

Passive Knee Orthoses Assistance in Functional Electrical Stimulation Cycling in an Individual With Spinal Cord Injury

Ana C. C. de Sousa¹, Member, IEEE, Felipe S. C. Sousa², Roberto de S. Baptista³, Member, IEEE, and Antônio P. L. Bó⁴, Senior Member, IEEE

Abstract—Functional Electrical Stimulation (FES) may be used in rehabilitation and assistance of people with Spinal Cord Injury (SCI). One significant application is facilitating physical exercise, mainly when combining FES with mechanical platforms, such as tricycles. However, there are still technical challenges in FES cycling protocols, such as improving control and cycling performance. Here we show how passive elements in knee orthoses during FES cycling could increase the average cadence, taking advantage of the cycling movement. Our approach is twofold. First, we simulated the forward dynamics of a detailed musculoskeletal model with passive elements over the knees. Simulations showed that specific spring stiffness ranges increased the crankset speed during cycling by more than 50%. Using parameters found in simulations, we built a pair of passive orthoses and performed experiments with one individual with SCI. During two days, the volunteer cycled with similar stimulation magnitude with and without the passive elements. We observed that the average crankset speed was higher by more than 10% when the springs were attached to the passive orthoses. These results show the potential of using passive elements to increase cycling speed for FES cycling with similar or even lower stimulation magnitude, leading to longer exercise duration.

Index Terms—Functional electrical stimulation, lower limbs, neuroprosthesis, passive orthoses, assisted cycling.

I. INTRODUCTION

SPINAL cord injury (SCI) is a severe clinical condition with substantial physical, psychic, and social repercussions. Beyond the sensorimotor impairment, an individual

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Ana C. C. de Sousa was with the Department of Electrical Engineering, University of Brasília, Brasília 70910-900, Brazil. She is now with the School of Allied Health Sciences, Griffith University, Gold Coast, QLD 4222, Australia (e-mail: anacsousa@lara.unb.br).

Felipe S. C. Sousa and Roberto de S. Baptista are with the Department of Electrical Engineering, University of Brasília, Brasília 70910-900, Brazil.

Antônio P. L. Bó was with the Department of Electrical Engineering, University of Brasília, Brasília 70910-900, Brazil. He is now with the School of Information Technology and Electrical Engineering, The University of Queensland, Brisbane, QLD 4072, Australia.

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with SCI may present uncontrolled physiological function of the bladder and bowel, edemas due to circulation impairments, spasticity, ossification, and other complications [1]. There are several rehabilitation exercises targeting muscle strengthening, circulation improvement, pressure release, and bone density increase to diminish some of these conditions. However, since the outcome is often limited, there is a demand to enhance rehabilitation, so that patients achieve their best possible functional outcome at the shortest period. In general, functional electrical stimulation (FES) and orthoses assistance increase these rehabilitation benefits [2]. FES stands for a known rehabilitation technique, in which the stimulation may generate muscle contraction for applications such as cycling [3]–[5].

However, early works on FES cycling [6] have already suggested that unfavorable biomechanics may be a contributory factor to a low efficiency [3], [7]. These biomechanics include poor recruitment of muscle groups, non-optimal timing of muscle activation, and lack of synergistic and antagonistic joint control. For these reasons, researchers focus on optimizing the recruitment of muscle groups (e.g., [8]) and timing the muscle activation (e.g., [9]), which have the most effect on the overall cycling performance. A competition called Cybathlon promoted the use of powered assistive technology, including FES cycling, which boosted the literature contributions on the topic [10]. During the competition, any control strategy was allowed to maximize speed and fatigue resistance [5], e.g., manual triggering or automatic closed-loop control strategies using embedded sensors.

Nevertheless, other mechanical features implementations to improve performance remains unexplored, such as the use of energy storage devices to support FES cycling in experiments. The simple addition of energy storage devices (usually elastic with mechanical springs) could make the unfavorable biomechanics of cycling more favorable. In these conditions, we could increase cycling speed, providing a more natural movement for rehabilitation and competition.

This paper investigates whether it would be feasible to use passive elastic mechanical orthoses to increase cadence. Using a detailed musculoskeletal model, we (i) investigate candidate parameters, and then we (ii) evaluate experimentally training sets on one volunteer with complete SCI using passive knee orthoses assistance.

A. Related Work

There were efforts in the study of passive knee orthoses in other rehabilitation exercises, e.g., in rowing [11] and elliptical stepping [12]. Both works [11], [12] simulated passive knee orthoses with springs storing energy during knee extension and then releasing energy during flexion. The results showed that the addition of the passive element reduced the overall level of quadriceps excitation.

As for cycling, [13] simulated elastic elements built in the bicycle to eliminate dead points of the pedal cycle. In a dead point, the angular velocity equals zero. The authors modeled elastic cables attached to the bicycle, where the spring releases elastic potential energy during knee flexion. The simulation results of a quadriceps FES controller indicated that the passive elements could eliminate dead points as the elements generate power to the crank, and reduce the number of stimulation channels required in FES cycling.

Later, [14] simulated a flywheel and an electrical clutch mechanism at the crank to maintain cycling cadence. The clutch engaged and disengaged a flywheel to assist or retard the cycling when necessary, while fuzzy logic control stimulates the quadriceps. Results showed that the quadriceps stimulation intensity decreased when using the elastic mechanism, consequently delaying muscle fatigue. As [14] used an electronic controller that actuates the device at the bicycle, it is not considered entirely passive. We also interpret that [13] and [14] added elements to the bicycle, changing its mechanical structure; therefore, we do not classify these devices as orthoses.

[15] simulated energy storage passive knee orthoses for FES cycling. The author identified that the performance depended on the spring position and parameters (spring constant and spring natural length). Therefore, he simulated and evaluated efficiency (work accomplished over the energy expended) for some basic setups: three spring positions, ten spring constants, spring only released energy during flexion, and they only stimulated the quadriceps muscles. Results led to the conclusion that when the spring constant (stiffness) increases, the cycling efficiency also increases.

