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Integrating hip exosuit and FES for lower limb rehabilitation in a simulation environment

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Abstract: Lower-limb rehabilitation for spinal cord injury (SCI) and other motor disorders is often a lengthy process for the patient. The combination of active orthoses and functional electrical stimulation (FES) promises to accelerate therapy outcome, while simultaneously reducing the physical burden of the therapist. In this work, we propose a controller to a hybrid neuroprosthesis (HNP) composed of a hip orthosis and FES-controlled knee motion. In our simulation analysis using a detailed musculoskeletal model, we use experimental data from an able-bodied subject during slow-speed walking to compare the performance provided by such a system. Furthermore, we analyzed the obtained results in comparison to gait data collected from experiments where we used an active hip orthosis. Although the knee stimulation controller still oscillated during gait, we acquired control results with errors smaller than five degrees. Besides, we were able to examine the performance at very slow speeds.

Keywords: Electrical stimulation, Neural control, Biomedical control, Neuroprosthesis, Rehabilitation, Robotics, Exoskeleton

1. INTRODUCTION

Motor disabilities caused by spinal cord injury (SCI) have devastating effects on individual quality of life and affect millions worldwide. Hence, there is a demand to improve rehabilitation techniques in a way that patients achieve the best possible functional outcome. In general, repetitive and intensive movement physiotherapy provides physiological benefits (e.g., prevention of osteoporosis, reduced incidence of fractures and reduced spasticity) when applied along with other equipment, such as active orthoses and functional electrical stimulation (FES).

The term active orthosis typically describes a device used to increase the capacity of a person suffering from a motor pathology (Dollar and Herr, 2008). This expression includes commercially available body weight-supported (BWS) treadmills, full lower limbs active orthoses and mobile active orthoses. BWS treadmills are commercially available for rehabilitation even though the equipment remains expensive, heavy and not user-friendly. Besides, BWS treadmills are unable to directly aid activities of the daily life (ADLs) and may only be usable for indoors training. On the other hand, full lower limbs active orthoses may assist individuals during their ADLs (Donati et al., 2016), yet, are not yet used for clinical therapy purposes. The issue of portability is one of the factors that limit the application of full active orthoses inside and outside clinical therapy. With that purpose, mobile active orthosis results proved promising for both ADLs assist and clinical rehabilitation. Even if there is still a compromise between size, weight, and performance on choosing actuators and maximize benefits from the electromechanical system through more robust controllers (Lovrenovic and Doumit, 2016).

FES stands for a known rehabilitation technique for motor functions improvement, in which the stimulation generates muscle contraction with several rehabilitation benefits, e.g., prevention of atrophy, balance, regulation of spasticity and cardiovascular fitness (Martin et al., 2012). Extensive research work has been carried out for applications of FES systems, such as robot-aided assistance (Adorno et al., 2014), and balance regulation investigation (Vette et al., 2007). The fast muscle fatigue remains the main limitation on the use of only FES for long periods. Results from controllers attempting to compensate fatigue by applying dynamic models are still insufficient (Kirsch et al., 2016). Other FES complexities also include multiple muscles coordination and inadequate stimulation response characteristics (i.e., nonlinearity, electromechanical delays, and time-dependency). Besides, FES control generally intends to provide stimulation for the sagittal plane, yet this may cause scissoring (Farris et al., 2009). Although specific solutions demonstrated the use of learning control of foot eversion in walking stroke patients to avoid scissoring (Seel et al., 2016), this type of approach implies on a search for particular controllers for complex movements as gait.

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One of the purposes of stroke and SCI patients recovery is to increase muscular tonus based on simple gait movements repetition. For ensuring a safety development, this work aims to provide a simulation environment for comparison of HNPs controllers for low speeds in rehabilitation. In state of the art, we looked for recent papers ($\leq$ 15 years) about HNPs, so we could simulate an HNP controller using the open-source software OpenSim (Delp et al., 2007), and start the development an HNP for rehabilitation. Several applications have provided hip extension through surface electrodes on the gluteus maximus muscles, e.g., in FES cycling (Bo et al., 2017). However, FES is unable to recruit the hip flexors muscles (psoas major and iliacus muscle) through surface electrodes. Therefore, for aiding the entire gait movement, we developed a system that combines FES knee control with an active hip orthosis based on (Obinata et al., 2007; Kirsch et al., 2014; Farris et al., 2009; Kobetic et al., 2009). We started by measuring and estimating joint movements during gait. From these trajectories, we developed a closed-loop control strategy in Opensim. This environment allows torques simulation (i.e., active orthosis) and also muscle excitation (i.e., electrical stimulation). This simulation not only intends to ensure the system security but also proposes a simpler environment for more robust controllers inspections.

