Simulation of the assistance of passive knee orthoses in FES cycling*

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Abstract— **Although advances in technology promoted new physiotherapy approaches, there is still an urge for equipment and techniques to improve the quality of patients lives with motor disabilities. Functional electrical stimulation cycling (FES cycling) is an example of this type of technology, in which the control of stimulation parameters enables a spinal cord injured person to ride a bicycle. In this work, we aim to investigate the use of passive knee orthoses for FES cycling assistance. Hence, we compared the cycling cadence and quadriceps excitation using an FES cycling simulation platform for different spring torques and ranges. In this paper, we obtained spring parameters that increased cycling cadence by 10.60% while decreasing by 7.33% the quadriceps activity, which indicates that this type of passive orthosis may diminish fatigue caused by FES.**

I. INTRODUCTION

Spinal cord injury (SCI) may cause impairments on motor functions. These impairments not only affect mobility but also make these individuals more susceptible to other health complications, such as osteoporosis, higher incidences of fractures and instabilities of the cardiovascular, thermoregulatory and broncho-pulmonary systems. Through repetitive and intense exercises in physiotherapy, it is possible to increase motor function or diminish some of these comorbidities [1]. As disabilities have devastating effects on quality of life, there is a demand to enhance rehabilitation, so that patients achieve their best possible functional outcome at the smallest period. In general, functional electrical stimulation (FES) and orthoses assistance increase these rehabilitation benefits [1].

FES stands for a known rehabilitation technique for motor function improvement, in which the stimulation generates muscle contraction [2]. Further, FES may assist people with paraplegia in complex movements, such as FES cycling. This rehabilitation enhances muscle strength and cardiorespiratory fitness [3]. However, rapid muscle fatigue is a notable limitation for FES muscle contractions in individuals with SCI. Consequently, some adjustments aim to diminish this effect. The simple addition of passive orthoses may store energy (usually elastic with mechanical springs) to assist movement, reducing muscle fatigue effects and lowering the metabolic cost by providing a more natural and stable movement for rowing [4] or cycling [5].

To maintain the FES cycling cadence, [6] added a flywheel mechanism at the crank of the bicycle. The system engaged an electrical clutch that engaged/disengaged the flywheel with the crank while stimulating the quadriceps muscle group. As the flywheel absorbs excessive kinematic energy, the control engaged the spring to slow down motion. However, one of the primary challenges in FES control is to decrease the fast fatigue caused by the artificial stimulation of the muscles.

In an environment without FES, [7] introduced the concept of passive knee orthoses for cycling assistance, in which a spring stores energy from knee flexion to release it as the knee extends. They based the passive assistance on the unbalanced effort required from the quadriceps (knee extensor) and hamstrings (knee flexor) during the same cycling cadence [8]. Further, in [9], they performed tests with and without the knee orthoses with three able-bodied subjects. At the same cycling cadence, the passive knee orthoses decreased the quadriceps effort during cycling trials.

To our knowledge, no previous experiments used mechanical passive orthoses in a similar approach for FES cycling. Nevertheless, some authors have already designed passive knee orthoses to regulate FES gait [10]. Similar to [9], the preliminary experiments from [10] also decreased the quadriceps excitation, which consequently delays the fatigue.

The present work aims to model passive knee orthoses for FES cycling assistance, determining the spring parameters, and how the parameters relate to the average cycling cadence and the muscles excitation. As the accelerated fatigue generated by FES limits the duration of experiments, we employed a previously developed simulation environment for FES cycling [11]. Section II describes our methods. We adapted the model, adding passive orthoses to both knees (Section II-A). After determining the spring and cycling parameters, we conducted simulations on how the spring affects FES cycling (Section II-B). Subsequently, Section III presents the results of these simulations. The final discussion and conclusions are found in Section IV.

