

Quantitative evaluation of stimulation patterns for FES cycling

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Abstract

The purpose of this work is to evaluate the stimulation patterns used to produce cycling by functional electrical stimulation. Specific experiments were carried out on two SCI patients using a motor to keep a constant cadence. We stimulated one muscle at a time for the whole cycle and measured the torque produced at the crank. We evaluated the contributions of the muscles to the movement to identify the ranges of useful functionality, i.e., the positive torque. The peak torques were also identified. This way we propose an experimental-based method to distinguish when the stimulation of each muscle facilitates or hinders the cycling movement and at what angles the muscles provide the most functionality. The results were compared to currently implemented patterns and it was shown that these patterns are perhaps not making the most use of the muscle-crank angle relationship.

1. INTRODUCTION

The use of functional electrical stimulation (FES) for cycling is now becoming an established choice for the rehabilitation of SCI patients. The maximum power generation and the muscle stamina are, however, still major cause of concern and a barrier to recreational mobile cycling. Due to these short comings it is therefore of increased importance that the cycling stimulation pattern is optimized to maximize performance.

There have been several approaches in the design of stimulation patterns for FES cycling. A paper by Perkins et al. [1] demonstrated a direct approach by testing static forces produced at set angles around the “likely” positions for each muscle group. This approach applied results from

statically measured forces to a dynamic system. The rate of change of length of a muscle will alter the force it produces. It is also a slow manual task to take incremental static measurements for each muscle group - even using a 45 degree increment. Chen et al. [2] proposed a stimulation pattern designed via a mathematical consideration of the geometry and limb weight. However this does not take into consideration the ability of the subject’s atrophied muscles. A muscle may be stronger in one orientation than in another. A different approach by Petrofsky [3] used EMG measurements from able-bodied cyclists to design the stimulation patterns. This assumes that electrically stimulated muscle produces a similar reaction to voluntary contractions.

It is hard to identify for each muscle an optimal length for force generation and optimal joint angles for force transfer. It is therefore the aim of this paper to examine the resulting torques produced from stimulation of the individual muscle groups used in FES cycling. By determining the crossover angles from assisting to retardation of the cycling motion and by identifying peaks in the torque profile a more appropriate evaluation of the stimulation pattern can be designed.

2. METHODS

The experimental setup consisted of a motorized recumbent tricycle modified for FES cycling with a shaft encoder to measure the crank angle and with orthoses mounted on the pedals to fix the ankles. An SRM sensor was used to measure the torque between the crank shaft and the cycle chain. The tricycle was mounted on a home trainer and the system was connected to a PC

running Matlab Simulink for data acquisition and control over the motor and the stimulation pulse width. The motor is controlled via a cadence controller designed using a polynomial pole placement approach from an identified model. The identified model was a second order ARX model of the motorized trike with the test subject seated in the trike (no stimulation). In this work an 8 channel stimulator (Stanmore Stimulator®) was used and quadriceps, hamstrings, gluteal and calf muscle groups were stimulated. Two complete paraplegic FES cyclists, a female and a male with a T9 and T10 lesions were asked to take part in the experimental sessions. Once the test subject was seated on the trike and the stimulation electrodes were correctly attached, the test began. The cadence controller maintained the pedalling cadence. During the initial and final 20 seconds the subject was not stimulated at all and cycled passively. In the intermediate part of the test, each muscle group was individually stimulated constantly for 20 seconds each. This test was repeated with cadences of 10 revolution per minute (rpm), 30 rpm and 50 rpm and with stimulation pulse widths of 300 μ s and 500 μ s. The stimulation currents were selected depending on the test subject, at a value that produced a tetanic contraction using a pulse width of 300 μ s. The stimulation frequency was set at 20Hz and data was acquired every 10ms.

3. RESULTS

Figure 1 shows an example of the results for the different muscles of the right leg, in the trial with a stimulation level of 300 μ s at a cadence of 30 rpm. The average torque measured during passive cycling (first and final 20 seconds) was subtracted from the torque measured with the stimulation for each single muscle. Analysing this difference of torque it was possible to neglect the contribution due to the inertia and the weight of the legs. This way torque differences greater than 0 were assisting the cycling motion while torques below 0 were resisting the cycling motion.

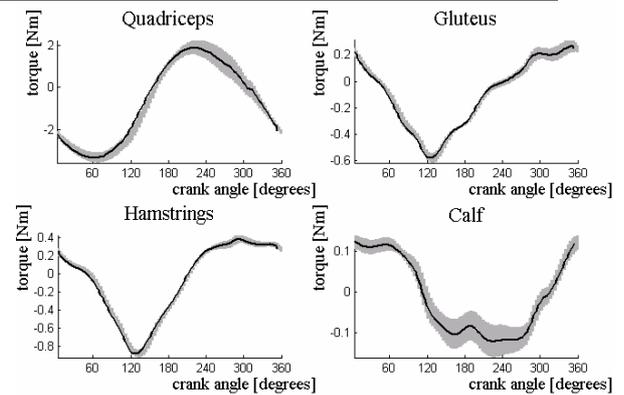


Figure 1: Difference between the torque produced during the stimulation test and the one obtained during the passive cycling against the crank angle. Results are reported in terms of mean value and standard deviation between all the cycles of the trial at a cadence of 30 rpm and with a pulse width of 300 μ s.