Section 2 describes a comparison on the state-of-the-art of HNPs. Section 3 presents an overview on the proposed materials and methods followed by the simulation and experiments evaluation in Section 4. The final discussion and conclusions are described in Section 5.

2. STATE OF THE ART

Early HNPs proposals only described simple robotics control techniques with an open loop control of FES (Perez-Orive and Mayagoitia, 1994; Ohashi et al., 1993). Later, experimentally, researchers confirmed that active orthosis provided effective control of movement, and stimulation reduced energy requirements (Popović and Sinkjær, 2000). More recent works evaluated the energy limitations of the mechanical orthosis and then combined to FES to benefit from muscle power (Ha et al., 2016), as well as the other way around (Jailani et al., 2010). Both these approaches lead to lighter systems.

Observing studies that conducted trials, most systems employed surface electrodes for FES with one (Jailani et al., 2010; del Ama et al., 2011; Obinata et al., 2007; Ha et al., 2016), two (Ren and Zhang, 2014; Ha et al., 2016) or three (Alibeji et al., 2016) pairs of surface electrodes on each lower limb. The only exception was the system presented by (Chang et al., 2016), in which they applied a 16 channel intramuscular electrodes system on an individual with mid-thoracic paraplegia. Although intramuscular electrodes provide more selectiveness and require fewer pulse intensities, the use of surface electrodes remains simpler and non-invasive to implement on experiments with different subjects. Other developer decisions remain on which muscle to stimulate. Most methods combined quadriceps and hamstrings. These are the stronger and most essential muscles for knee flexion and extension, showed to be crucial to gait (Perry, 1992). The hip tilt is another essential joint for gait. However, FES is unable to provide a hip extension through the surface electrode. As for the ankle joint, although necessary for a clear step, the passive orthosis is an efficient solution for the task, which simplifies the control system, by reducing one degree of freedom.

As for the type of actuators, most systems employ DC motors (Ren and Zhang, 2014; Obinata et al., 2007; Kirsch et al., 2014; Ha et al., 2016), probably because of their lower cost and simple controllability. Furthermore, another usual employment is the combination of DC motors and spring breaks, which promotes hip extension using the restoring force from gait.

For managing both FES and active orthosis, controllers often considered an offline predetermined trajectory based on the application (e.g., rhythmic swing or gait). As for feedback, most used variables are position (Jailani et al., 2011; Alibeji et al., 2016; Bulea et al., 2014; Ren and Zhang, 2014; Ha et al., 2016), torque (Vallery et al., 2005) or even the relationship of both (Chen et al., 2013). Also, experiments employed a finite state machine at a high level and cooperative combined stimulation and actuators at a lower level. Most projects focused on the first-actuator approach and used stimulation as a way to reduce motor torque (Ren and Zhang, 2014; Ha et al., 2016). However, (Kirsch et al., 2014; Chang et al., 2016) explicit presented a muscle-first approach, in which actuators just adjusted the trajectory or providing movement that stimulation is unable to.

We also identified two simulation projects that evaluated HNPs for rehabilitation exercise for one leg (Vallery et al., 2005; Chen et al., 2013). Using both legs, (Vallery et al., 2005) designed and stimulated a cooperative controller to avoid fatigue for knee gait assist. While (Chen et al., 2013) proposed a press leg exercise in which stimulation and actuators complement the patient effort. As for projects that handled experiments on able-bodied subjects, they conducted most trials on full gait trials (Jailani et al., 2011; del Ama et al., 2014; Obinata et al., 2007; Bao et al., 2016; Ha et al., 2016), even though designing, developing and experimenting demand more funds, security checks and time, which indicates an inclination to employ HNPs for gait assistance. Even trials that experimented with only one leg and rehabilitation exercises expressed the intention to also employ HNPs for gait in the future (Ren and Zhang, 2014). We also recognized fewer groups working with SCI individuals (Ha et al., 2016; Chang et al., 2016) compared to the number of able-bodied subjects experiments. This difference probably indicates the challenge to find volunteers available for tests, in addition to the responsibility of security checks for this population to use the HNP. Besides, we also acknowledge that the projects for HNPs intend to apply the system to SCI individuals.