II. METHODS

We simulated the FES cycling environment using OpenSim [12] integrated with Matlab. OpenSim is an open source software for modeling and simulation of musculoskeletal systems. The software offers a forward dynamic simulation, in which precise muscle excitation patterns may be defined, simulating the effect of FES. In a previous model [11], we have already provided a platform for FES cycling with PID control for quadriceps, hamstrings and/or gluteus[1]. To model

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

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the passive orthoses, we adapted the framework (Section II-A) by adding passive knee orthoses and a wheel accelerating system (Section II-A.1).

A. Basic Framework

The model described in [11] is based on the EMA Trike [13]. Fig. 1 illustrates the model with its adaptations. The hips and knees of the model run freely, but, the pelvis and ankles are locked. The pedal holders immobilize the ankles and connect the feet to the pedals through a box in which the pedal accommodates. Consequently, the pedal holders transmit the forces to the pedal using contact geometries (physical shapes that allow collisions in OpenSim).

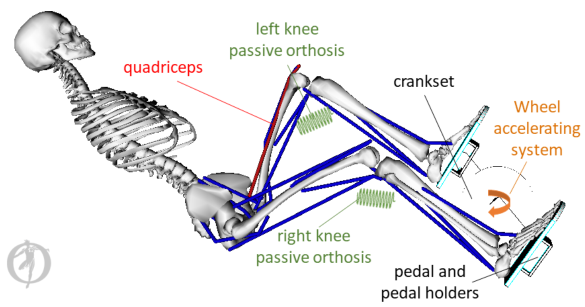

Fig. 1. Complete model description for FES cycling positioned similarly to the EMA Trike [13]. OpenSim represents excited muscles as red lines, and not excited muscles as blue lines. We adapted the previous system with left and right passive knee orthoses (represented in green) and a wheel accelerating system at the crankset (represented in orange). In this article, we excited only the quadriceps.

The excitation control incorporates predefined muscles range angles. We have previously defined these ranges to excite each muscle group in [11]. During one pedal stroke, quadriceps provide most torque for the pedal stroke though knee extension. Therefore, we choose to apply excitation only on the right and left quadriceps muscles (u_{exc}). In OpenSim, the interval of muscles excitation is $[0, 1]$, in which 1 is the maximum excitation force. For the passive knee orthoses analyses, we kept the quadriceps at the maximum. This response leads to a maximum cycling cadence for later comparison with and without the passive orthoses, i.e., we may recognize the interference essentially from the passive orthoses. Fig. 2 illustrates the quadriceps and the passive knee orthoses range angles.

1) Passive knee orthoses and wheel accelerating system: We modeled the passive orthoses based in [7]. The supporting knee torque (τ_{spr}) operates as a rotational spring

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

where K represents the spring stiffness, θ_j represents the knee joint angle, and θ_s the starting angle. Fig. 3 illustrates an example of a spring with $\theta_s = 67º$ and $K = 0.69\, Nm/º$. At this representation, 0º refers to the maximum knee extension, i.e., while the knee angle increases, the leg flexes, and while it decreases, the leg extends. Therefore, the spring may provide a maximum torque τ_{max}.

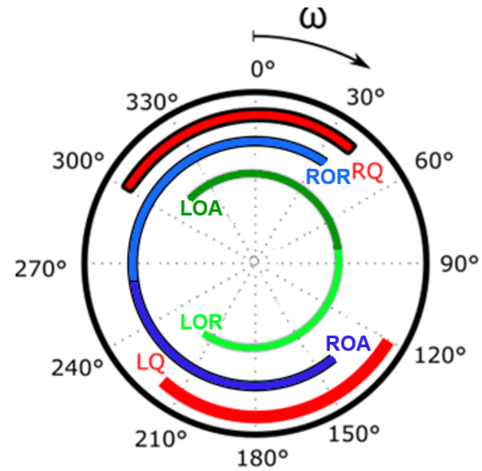

Fig. 2. Muscles and spring range angles for excitation during a pedal stroke. Right and left quadriceps (RQ and LQ) marked as red, right orthosis accumulating (ROA) and releasing (ROR) energy and left orthosis accumulating (LOA) and releasing (LOR) energy.