Although [13]–[15] indicated that passive elements could positively affect cycling performance; these works only simulated basic setups and lacked experimental validation.

In a setup without FES, [16], [17] experimentally validated passive knee orthoses for non-disabled cyclists. In their concept validation [16], the spring stored energy from knee flexion and released in knee extension. They based their approach on the unbalanced effort required from the quadriceps and hamstrings during the same cycling cadence [18]. The quadriceps electromyography (EMG) signal from one user (using the passive orthoses) showed a decrease in the activity as the leg moves around the pedal crank top dead center. After concept validation, they compared the quadriceps EMG of three participants with and without the orthoses [17]. With the same cycling power, results presented less muscle activity in all participants when using the spring support. Although these results validated cycling performance with passive orthoses with experiments, this device intends to improve

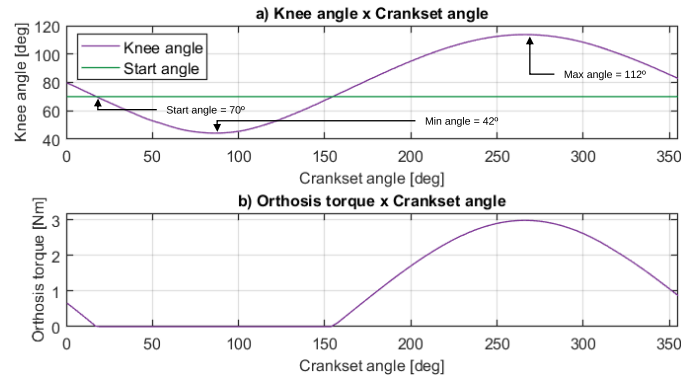


Fig. 1. Example of the spring with starting angle of 52° and maximum torque of 15 Nm° during a complete crankset cycle. (a) Right knee (purple line) and spring start angle (green line). (b) Passive orthosis torque (purple line), the torque becomes zeros when the right knee angle is lower than 52° .

athletes training, not FES rehabilitation for individuals with SCI.

To our knowledge, no previous experiments combined passive orthoses with FES cycling in either disabled or non-disabled volunteers.

II. MATERIALS AND METHODS

A. Passive Knee Orthoses: Model and Design

In a previous work [19], we modeled the passive orthoses based in [16]. The supporting knee torque (τ_{spr}) operates as a rotational spring

$$\tau_{spr} = \begin{cases} K_s(\theta_j - \theta_s), & \theta_j \geq \theta_s \\ 0, & \theta_j < \theta_s \end{cases} \quad (1)$$

where K_s represents the spring stiffness, θ_j represents the knee joint angle, and θ_s the starting angle at which the spring engages and disengages. Fig. 1 illustrates an example of a spring with $\theta_s = 70^\circ$ and $\tau_{max} = 3 \text{ Nm}$. At this representation, 0° refers to the maximum knee extension, i.e., while the knee angle increases, the leg flexes, and while it decreases, the leg extends. In Fig. 1b, we may observe that the spring provides the maximum torque τ_{max} during maximum flexion. In this work, intensity of elasticity is referred to as the maximum torque, an approach that is often employed in the corresponding biomechanics literature [18].

In this model, the spring, while activated, releases energy (i.e., aids the cycling movement) for half the time and stores energy (i.e., resists the cycling movement) for the other half. We expect to determine if the spring should first release or store energy. From Fig. 1, we observe two knee inflection angles (θ_i), where the knee stops flexing and starts extending, or the other way around. In our simulation study and experimental evaluation, we constrained the inflection point of the orthosis torque at the same inflection angle of the knee. As the movement is a cycle, and the bicycle's and user's geometry does not change, the knee angles remain the same.

Regarding our prototype for experiments (Fig. 2), we used a pair of passive knee orthoses with torsion spring

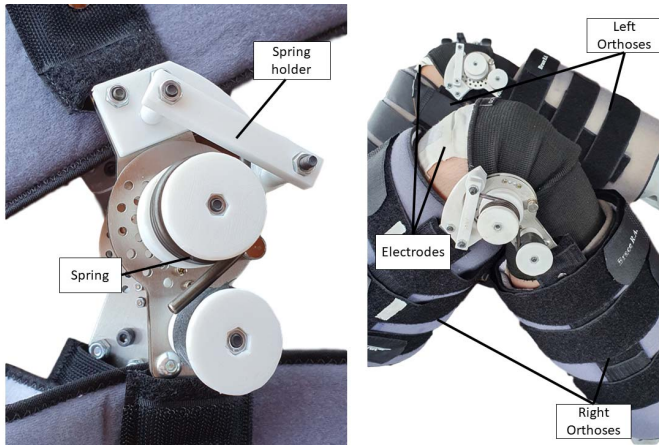


Fig. 2. Details of the knee passive orthoses for FES cycling.

support (Fig. 2) to validate the concept of energy storage devices in FES cycling. For the prototype, we used a commercial knee orthosis (Knee Brace R.A, Salvapé, Brazil) to ensure proper support of the orthosis without injuring the skin or knee joint. We adapted the original structure to receive the elastic element, placing supports (spring holders) for fixing the spring. The knee orthosis accommodates the full range of cycling leg motion, only limiting knee hyperextension.

We fabricated a spring holder prototype with 3D printed polyethylene terephthalate glycol to position the spring over the knee and created torsion in a specific cycling range. Therefore, the spring holder mechanically defines the range selection. We added two pieces of equal-stiffness torsion AISI 1080 carbon steel springs to our design at each spring holder. It is possible to change these springs to different stiffness for future evaluation.

B. Simulation Environment

The underlying simulation framework requires OpenSim and its integration with Matlab (Section II-B.1). Within this environment, we control cadence in 20s simulated trials, where different simulation parameters are modified in each run (Section II-B.2).