3. METHODS

In addition to the combination of active orthosis and FES, HNPs must consider that low walking speeds generate different joint movement patterns. Accordingly, we started our work by measuring joint kinematics without any external intervention during gait (Section 3.1). With these

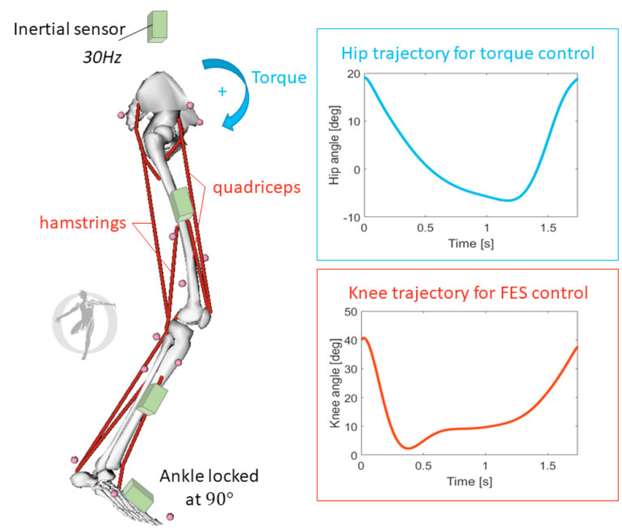

Fig. 1. The complete model for knee flexion and extension in OpenSim, representing the muscles as red lines. We locked the ankle joint at 90 degrees. The model also illustrates the hip and knee trajectory for the controllers. We represented the inertial sensors as blue cubes (on the trunk, upper leg, lower leg, and foot).

measurements, we found the knee and hip patterns, used as trajectories for the control architecture simulated in OpenSim (Section 3.2). Finally, we applied these patterns in a hip actuator controller (Section 3.3).

3.1 Gait kinematics

To measure the joint angles from one leg, we attached four inertial sensors (3-space, Yost Labs, United States) on the trunk, upper leg, lower leg and foot on a non-disabled volunteer (Fig. 1). This volunteer walked on a treadmill (Jog Forma, Technogym, Italy) for 120 seconds at two speeds. For the first 60 seconds, the treadmill maintained first the slowest possible speed ($0.1\,m/s$), and, after, a conservative rehabilitation speed ($0.3\,m/s$). Subsequently, we imported the position datasets into an OpenSim model that allowed the movements of the hip, knee, and ankle joints. OpenSim allows the visualization of musculoskeletal data or simulation results within the GUI.

For each dataset, we also conducted polynomials approximation with an error-based filter and averages to create unique averaged regression functions (Freire et al., 2018). These curves are the step trajectories to be followed by the controllers during gait (hip and knee trajectories are illustrated in Fig. 1).

3.2 Control development

The control architecture is composed of trajectory PID controllers and some mechanical constraints. The two controllers are: (1) a torque PID controller that represents an active orthosis to generate hip movement, and (2) an FES PID controller that makes use of muscles excitation to generate knee movement. For initial simplification of the number of degrees of freedom, we constrained the trunk and ankle: the model does not represent the trunk movement, and we locked the ankle is locked at 90 degrees, representing a passive orthosis. We hypothesize that this simulation should be able to provide movements similar to the regular walking from the volunteer during a rehabilitation scenario.

We simulated this environment on OpenSim and its integration with Matlab. Previous work already proposed and compared controllers with a similar platform for FES cycling (de Sousa et al., 2016). For a forward dynamic simulation, the OpenSim Excitation Editor allows the specification of excitation patterns to muscles (excitation control signal: from 0 to 1). Therefore, we controlled the knee angle position by an FES PID controller actuating on the quadriceps to generate knee extension or on hamstrings to generate knee flexion (Fig. 1). For simplicity, we have not excited quadriceps and hamstrings at the same time, i.e., there is no co-contraction. To simulate the active orthosis, we employed the OpenSim Torque Actuator API to add external torques (actuator control signal: from 0 to 1 multiplied by the maximum torque). Therefore, the torque PID controller controls this torque intensity and direction, i.e., hip extension and flexion (Fig. 1).