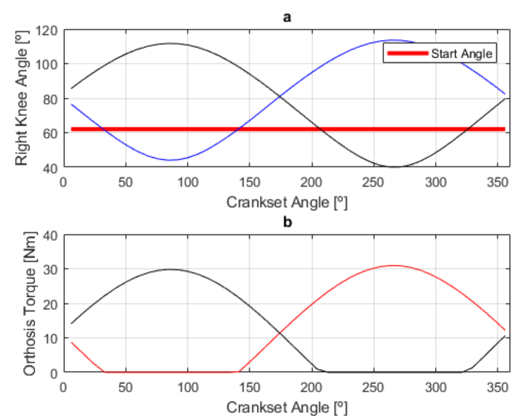

Fig. 3. Example of a spring with starting angle at $\theta_s = 67º$, and stiffness of $K = 0.69\, Nm/º$ during a complete crankset cycle. The crankset angle reference is related to the position of the right foot. (a) Right knee (blue line) and spring start angle (red line). (b) Passive orthosis torque (red).

Consequently, the spring 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. As the spring may significantly resist movement at some states, sometimes the model is unable to cycle without external aid. Hence, we modeled a wheel acceleration system that avoids this dependency at the beginning of cycling. After the cycle achieves the target crankset cadence $\dot{\theta}_t$, the system may keep cycling due to inertia and geometry conditions of the bicycle. Therefore, the acceleration system provides the torque

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

where $\dot{\theta}_c$ is the cycling cadence, calculated by the differentiation of the crankset angle during the time.

The passive orthosis has an inflection angle (θ_i), in which the spring stops accumulating energy and starts releasing it. For an easier mechanism prototyping in the future, this point should be at the same inflection angle of the knee (i.e., the knee shifts from extension to flexion, or vice versa).

B. Simulations to determine the spring parameters

After modeling the new FES system, we performed simulations to determine the spring parameters τ_{max} and θ_s, described in Sections II-B.1 to II-B.3 and summarized in Fig. 4. Our performance criteria for cycling were the average crankset cadence and total quadriceps excitation (i.e., the percentage of time that the control excited the quadriceps) at the last complete cycle. Hence, we expect to choose spring parameters that increase the average cadence of a complete cycle with a similar quadriceps excitation. In OpenSim, we fixed the control frequency at 50 Hz, thus at transitions between activating or deactivating quadriceps, the start of excitation may be delayed or advanced, changing the total quadriceps excitation.

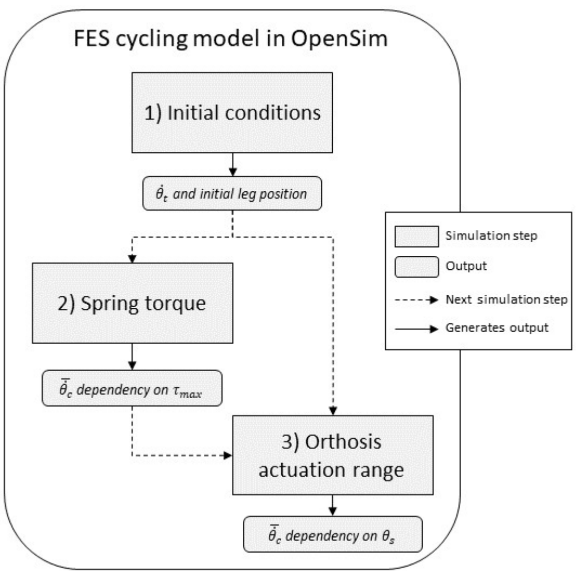

Fig. 4. Flow diagram of the simulations steps to determine the spring parameters (τ_{max} and θ_s) relation to the cycling cadence. We performed all simulations with the same FES cycling model in OpenSim.