1) *Basic Simulation Framework*: Fig. 3 illustrates the model initially introduced in [19].¹ The model uses the OpenSim platform [20], which is an open-source software to simulate highly detailed musculoskeletal models. Using a graphical interface, users can generate simulations with default models or develop new models and controllers.

The hips and knees of the model run freely, whereas the pelvis and ankles are locked. The pedal holders connect the feet to the pedals and immobilize the ankles. Consequently, the pedal holders transfer elastic forces to the pedal using contact geometries (more details in [21]). We should observe that the simplification of the pelvis of the cyclist locked over the seat is unrealistic. During an experiment, the pelvis moves

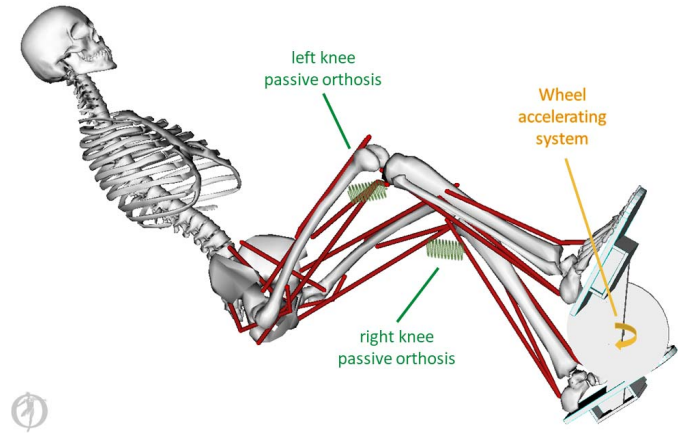


Fig. 3. Model for investigating FES cycling and passive orthoses in simulation. It has been adapted from previous works with the addition of left and right passive knee orthoses (green) and a wheel accelerating system at the crankset (orange).

and creates forces against the support, decreasing the overall performance.

During one pedal stroke, the quadriceps provide most torque for the pedal stroke through knee extension. Hamstrings pull the feet to the top while the gluteus provides more power for knee extension. For efficient cycling, these muscle groups must be excited in specific ranges, depending on crankset angle and speed. The excitation control incorporates predefined muscles range angles, which were previously defined in [21]. In OpenSim, the interval of muscles excitation (u_{exc}) is [0, 1].

We also modeled a wheel acceleration system that guarantees that the system starts cycling by applying any possible initial spring resistance

$$\tau_{acc} = \begin{cases} 15 \text{ Nm}, & \dot{\theta}_c < \dot{\theta}_t \\ 0, & \dot{\theta}_c \geq \dot{\theta}_t \end{cases} \quad (2)$$

where $\dot{\theta}_c$ is the cycling cadence. We selected the wheel acceleration system torque (15 Nm) empirically. After the cycle achieves the target crankset cadence $\dot{\theta}_t$, the system may keep cycling due to the bicycle's inertia and geometry conditions. Therefore, we turn off the acceleration.

2) *Simulation Methodology*: Previously, we simulated the spring torque in a narrow configuration [19]. In the current work, we extended this simulation study to investigate how friction losses, different quadriceps excitation, and different muscles affect the cycling movement.

In OpenSim, we fixed the control frequency at 50 Hz, the duration of simulation as 20 s, and the passive orthoses inflection point around 112° , which is the same inflection knee angle.

The purpose of this work is to find the maximum cadence that the volunteer could achieve. Using a bang-bang (BB) controller limits the number of variables impacting the experiment. Therefore, we designed a bang-bang controller (i.e., fixed excitation). We set the controller with three different muscle group sets, only quadriceps (Q), quadriceps, and hamstrings (QH), and quadriceps, hamstrings, and gluteus (QHG). For each of these muscle sets, we compared the effect of dif-

¹It is being developed and maintained in https://simtk.org/docman/?group_id=1553.

TABLE I
CONFIGURATIONS FOR SIMULATIONS TO DETERMINE THE EFFECT OF
MAXIMUM TORQUE AND THE STARTING ANGLE

Features for τ_{max}	Configuration (480 simulations)
Muscles set	{Q, QH, QHG}
Excitation	$h = \{1.0, 0.9, 0.8, 0.7, 0.6\}$
Load [Nm]	$L = \{0, 0.01\}$
Maximum spring torque [Nm]	$\tau_{max} = \{-40, -35, -30, \dots, 35, 40\}$
Features for $\Delta\theta$	Configuration (180 simulations)
Muscles set	{Q, QH, QHG}
Excitation	$h = \{1.0, 0.9, 0.8, 0.7, 0.6\}$
Load [Nm]	$L = \{0, 0.01\}$
Angle range [°]	$\Delta\theta = \{60, 50, 40, 30, 20, 10\}$

ferent excitation magnitudes ($h = \{1.0, 0.9, 0.8, 0.7, 0.6\}$) with and without a load at the crankset ($L = 0$ and $L = 0.01$ Nm). Although we defined the manufacturing requirement elastic spring constant lower than 0.5 Nm° , we first performed a broader study with spring parameters that did not necessarily lead to appropriate elastic values.

The first step of the simulations is to determine the most efficient maximum spring torque τ_{max} . Therefore, we simulated the system with a fixed θ_s and ranged the τ_{max} from -40 Nm to 40 Nm, in which the negative signal determines that the spring releases energy at extension, and the positive signal determines that the spring releases at flexion. A τ_{max} higher than $|40|$ Nm may lead to simulations where cycling movement is not achieved, i.e., muscles cannot provide enough torque for cycling movement when the spring is storing energy. Figure 1 illustrates the range, in which the orthosis actuates: when the angle is greater than 70° . Therefore, the spring actuates in almost the entire range of movement, $\Delta\theta = \theta_i - \theta_s = 42^\circ$. Table I summarizes all 480 simulations (3 muscle group sets, 5 excitation magnitudes, 2 loads and 16 maximum torques (τ_{max})).