Before combining both PID controllers on the gait simulation, we first tuned each PID control parameters individually for the speed $0.1\,m/s$. In this speed, the inertia is less effective; therefore the controllers must be further activated. We tuned the torque control locking the knee at the maximum and minimum values from the trajectories so that we could evaluate the controller performance for both knee angle extremes. We set a maximum mean error of 5 degrees and setting time lower than one gait cycle. After tuning the torque controller, we similarly tuned the FES controller. Finally, with both sets of tuned PID control parameters and the trajectories defined in section 3.1, we simulated 5 steps for both speeds.

3.3 Exosuit trial

Based on the hip trajectory for $0.3m/s$ from section 3.1, we performed one gait test on the same volunteer with the *Hip Exosuit* (Freire et al., 2018) (Fig. 2). The hip actuator (ANT-38, Sito Motors Co., China) generates a linear displacement pattern that, with the transmission system, is converted in rotational movement of the hip joint. Additionally, the orthosis structure stabilizes the trunk by gears around the user's back. During this test, we did not apply FES, and the volunteer did not walk on a treadmill. We instructed the volunteer to walk with normal knee and ankle movements, but with no interference on the hip joint. We measured the gait period, the gait step length, and the hip angle trajectory for 60 seconds, and later we compared these results to the simulation.

4. SIMULATION AND EXPERIMENTS EVALUATION

For the kinematic analysis, we evaluated the measurements acquired from the experiment described in section 3.1 at $15\,Hz$. The volunteer reported struggle to maintain balance at $0.1\,m/s$. We imported these measurements into OpenSim and visualized the model joints performing the movement, which exposed the dragging foot at $0.1\,m/s$. Nevertheless, even at a low sample frequency and the dragging foot, the polynomials approximation were able to

Fig. 2. The Hip Exosuit used by the volunteer. The ANT-38 actuator provides torque on the hip, while the Yost Labs IMU measures the hip angle position.

create the gait trajectories functions, creating trajectories with enough points for the controller frequency.

For both speeds, we executed the torque and FES PID controllers for one step gait trajectory at their respective individual tuning as described in section 3.2. However, when we applied both controllers simultaneously, an additional tuning was required for the torque PID controller. During gait, the knee flexes while the hip extends (i.e., the knee pushes against the hip). Therefore, we increased the proportional gain from the PID to increase the hip actuation. Figures 3 and 4 show two steady-state gait cycles for the hybrid control at $0.1\,m/s$ and $0.3\,m/s$, respectively. For both simulations, we set the control frequency at $50\,Hz$. At $0.1\,m/s$, the root-mean-square deviation error (RMSE) for the hip was 3.07 degrees and for the knee was 3.37 degrees, and, at $0.3\,m/s$, was 3.10 and 2.72, for hip and knee, respectively. The results from Fig. 3 and 4 show that the lower speed demands more from the controllers, probably due to lack of inertia effect. Also, the FES PID controller still oscillates, compared to the torque PID controller, especially when the knee pushes the hip.

For a first trial, we tested the Hipsuit actuation performance with the Hip Exosuit. We set a constant torque, only changing the direction of the actuation respecting the hip trajectory (i.e., positive for flexion and negative hip extension). As previously presented on (Freire et al., 2018), due to development limitations, the active orthosis only achieved a maximum speed of $0.15\,m/s$ (gait period of $5.35\,s$) even at maximum torque, when we aimed 0.3 m/s (Fig. 5). The actuators were not able to provide enough actuation especially on the moments in which the knee pushes the hip. The volunteer and the orthosis performance compensated the efficiency losses and transition delay by performing longer steps ($0.75\,m$) compared to $0.53\,m$ for $0.1\,m/s$ and $0.25\,m$ for $0.3\,m/s$. We should fix these development problems before testing the controllers.