1) Initial conditions: Before starting the simulations to determine the spring parameters, we performed some tests to determine the proper initial leg position and the target cadence $\dot{\theta}_t$ for the wheel accelerating system. We defined the initial leg position based on the position of the right foot on the crankset angle. We performed $5\,s$ simulations of 12 initial positions, from the crankset angles 0º to 165º in steps of 15º. As the leg positions in cycling are symmetric, the 180º to 360º variations are not necessary. We chose the initial position as the result that performed the highest average crankset cadence $\bar{\dot{\theta}}_c$ at the last complete cycle. We also chose $\dot{\theta}_t$ as this highest cadence value.

2) Spring torque: To decide if the spring should first release or first store energy, we performed $20\,s$ simulations with several springs that reach the maximum torque at the inflection angle defined in Section II-B.1. We ranged the maximum spring torque from $-50\,Nm$ to $50\,Nm$, in which the negative signal determines that the spring flexes the knee, and the positive signal determines that the spring extends it. A maximum torque higher than $|50|\,Nm$ could lead to spring constants higher than the prefabricated options. We selected the θ_s, so the spring flexes and extends for approximately a quarter of the cycle period.

3) Orthoses actuation range: We defined the actuation range $\Delta\theta$ as the angular range that the orthosis actuates from its minimum to its maximum torque value (i.e., $\Delta\theta = \theta_i - \theta_s$). Based on the torque results in Section II-B.2, we analyze the effect of the spring ranges. As the complete range of the knee trajectory is approximate 70º, we ranged $\Delta\theta$ from 5º to 65º, increasing 5º at each simulation.

III. RESULTS

The final average crankset cadence depends on the bicycle geometry and the quadriceps range. For our system, the 120º position generated the highest $\bar{\dot{\theta}}_c = 270.5$ °/s. Therefore, we defined this cadence as $\dot{\theta}_t$ for the wheel acceleration system.

Using the simulation results from the 120º initial position, we found two knee inflection angles, at 44º (extending to flexing) and 112º (flexing to extending). As at 112º the spring is closer to the right quadriceps excitation ($\theta_c = 280$º) we chose this inflection angle so that we could synchronize the energy released to the leg, as illustrated in Fig. 2.

Further, we found that with $\theta_s = 92$º, the spring acts during a quarter of the cycle period. Fig. 5 presents the average crankset cadence and the quadriceps excitation period of the last cycle. From these results, we found that positive torques (i.e., the torque that extends the leg) lead to higher cadence. The negative torques lead to similar or lower cadence. Further, the quadriceps excitation remained almost the same (28%), which implies that with the same muscle excitation, we may achieve higher cadence (i.e., possibly delaying muscle fatigue).

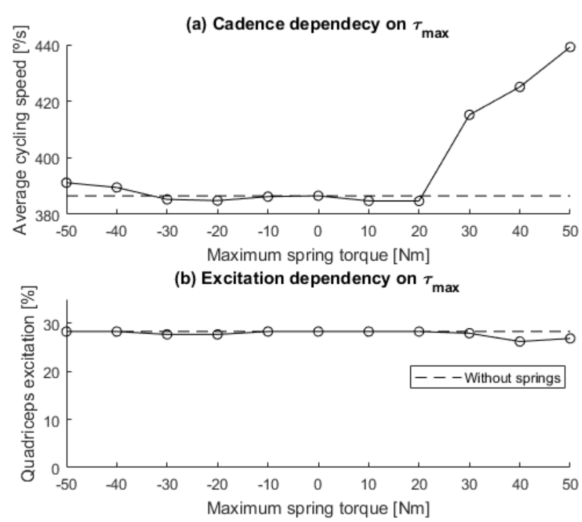

Fig. 5. Average cadence and quadriceps excitation after ranging the maximum spring torque. The dashed line represents the reference crankset cadence for a $20\,s$ simulation without the passive orthoses. (a) Average cadence dependency on maximum spring torque with a constant start angle. (b) Total time percentage of the right leg quadriceps excitation.