After these simulations, we fixed τ_{max} and then investigated $\Delta\theta$. The complete range of this model's knee trajectory is approximately 70° . Therefore, we ranged $\Delta\theta$ from 10° to 60° . Table I summarizes all 180 simulations (3 muscle group sets, 5 excitation magnitudes, 6 angle ranges ($\Delta\theta$) and 2 loads).

For each simulation, we observed cadence over time, and calculated two performance parameters: the average crankset cadence $\bar{\theta}_c$ and the standard deviation $\sigma_{\bar{\theta}}$. We set cadence as a parameter performance, as often used in other works [10]. From these performance parameters, we analyze how τ_{max} and $\Delta\theta$ could modify the average cadence of a complete cycle with similar muscle excitations (h).

C. Experimental Setup

In our experiments, we used similarly a bang-bang (BB) controller and stimulated only the quadriceps muscles based on a similar range of Fig. 4. The left orthosis starts actuating when the crankset angle reaches around 154° until 17° . As similarly presented in Fig. 1, the knee angle range differs from the crankset. In our experiment, the knee range is 42° , and the crankset range is 223° .

The experimental setup (Fig. 5) is based on a RehaStim stimulator (Hasomed, Germany), a wireless inertial measurement unit (IMU, 3-space, YEI Technology, USA),

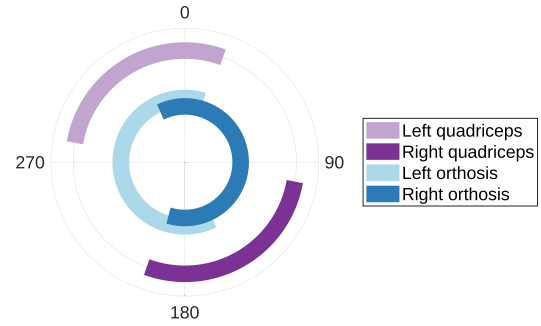


Fig. 4. Muscles and spring range angles for excitation during a pedal stroke. Left and right quadriceps in blue and left and right orthoses in green. We represent the left side in light colors and the right side in dark colors.

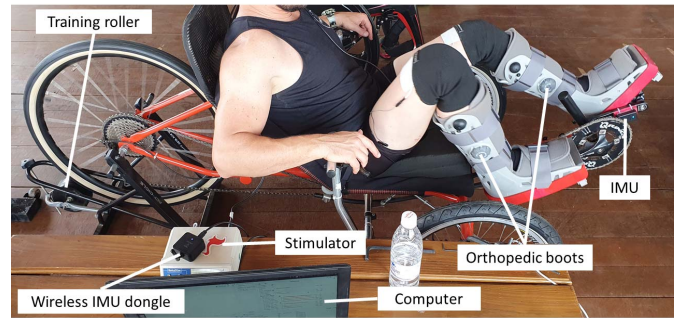


Fig. 5. The EMA trike complete system. The computer provides the signal to the stimulator based on the crankset angle measured by the IMU. Through surface electrodes on the subject, the stimulator applies the corresponded signal to the muscles to cycle the tricycle. The orthopedic boots attached to the pedals avoid the volunteer's ankle movement and align the legs, so it does not fall out of the sagittal plane.

the EMA tricycle setup (HP3 Trikes, Brazil), customized passive orthoses, and a computer running the control system. The IMU measures the current position at 160 Hz and, based on this information, the controller calculates the current cadence and determines the stimulation signal to send to the stimulator. We kept the stimulation frequency at 50 Hz and the pulse width at 500 us .

We integrated the crankset angle information with the stimulator signal using the Robot Operating System (ROS),² running at 50 Hz .

D. Subject and Protocol

The subject was a 41-year-old male, with American Spinal Injury Association Impairment Scale designation A (AIS A), level T9, injured seven years earlier. This volunteer already uses FES in a regular basis and presents healthy skin and normal cardiovascular response to stress and physical exercise. A local ethical committee³ approved the experimental tests, and the subject provided written consent.

We monitored the volunteer during two days of FES cycling training. The participant cycled with and without the spring on

²The software is being developed and maintained in https://github.com/lara-unb/ema_fes_cycling.

³Presentation Certificate for Ethical Appreciation (CAAE): 11717119.3.0000.0030.

each day to guarantee the same electrode positioning. On the first day, the user cycled first with the springs attached, and on the second day, the user cycled first without it. Before the experiments, the volunteer did not use any electric stimulation at the quadriceps for at least 24h.

Each training consisted of two phases: the warming-up phase and the training phase. After adjusting all equipment and the cyclist to the bike, we start a warming-up phase by manually moving the pedals, and we slowly increase the stimulation current applied to muscles. We consider the warming-up phase finished when the volunteer stimulation current reaches around 50mA. At this current, the volunteer can complete a few cycles without any external aid (manually moving the pedals). After the warming-up phase, we kept all external interferences at a minimum.

Then, we start the training phase, from 50mA until a maximum of 80mA. For data analysis, we selected datasets within this range. The volunteer may only keep cycling with the same stimulation current for a few minutes, stopping at the dead points of the pedal cycle (in this setup, these points are around 90° and 270°). Therefore, we adapted the stimulation current: when the cycle stopped, we increased the stimulation current by 2mA.

III. RESULTS

A. Simulations

To analyze the interference from the passive orthoses, we compared results with passive orthoses assistance and without orthoses ($\tau_{max} = 0$), always considering a similar h . Sections III-A.1 and III-A.2 present the results when we ranged τ_{max} and θ_s , respectively.