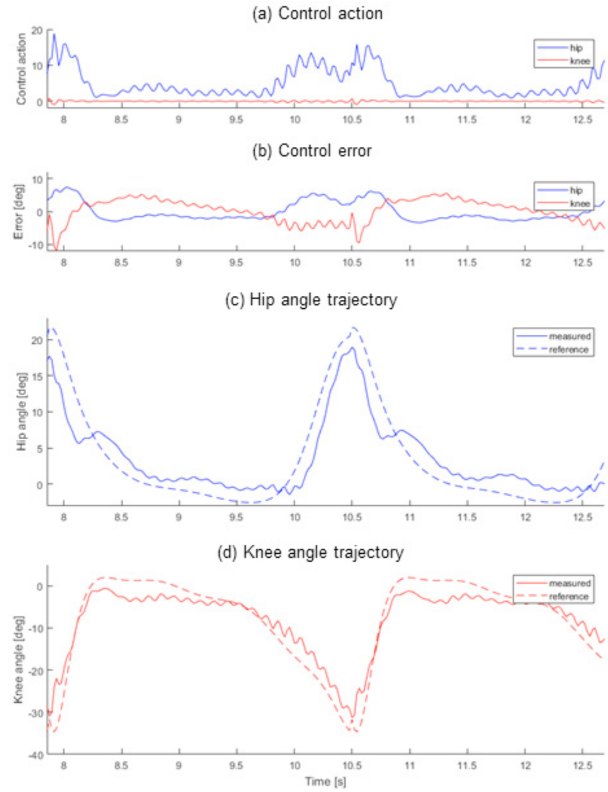

Fig. 3. Simulation results for 2 steps during permanent regime for the HNP control at $0.1\,m/s$ (gait period was $2.54\,s$). (a) Control signal for torque (hip, blue) and FES (knee, red) controllers. (b) Control errors. (c) Hip angle measurements and trajectory reference. (d) Knee angle measurements and trajectory reference.

5. CONCLUSIONS

Other HNPs already employed active orthoses on hips and FES on knees (Obinata et al., 2007; Kirsch et al., 2014; Farris et al., 2009; Kobetic et al., 2009). However, to our knowledge, there is no proposal for a simulation platform environment for HNPs. In this work, we presented a platform using OpenSim tools for musculoskeletal movement analysis to expose controllers particularities before testing on gait rehabilitation. Further, we described and conducted a methodology for creating trajectories for gait, and we worked on a methodology for tuning PID controllers for two joints. Even though a second refinement was required, from the initial parameters, the system was stable and operating. The simulation results presented maximum absolute errors below 10 degrees and RMSE below 4 degrees for all simulations and joints.

For both speeds, the controlled joint trajectories were similar to the original patterns, considering timing and shape. The nuances revealed that the FES control oscillates more intensively for the lower speed. From these results, we identified the critical movement (i.e., knee flexion along with hip extension) and difficulty for lower speeds. As

