Proceedings Paper:

Abdulla, S.C., Sayidmarie, O.K., Tokhi, M.O. et al. (1 more author) (2012) Modeling and control of a novel FES driven assisted cycling mechanism. In: Methods and Models in Automation and Robotics (MMAR), 17th International Conference on. 17th International Conference on Methods and Models in Automation and Robotics (MMAR), 2012, 27-30 Aug 2012, Miedzyzdroje, Poland. IEEE , pp. 464-469. ISBN 978-1-4673-2124-2

https://doi.org/10.1109/MMAR.2012.6347842

Modeling and Control of a Novel FES Driven Assisted Cycling Mechanism

Sh. Ch. Abdulla, O. Sayidmarie, S. Gharooni, O. Tokhi
Department of Automatic Control and Systems Engineering
University of Sheffield
Sheffield, UK
cop11sca@sheffield.ac.uk

***Abstract*—Functional Electrical Stimulation (FES) driven cycling using single muscle group, the quadriceps, is achieved using PID controller with a novel assisting mechanism represented by a flywheel with an electrical clutch. This mechanism is useful for disabled individuals whose muscles are weak and unable to push a fix-geared flywheel. The flywheel is engaged and disengaged by the clutch when necessary to assist or retard the cycling without imposing additional load on the person`s leg muscle. A comparison between this new mechanism and a previously proposed method show positive results of the new mechanism towards reducing the number of muscle stimulation which leads to delay muscle fatigue and consequently promoting prolonged FES driven cycling for paralyzed people.**

I. INTRODUCTION

The use of electrical signals to restore the function of paralyzed muscles is called functional electrical stimulation (FES). Closed-loop control can be used to coordinate FES-based movement in lower limbs to perform pedaling as training for individuals with spinal cord injury (SCI). FES cycling training can stop denervation atrophy of skeletal muscle, improve the cardiovascular fitness, increase muscle bulk and raise blood circulation in lower limbs, as well as offer psychological benefits [1-3]. Muscle stimulation can be done either by using implanted electrodes [4] or by surface electrodes mounted on the skin.

Several ergometers for FES-cycling training are commercially available such as Monark, Regys, Ergys. One of the handicaps that affect the performance and the smoothness of FES driven cycling is the dead points, i.e., the two points on the pedaling cycle at which it is difficult to produce sufficient crank torque. By means of inertia and a complex interplay of muscle actions, healthy people can overcome these points, while it`s difficult to produce such muscle actions by means of FES because the muscles involved are too deep to be stimulated by surface electrodes [5]. