in *2018 IEEE international conference on innovative research and development (ICIRD)*. IEEE, 2018, pp. 1–5.

[12] S. Hasan and A. K. Dhingra, "Performance verification of different control schemes in human lower extremity rehabilitation robot," *Results in Control and Optimization*, vol. 4, p. 100028, 2021.

[13] G. Xu, A. Song, and H. Li, "Control system design for an upper-limb rehabilitation robot," *Advanced Robotics*, vol. 25, no. 1-2, pp. 229–251, 2011.

[14] Z. Shen, J. Zhou, J. Gao, and R. Song, "Fuzzy logic based pid control of a 3 dof lower limb rehabilitation robot," in *2018 IEEE 8th Annual International Conference on CYBER Technology in Automation, Control, and Intelligent Systems (CYBER)*. IEEE, 2018, pp. 818–821.

[15] J. Cao, S. Q. Xie, R. Das, and G. L. Zhu, "Control strategies for effective robot assisted gait rehabilitation: the state of art and future prospects," *Medical engineering & physics*, vol. 36, no. 12, pp. 1555–1566, 2014.

[16] L. Marchal-Crespo and D. J. Reinkensmeyer, "Review of control strategies for robotic movement training after neurologic injury," *Journal of neuroengineering and rehabilitation*, vol. 6, no. 1, pp. 1–15, 2009.

[17] M. Manzari, "Modified lokomat simulation," https://brightspace.tudelft.nl/d2l/le/content/500585/viewContent/3154161/View, 2023.

[18] M. Zhang, A. McDaid, A. J. Veale, Y. Peng, and S. Q. Xie, "Adaptive trajectory tracking control of a parallel ankle rehabilitation robot with joint-space force distribution," *IEEE Access*, vol. 7, pp. 85 812–85 820, 2019.

[19] C. ter Welle, "Pd-controller lokomat repository [source code]," https://github.com/karienterwelle/IER_2023, 2023.