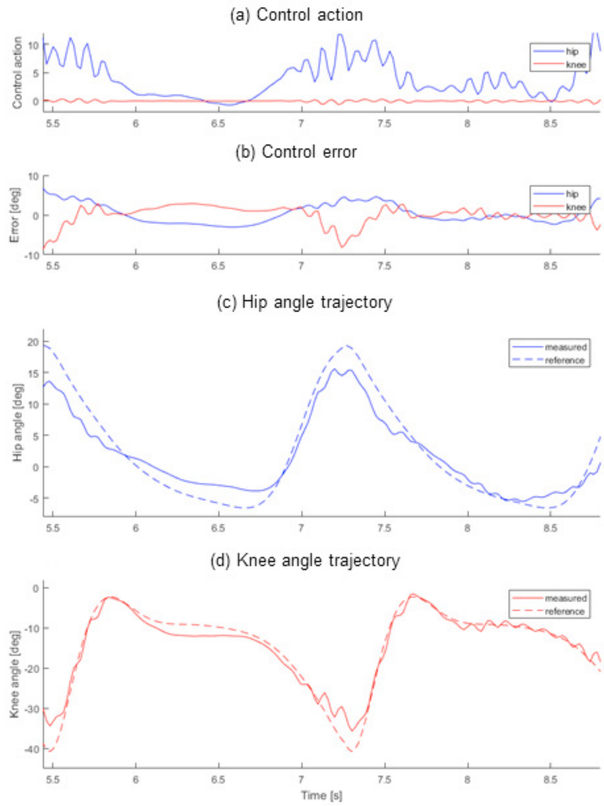

Fig. 4. Simulation results for 2 steps during permanent regime for the HNP control at $0.3\,m/s$ (gait period was $1.76\,s$). (a) Control signal for torque (hip, blue) and FES (knee, red) controllers. (b) Control errors. (c) Hip angle measurements and trajectory reference. (d) Knee angle measurements and trajectory reference.

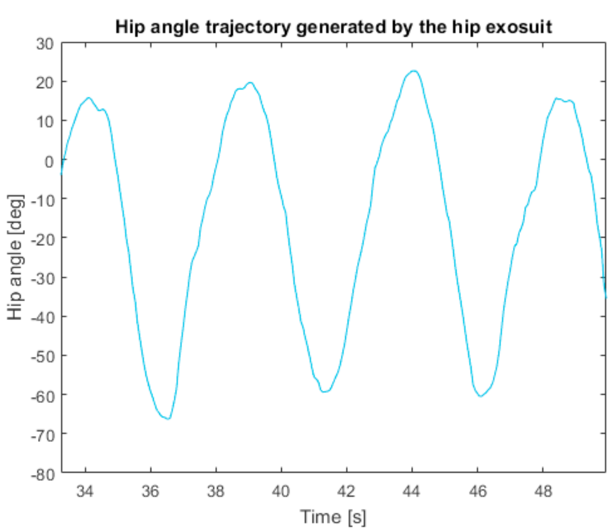

Fig. 5. Experimental results for the active orthosis (gait period was $5.35\,s$).

rehabilitation gait is closer to $0.3\,m/s.$, we consider these results adequate. Even though our practical tests with the new Hip Exosuit were not satisfactory, changing the actuator should increase torque allowing following tests with PID controllers. Therefore, in the future, we intend to use the simulation results for comparison.

For future work, we intend to include stability analysis for different patients (model scales). Furthermore, we should take into account the trunk mass/inertia in our simulation. Besides, we intend to explore the use of co-activation of antagonist muscles using FES, which may improve disturbance rejection (Bó et al., 2016), in addition to other control approaches, such as (Müller et al., 2017) that could eliminate the oscillating behavior of the FES controller.

REFERENCES

Adorno, B.V., Bó, A.P.L., and Fraisse, P. (2014). Kinematic modeling and control for human-robot cooperation considering different interaction roles. *Robotica*, 33(02), 314–331. doi:10.1017/s0263574714000356.

Alibeji, N., Kirsch, N., and Sharma, N. (2016). An adaptive low-dimensional control to compensate for actuator redundancy and FES-induced muscle fatigue in a hybrid neuroprosthesis. *Control Engineering Practice*. doi: 10.1016/j.conengprac.2016.07.015.

Bao, X., Kirsch, N., and Sharma, N. (2016). Dynamic control allocation of a feedback linearized hybrid neuroprosthetic system. In *2016 American Control Conference (ACC)*, 3976–3981. IEEE. doi: 10.1109/ACC.2016.7525534.

Bó, A.P.L., da Fonseca, L.O., and de Sousa, A.C.C. (2016). FES-induced co-activation of antagonist muscles for upper limb control and disturbance rejection. *Medical Engineering & Physics*, 38(11), 1176–1184. doi: 10.1016/j.medengphy.2016.07.004.

Bo, A.P., da Fonseca, L.O., Guimaraes, J.A., Fachin-Martins, E., Paredes, M.E., Brindeiro, G.A., de Sousa, A.C.C., Dorado, M.C., and Ramos, F.M. (2017). Cycling with spinal cord injury: A novel system for cycling using electrical stimulation for individuals with paraplegia, and preparation for cybathlon 2016. *IEEE Robotics & Automation Magazine*, 24(4), 58–65. doi: 10.1109/mra.2017.2751660.