1) *Maximum Torque*: For all simulations, cycling achieved steady-state response, i.e., the speed during a subsequent cycle was the same. Therefore, we plotted the average cycling speed at the last cycle ($\bar{\theta}_c$) of all simulation results in Fig. 6, as a representation of the simulation performance. We define performance as the ratio between $\bar{\theta}_c$ and the baseline $\bar{\theta}_B$ (i.e., same muscle set, h and L). Blue represents values higher than 100%, i.e., $\bar{\theta}_c$ was higher than the baseline. Similarly, we plotted the performance of the standard deviation of the cycling at the last cycle ($\sigma_{\dot{\theta}}$) in Fig. 7. Blue represents lower $\sigma_{\dot{\theta}}$ compared to the baseline, i.e., less variability in cadence. In this analysis, we consider a more efficient cycling, a cycling with higher $\bar{\theta}_c$ and lower $\sigma_{\dot{\theta}}$, i.e., color map in blue on both figures. We inverted the gradient in Figs. 6 and 7 because we chose to represent the improvements as blue. For Fig. 6, the improvement is higher %, and for Fig. 7 is a lower %.

From Fig. 6(a), we observed that most negative torques lead to higher average cycling speed ($\bar{\theta}_c$), and most negative torques lead to higher average cycling speed compared to simulations without the passive orthoses. We also observed in Fig. 7(a) that positive torques lead to lower standard deviations ($\sigma_{\dot{\theta}}$). Henceforth, simulations in which the spring aids knee extension presented higher cadences with lower variation in the cycle.

Furthermore, when we observe the performance related to the cadence (Fig. 6(b)), we noticed that most higher

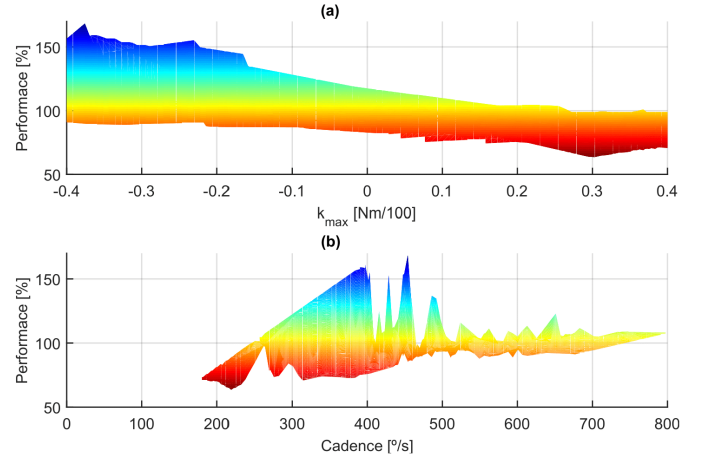


Fig. 6. Simulation performance of the average cycling ($\bar{\theta}_c$) for all k_{max} ($k_{max} = \tau_{max}/100$). Blue represents higher $\bar{\theta}_c$ (compared to the previously simulated baseline that corresponds to the same muscle set, h and L), and red, lower $\bar{\theta}_c$. (a) Detail of performance and k_{max} . (b) Detail of performance and $\bar{\theta}_c$.

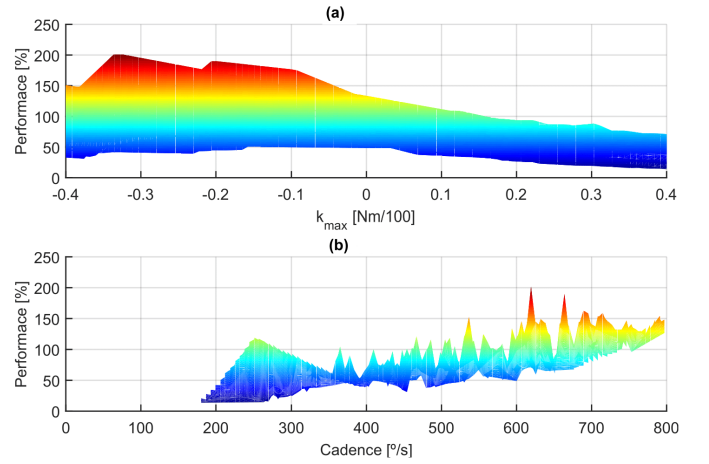


Fig. 7. Simulation performance of the standard deviation of the cycling cadence ($\sigma_{\dot{\theta}}$) for all k_{max} ($k_{max} = \tau_{max}/100$). Red represents higher $\sigma_{\dot{\theta}}$ (compared to the previously simulated baseline that corresponds to the same muscle set, h and L), and blue, lower $\sigma_{\dot{\theta}}$. (a) Detail of performance and k_{max} . (b) Detail of performance and $\sigma_{\dot{\theta}}$.

performances happened between $300^\circ/s$ and $400^\circ/s$, indicating that the cadence also influences the performance, not only the spring.

2) *Actuation Range*: Likewise, Fig. 8 and 9 show the plots with the results when we ranged θ_s . For most ranges, the cadence and standard deviation performances increased, with a slightly higher rate for lower starting ranges, i.e., spring actuates longer (Fig. 8(a)). In these figures, we also see that the cadence seems to influence on the cadence performance, between $350^\circ/s$ and $450^\circ/s$, we see higher cadences (Fig. 8(b)) and lower standard deviations (Fig. 9(b)).