Almost all results for different τ_{max} led to a higher average cadence with similar excitation (around 28%), as shown in Fig. 6. The plots also indicate that lower θ_s (i.e., higher ranges) usually led to higher cadence, probably due to a more extended spring actuation, accumulating and releasing energy. Moreover, we may assume that there is a boundary for the τ_{max}, as $50\,Nm$ presented lower average cadence for lower θ_s. This restriction is due to the maximum torque that the quadriceps provide to the system, i.e., if the spring exceeds this torque, the model is unable to cycle.

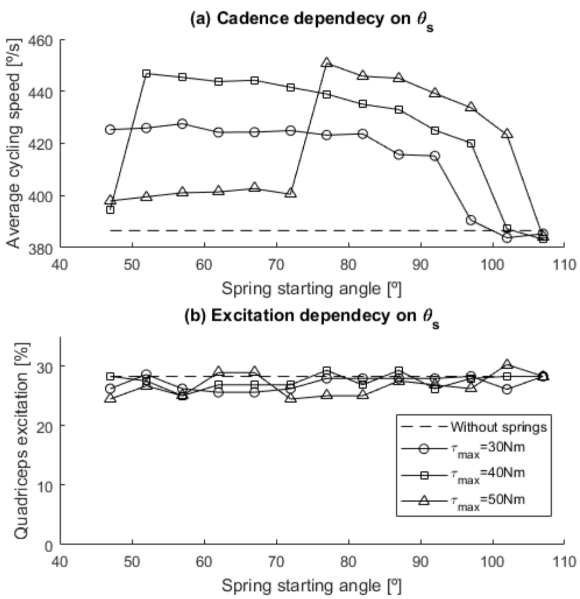

Fig. 6. Average cadence and quadriceps excitation ranging the spring range angle for $\tau_{max} = \{30, 40, 50\}$ Nm. The dashed line represents the reference crankset cadence for a $20\,s$ simulation without the passive orthoses. (a) Average cadence dependency on θ_s for $\tau_{max} = \{30, 40, 50\}$ Nm. (b) Total time percentage of the right leg quadriceps excitation.

Moreover, higher cadence results not necessarily lead to tolerable elastic spring constant values. We intend to keep this constant lower than $0.55\,Nm/^{\circ}$. Some results meet this requirement, e.g., $\tau_{max} = 30\,Nm$ and $\theta_s = 57^{\circ}$ ($K = 0.54\,N/m$). These spring parameters increased the cycling cadence by 10.60% with 7.33% less quadriceps activity.

IV. CONCLUSIONS

The OpenSim FES cycling environment allowed the investigation of passive knee orthoses parameters. We found the knee inflection and the spring parameters θ_s and τ_{max}. With similar quadriceps excitation, we observed that the spring torque that extends the leg lead to higher cycling cadence, which implies that the system overcomes dead zones during the intracycle cadence.

Although we were able to define parameters for manufacturing, the presented paper still lacks considerations about friction losses and controllers design (e.g., PID control adapted to the spring model). This study also misses a force analysis on the knee and the pedals. As we intend to use the passive orthoses to assist FES cycling for SCI individuals, the system must guarantee that it does not cause injuries.

Besides, it is still unclear how the metabolic cost and fatigue decrease using passive orthoses in cycling. Therefore, with these simulation results, it is challenging to evaluate the gain obtained in the rehabilitation process yet. Further, we should consider that it may be challenging to start real mobile cycling with these passive orthoses without a motor.

In this work, we noticed evidence that the use of passive orthoses may increase average cadences for FES cycling using the same (or even lower) quadriceps excitation, similar to [9]. In the future, we intend to expand the simulations to consider friction losses, controllers, and muscle perturbations (e.g., fatigue or spasms, common FES difficulties) before manufacturing and testing the spring with able-bodied and SCI subjects.

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