The start and end values of the positive torque and the start and end values of the maximal positive torque (max -10%) were averaged over all the trials (different pulse widths and cadences and for both patients). These are presented, with the standard deviations of the start and end values, for each muscle group in figure 2.

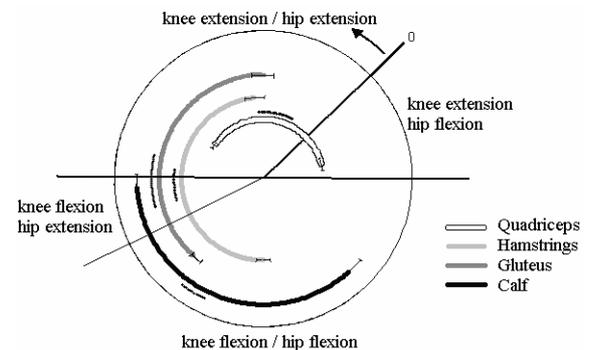


Figure 2: Limits of functionality of the different muscles. The phase in which each muscle gave its maximum contribution is reported with black lines. The crank reference is indicated with 0 which is the point of maximum hip flexion.

For a full understanding of the functionality of the different muscle groups, the different phases of the movement in terms of flexion and extension of the hip and the knee have also been included in figure 2. As figure 2 shows, responses were quite repeatable. Start and end variation over all conditions (3 speeds, 2 pulse widths) and for both subjects were small, about a maximum of 10 degrees.

From this representation the main action of the biarticular muscles can be understood. The quadriceps muscles acted mostly as knee extensors while the hamstrings were hip extensors at the start and knee flexors at the end of their action. Calf muscle action was limited to knee flexion because the orthoses fixed the ankle angle. Surprisingly, positive gluteus action extended into the third phase, which is not hip extension.

Figure 3 shows two stimulation patterns already in the literature [1], [4] compared with our calculated limits.

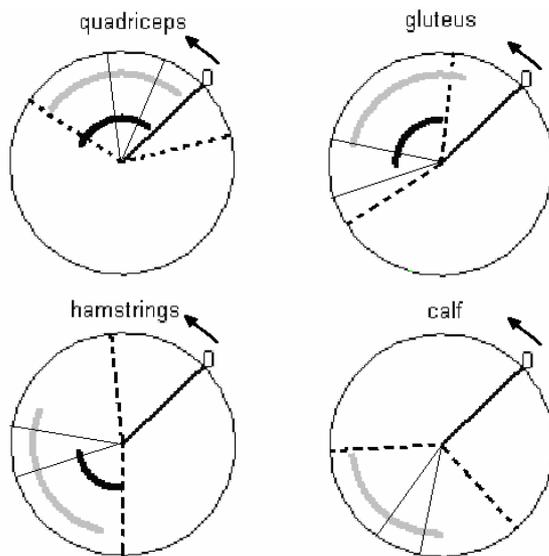


Figure 3: Two stimulation strategies are reported: grey lines showed the strategy used by Perkins [1] (Lit1), the thick black lines the Hunt one [4] (Lit2). Limits of functionality are shown in dashed lines. Thin black lines show maximal torque.

The Lit2 stimulation of the quadriceps switched off a bit too late overstepping the end of the identified range of functionality while Lit1 was within this range. Both the strategies included the maximal activation phase. Both Lit1 and Lit2 were inside the limits of functionality for the hamstrings but Lit2 did not stimulate this muscle group in the phase of maximal contribution. For gluteus stimulation Lit1 started a little bit in advance and neither exploited the entire phase of maximal contribution for this muscle group. The calf is used only by Lit1 and is stimulated inside the limits and over the maximal phase of functionality.

4. DISCUSSION AND CONCLUSIONS

Several ways have been used to decide how to stimulate legs for cycling, however, for people with spinal cord injury, each of their leg muscles is in some state of de/retraining. Also, the method of recruitment means that not all the target muscle groups may be stimulated. The situation is therefore complicated and these methods must be used with caution. We think that an experimental method, that allows the stimulation pattern to be set according to the actual stimulation responses of each user, will be valuable. The method described in this paper is quicker than the Perkins static procedure [1]. The key innovation is to use the motor to maintain the speed, thus removing one variable, and making the tests practicable. The simplest algorithm for determining the switching angles is to use the zero-crossing angles (Figure 2). However, this is unlikely to be optimal for normal cycling when, presumably, the current need for power is in conflict with endurance. This, and the time advance needed to allow for the delay between stimulation and muscle response, is the focus of our continuing studies.

References

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