B. Experiments With the Volunteer

Considering the simulation results, we customized a pair of passive orthoses with springs of $0.071\text{Nm}/^\circ$ stiffness around each knee joint, in which the spring stores energy from

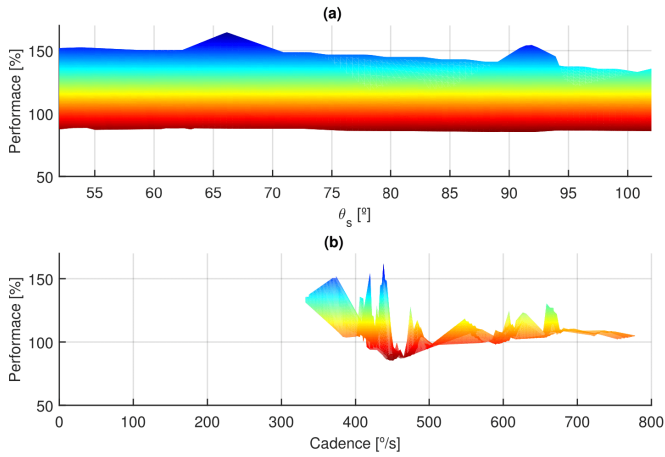


Fig. 8. Simulation performance of the average cycling ($\bar{\theta}_c$) for all θ_s . Blue represents higher $\bar{\theta}_c$ (compared to the previously simulated baseline that corresponds to the same muscle set, h and L), and red, lower $\bar{\theta}_c$. (a) Detail of performance and θ_s ($\Delta\theta = \theta_i - \theta_s$). (b) Detail of performance and $\bar{\theta}_c$.

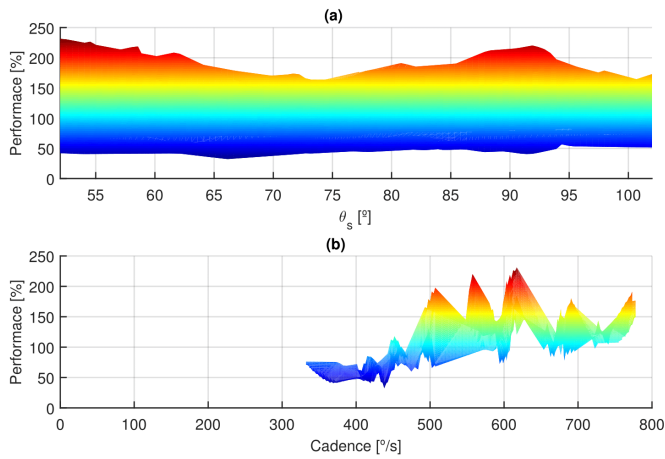


Fig. 9. Simulation performance of the standard deviation of the cycling cadence ($\sigma_{\dot{\theta}}$) for all θ_s . Red represents higher $\sigma_{\dot{\theta}}$ (compared to the previously simulated baseline that corresponds to the same muscle set, h and L), and blue, lower $\sigma_{\dot{\theta}}$. (a) Detail of performance and θ_s ($\Delta\theta = \theta_i - \theta_s$). (b) Detail of performance and $\sigma_{\dot{\theta}}$.

knee flexion to release it as the knee extends. As the spring actuation range is 42° , $\tau_{max} = 2.9$ Nm. For simulations, we selected the same value as [17] ($k = 0.25$ Nm/ $^\circ$; range = 55° ; $\tau_{max} = 13.75$ Nm). They worked with cyclists without SCI and achieved powers above 200 W. However, in our experiments, the volunteer with SCI could not achieve powers higher than 12 W. We tested with other springs ($\tau_{max} = 5.9$ Nm, 11.8 Nm, and 17.6 Nm), but he could not cycle as FES power is much lower.

During the experimental setup, the volunteer cycled with and without springs in two days (the first day lasted approximately 30 minutes, and the second, 40 minutes). We monitored the participant, and he did not experience any motor conditions that affect his movement (e.g., spasticity). The cyclist has a history of fracture in the left knee, which causes a loss of strength in this leg. However, during the experiments, it is unnoticeable as both legs provide enough energy for cycling.

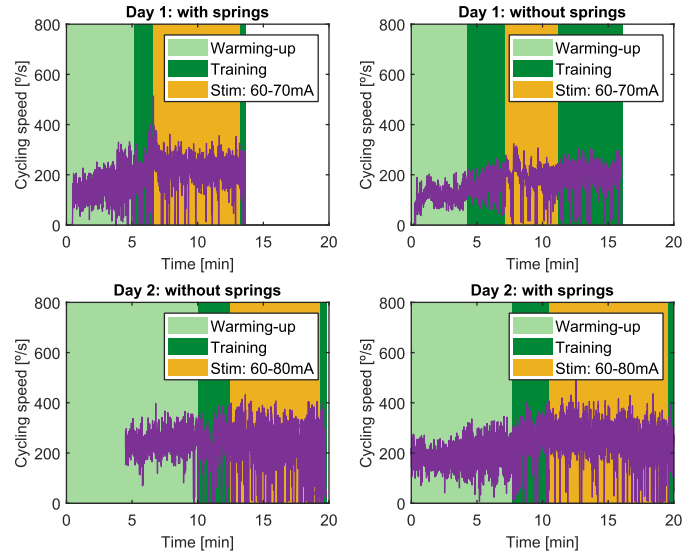


Fig. 10. Cycling speed during both days of experiments. At each day, the volunteer cycled with and without the springs. The exercise divides into warming-up (light green) and training phases (dark green and yellow). On the first day, we compare cycling speed between 60mA and 70mA, and on the second day, between 60mA and 80mA (yellow).

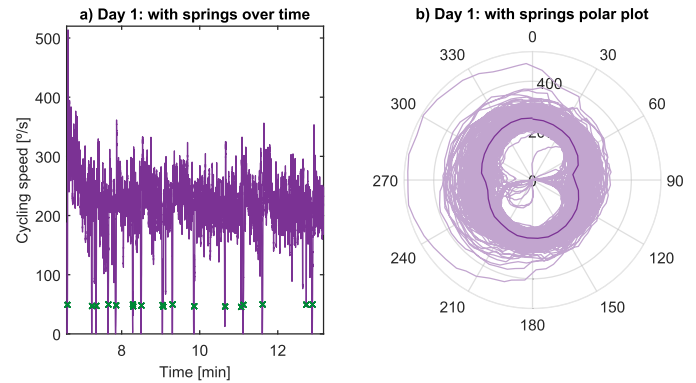


Fig. 11. Cycling speed during first day of training with the springs for 60mA and 70mA of stimulation current (around 6min). (a) Cycling speed over time. (b) Cycling speed over the position of the crankset (left crankset forming 90° with the ground represents the zero position).