Bulea, T.C., Kobetic, R., Audu, M.L., Schnellenberger, J.R., Pinault, G., and Triolo, R.J. (2014). Forward stair descent with hybrid neuroprosthesis after paralysis: Single case study demonstrating feasibility. *Journal of rehabilitation research and development*, 51(7), 1077–94. doi:10.1682/JRRD.2013.12.0257.

Chang, S.R., Nandor, M.J., Li, L., Foglyano, K.M., Schnellenberger, J.R., Kobetic, R., Quinn, R.D., and Triolo, R.J. (2016). A stimulation-driven exoskeleton for walking after paraplegia. In *2016 38th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, volume 44106, 6369–6372. IEEE. doi:10.1109/EMBC.2016.7592185.

Chen, Y., Hu, J., Wang, W., Peng, L., Peng, L., and Hou, Z.G. (2013). An FES-assisted training strategy combined with impedance control for a lower limb rehabilitation robot. In *2013 IEEE International Conference on Robotics and Biomimetics (ROBIO)*, 2037–2042. IEEE. doi:10.1109/ROBIO.2013.6739769.

de Sousa, A.C.C., Ramos, F.M., Dorado, M.C.N., da Fonseca, L.O., and Bó, A.P.L. (2016). A comparative study on control strategies for FES cycling using a detailed musculoskeletal model. *IFAC-PapersOnLine*, 49(32), 204–209. doi:10.1016/j.ifacol.2016.12.215.

del Ama, A.J., Moreno, J.C., Gil-Agudo, A., de-los Reyes, A., and Pons, J.L. (2011). Online assessment of human-robot interaction for hybrid control of walking. *Sensors*, 12(1), 215–225.

del Ama, A.J., Moreno, J.C., Gil-Agudo, Á., and Pons, J.L. (2014). Hybrid fes-robot cooperative control of ambulatory gait rehabilitation exoskeleton. *Journal of NeuroEngineering and Rehabilitation*, 11(1), 27. doi: 10.1186/1743-0003-11-27.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., and Thelen, D.G. (2007). OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement. *IEEE Transactions on Biomedical Engineering*, 54(11), 1940–1950. doi:10.1109/TBME.2007.901024.

Dollar, A.M. and Herr, H. (2008). Lower extremity exoskeletons and active orthoses: Challenges and state-of-the-art. *IEEE Transactions on Robotics*, 24(1), 144–158. doi:10.1109/tro.2008.915453.

Donati, A.R.C., Shokur, S., Morya, E., Campos, D.S.F., Moioli, R.C., Gitti, C.M., Augusto, P.B., Tripodi, S., Pires, C.G., Pereira, G.A., and et al. (2016). Long-term training with a brain-machine interface-based gait protocol induces partial neurological recovery in paraplegic patients. *Scientific Reports*, 6(1). doi: 10.1038/srep30383.

Farris, R.J., Quintero, H.a., Withrow, T.J., and Goldfarb, M. (2009). Design and simulation of a joint-coupled orthosis for regulating FES-aided gait. In *2009 IEEE International Conference on Robotics and Automation*, 1916–1922. IEEE. doi:10.1109/ROBOT.2009.5152634.

Freire, J.P., Bó, A.P.L., and Rocha, T.S. (2018). Exosuit for alternative hip actuation: a proof of concept. In *6 Encontro Nacional de Engenharia Biomecânica*.

Ha, K.H., Murray, S.A., and Goldfarb, M. (2016). An Approach for the Cooperative Control of FES With a Powered Exoskeleton During Level Walking for Persons With Paraplegia. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 24(4), 455–466. doi: 10.1109/TNSRE.2015.2421052.

Jailani, R., Tokhi, M., and Gharooni, S. (2010). Fuzzy logic control of knee extension for FES-assisted walking with spring brake orthosis. In *2010 IEEE International Conference on Systems, Man and Cybernetics*, 379–385. IEEE. doi:10.1109/ICSMC.2010.5641745.

Jailani, R., Tokhi, M., Gharooni, S., and Jogtaei, M. (2011). Finite State Control of FES-Assisted Walking with Spring Brake Orthosis. In *2011 UkSim 13th International Conference on Computer Modelling and Simulation*, 183–188. IEEE. doi:10.1109/UKSIM.2011.43.