The duration of experiments differs each day due to the volunteer state of tiredness. Fig. 10 shows the crankset cycling speed during the entire protocol. We recorded the warming-up (light green) and training phases (dark green and yellow) from exercises after applying a moving average filter with window size 30. In this work, we observed between 60 mA and 70 mA on the first day and between 60 mA and 80 mA on the second day (yellow marks).

On the first day, we observed the cycling movement from Fig. 11 and 12, with and without springs respectively (stimulation amplitude between 60 mA and 70 mA). In each figure, we represent the cycling speed through time (Fig. 11a and 12a) and over the crankset position (Fig. 11b and 12b). Table II shows the average cycling speed and standard deviation during the section between 60 mA and 70 mA. We may observe an increase of 17% of average cycling speed when we attached the springs.

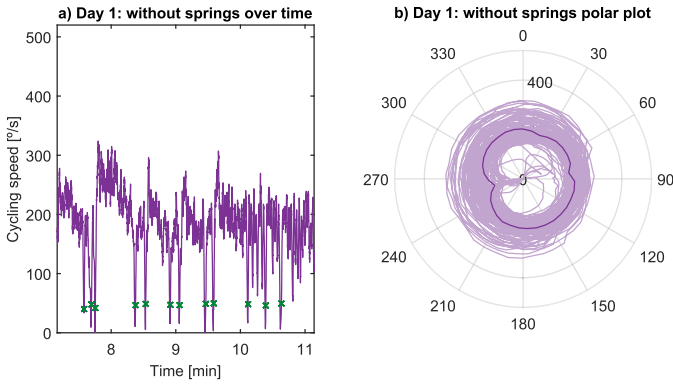


Fig. 12. Cycling speed during first day of training without the springs for 60 mA and 70 mA of stimulation current (around 4 minutes). (a) Cycling speed over time. (b) Cycling speed over the position of the crankset (left crankset forming 90° with the ground represents the zero position).

TABLE II

AVERAGE CYCLING SPEED [$^\circ/\text{s}$] FOR THE TRAINING SECTION BETWEEN 60mA AND 70mA OF STIMULATION CURRENT FOR THE FIRST DAY OF TRAINING

	With springs	Without the springs
Average cycling speed	215.14	183.88
Standard deviation	60.30	57.08

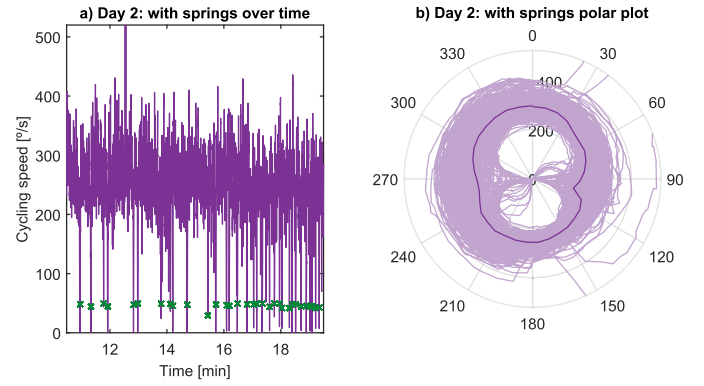


Fig. 14. Cycling speed during second day of training with the springs for 60 mA and 80 mA of stimulation current (around 9 minutes). (a) Cycling speed over time. (b) Cycling speed over the position of the crankset (left crankset forming 90° with the ground represents the zero position).

TABLE III

AVERAGE CYCLING SPEED [$^\circ/\text{s}$] FOR THE TRAINING SECTION BETWEEN 60mA AND 80mA OF STIMULATION CURRENT FOR THE SECOND DAY OF TRAINING

	With springs	Without the springs
Average cycling speed	245.35	220.42
Standard deviation	74.81	92.39

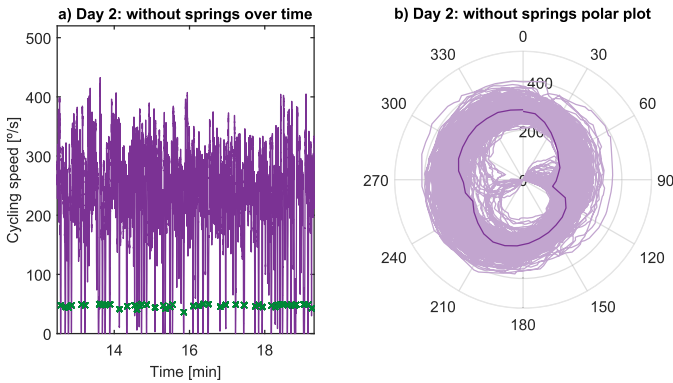


Fig. 13. Cycling speed during second day of training without the springs for 60 mA and 80 mA of stimulation current (around 7min). (a) Cycling speed over time. (b) Cycling speed over the position of the crankset (left crankset forming 90° with the ground represents the zero position).

On the second day, we observed the cycling movement from Fig. 14 and 13, with and without springs respectively (stimulation amplitude between between 60 mA and 80 mA). Table III shows the average cycling speed and standard deviation during the section between 60 mA and 80 mA. We may observe an increase of 11% of average cycling speed when we attached the springs.

In the polar plots (Fig. 11 to 14), we represent the zero position when the left crankset forms 90° with the ground, i.e., the left foot is at the highest point over the ground. We may observe points where the cycling speed almost reaches zero (usually over 90° and 270°). When we only excite the quadriceps, those are the dead points of the EMA trike, as observed in the profile in Fig. 4. There is no stimulation on the muscles at these points; if the inertia of movement did not guarantee enough force to pass those points, it usually

stops. We marked those points also as green crosses over Fig. 11(a) to 14(a). Observing the green crosses, we presume that the abrupt stop during pedaling does not affect the average speed's computation. On both days, there were a consistent number of times that the rider stops.

IV. DISCUSSION

Previous work assumed that elastic elements could positively affect cycling based on the simulations of just a few sets of spring parameters [15]. In this work, the detailed FES cycling simulation environment allowed further investigation of passive knee orthoses parameters, in which we found that the spring parameters τ_{max} and θ_s may positively influence cadence. Simulating a wide range of τ_{max} (Fig. 6 and 7), we observed the positive torques (i.e., torques that release energy during leg extension) usually lead to similar or lower cadences and negative torques lead to higher cadences. These results imply that, in simulations, we may achieve higher cadence with the same muscle excitation h with springs in comparison to a scenario with no springs.

In simulations ranging θ_s (Fig. 8 and 9), we observed that most θ_s increased the average cycling speed, with a slightly higher performance for lower θ_s (i.e., higher ranges). However, lower θ_s also increased standard deviation errors. One hypothesis is that the performance also relates to the synchronization between the spring actuation range and the knee dynamics. Studies considering the biomechanics of cycling, similar to [22], [23] could provide a better perception of the system, considering the synchronization between the starting angle, the spring actuation, the cycling stroke, and the muscles.

From Fig. 6(c) and 7(c), we also observed some relation between cadence and the improvement in cycling (between $300^\circ/\text{s}$ and $400^\circ/\text{s}$). These results also suggest that the model

biomechanics influences cycling performance. For further conclusions, future work should also analyze the model dynamics in synchronization with the spring range of actuation. Additionally, we should again recognize some modeling simplifications (e.g., locked pelvis), limiting the direct comparison between model and experiment results.

Based on the consistency of the results from Section III-A, we built a pair of passive knee orthoses to test this effect during FES cycling exercises performed by a volunteer with SCI. Past works have previously modeled and implemented similar concepts for cycling [13], [15]–[17], yet, this is the first time that an individual cycled with passive knee orthoses using FES. While using the passive element, the cadence increased in both days of training (Tables II and III). Results from Fig. 11 and 14 suggest that the use of springs aids overcoming dead zones during the intracycle cadence.

During the experimental setup, the positioning of the user on the bicycle took longer than usual with the passive knee orthoses, as it added weight to the system, making one leg out of the sagittal plane. This additional setup, i.e., putting the passive orthoses and repositioning on the tricycle, took about 15 minutes. The orthoses made it harder for him to start cycling than training without any orthoses. This prototype forces the skin and pushes the electrodes, reducing the contact area, consequently reducing its effectiveness. Furthermore, although the spring provides energy to the system half the time, the other half acts as an extra load, and, at lower cadences, the lack of inertia could obstruct the entire cycling. Our solution was manually interfering in the experiments during the warming-up phase (represented as the wheel-acceleration system in simulations). Optimal final solutions for FES cycling usually do not contain these external interferences, so the user has more autonomy when using the system [3]–[5].

The spring positioning to guarantee synchronization (timing) is one of the biggest challenges when moving from a simulation environment to an experimental environment. After we position the orthoses, the spring holders keep the spring range constant because the devices are rigid. However, the passive orthoses positioning over the cyclist may differ between experiments (more notably for flexible orthoses). There are slightly different ranges between each training. Nevertheless, real FES cycling experiments already present this type of discrepancy due to, e.g., electrode positioning and volunteer sitting position. Therefore, we should be aware that adding another element to a rehabilitation system increases the variety of the results and, sometimes, unpredictability.

Further studies should recognize if these external aids (e.g., manually assisting the rider to start cycling) are unavoidable and if this could limit the use of passive orthoses in rehabilitation scenarios. Moreover, in the future, it is also possible to create an improved and more stiff prototype, as developed in [17], which could better guarantee the correct positioning of the springs around the knee, reduce its weight and, consequently decrease the time of setup and the effect over the electrodes.

As we see a positive cadence change with the same setup, there is evidence that in a scenario with cadence control

(e.g., proportional integral derivative controller), passive orthoses could lead to a decrement in muscle excitation through stimulation. Increasing the stimulation current intensity remains the most effective way to alleviate fatigue [8]. Therefore, lower excitation could delay or even prevent fatigue, one of the central challenges for FES cycling performance. Results from Figure 10 indicate that, when we attach the springs, the exercise duration is more prolonged (considering the same stimulation range level, yellow blocks). Future efforts should test these theories in future experiments monitoring fatigue with more subjects and more spring sets.

Moreover, it is still unclear how the metabolic cost and fatigue decrease using passive orthoses in cycling. Studies investigating FES systems optimization often report power values instead of cadence, which could advance the discussion about passive orthoses use. In this work, we set up a similar system used for groups at the Cyathlon 2016 and 2020 competitions [5], [10]. Therefore, it is easier for other groups to repeat the methods and compare results, such as cadence performance.

Furthermore, future works should also evaluate scenarios with disturbance loads [24], [25]. The lack of optimal control strategies still limits FES cycling from achieving its full potential, and new experiments are essential to provide more robust conclusions and how to evaluate the gain obtained in the rehabilitation process.

V. CONCLUSION

In this work, we explored the use of passive elements in knee orthoses during FES cycling. First, we simulated a cycling model's dynamics with passive orthoses to find relations between parameters and performance. Simulations showed that passive orthoses increased cadence depending on the elastic spring constant. Then, based on these simulation results, we performed an initial experimental validation with one case study. The volunteer was not only able to cycle with the orthotic device, but his speed increased in 17% on the first day, and 11% on the second day of experiments, when compared to cycling sets without the springs. This increase potentially allows longer rehabilitation sessions or FES cycling, facilitating physical exercises for people with SCI.

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