Kirsch, N., Alibeji, N., Dicianno, B.E., and Sharma, N. (2016). Switching control of functional electrical stimulation and motor assist for muscle fatigue compensation. In *2016 American Control Conference (ACC)*, 4865–4870. IEEE. doi:10.1109/ACC.2016.7526123.

Kirsch, N., Alibeji, N., Fisher, L., Gregory, C., and Sharma, N. (2014). A semi-active hybrid neuroprosthesis for restoring lower limb function in paraplegics. In *2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, volume 2014, 2557–2560. IEEE. doi: 10.1109/EMBC.2014.6944144.

Kobetic, R., To, C.S., Schnellenberger, J.R., Audu, M.L., Bulea, T.C., Gaudio, R., Pinault, G., Tashman, S., and Triolo, R.J. (2009). Development of hybrid orthosis for standing, walking, and stair climbing after spinal cord injury. *The Journal of Rehabilitation Research and Development*, 46(3), 447. doi:10.1682/JRRD.2008.07.0087.

Lovrenovic, Z. and Doumit, M. (2016). Review and analysis of recent development of lower extremity exoskeletons for walking assist. In *2016 IEEE EMBS International Student Conference (ISC)*, 1–4. IEEE. doi: 10.1109/EMBSISC.2016.7508620.

Martin, R., Sadowsky, C., Obst, K., Meyer, B., and McDonald, J. (2012). Functional Electrical Stimulation in Spinal Cord Injury: From Theory to Practice. *Topics in Spinal Cord Injury Rehabilitation*, 18(1), 28–33. doi: 10.1310/sci1801-28.

Müller, P., Balligand, C., Seel, T., and Schauer, T. (2017). Iterative learning control and system identification of the antagonistic knee muscle complex during gait using functional electrical stimulation. *IFAC-PapersOnLine*, 50(1), 8786–8791. doi:10.1016/j.ifacol.2017.08.1738.

Obinata, G., Fukada, S., Matsunaga, T., Iwami, T., Shimada, Y., Miyawaki, K., Hase, K., and Nakayama, A. (2007). Hybrid Control of Powered Orthosis and Functional Neuromuscular Stimulation for Restoring Gait. In *2007 29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, volume 2, 4879–4882. IEEE. doi:10.1109/IEMBS.2007.4353433.

Ohashi, T., Obinata, G., Shimada, Y., and Ebata, K. (1993). Control of hybrid FES system for restoration of paraplegic locomotion. In *Proceedings of 1993 2nd IEEE International Workshop on Robot and Human Communication*, 96–101. IEEE. doi: 10.1109/ROMAN.1993.367741.

Perez-Orive, J. and Mayagoitia, R.E. (1994). Closed-loop control system to be used with a hybrid RGO system. *Science China Technological Sciences*, 16(pt 1), 410–411. doi:10.1109/IEMBS.1994.411979.

Perry, J. (1992). *Gait Analysis: Normal and Pathological Function*. SLACK Incorporated, Thorofare, New Jersey.

Popović, D.B. and Sinkjær, T. (2000). *Control of Movement for the Physically Disabled*. Springer London, London. doi:10.1007/978-1-4471-0433-9.

Ren, Y. and Zhang, D. (2014). FEXO knee: A rehabilitation device for knee joint combining functional electrical stimulation with a compliant exoskeleton. *Proceedings of the IEEE RAS and EMBS International Conference on Biomedical Robotics and Biomechatronics*, 683–688. doi:10.1109/BIOROB.2014.6913857.

Seel, T., Werner, C., and Schauer, T. (2016). The adaptive drop foot stimulator – multivariable learning control of foot pitch and roll motion in paretic gait. *Medical Engineering & Physics*, 38(11), 1205–1213. doi: 10.1016/j.medengphy.2016.06.009.

Vallery, H., Stützle, T., Buss, M., and Abel, D. (2005). Control of a hybrid motor prosthesis for the knee joint. *IFAC Proceedings Volumes*, 38(1), 76–81. doi: 10.3182/20050703-6-CZ-1902.01415.

Vette, A., Masani, K., and Popovic, M. (2007). Implementation of a physiologically identified PD feedback controller for regulating the active ankle torque during quiet stance. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 15(2), 235–243. doi: 10.1109/tnsre.2007.897016.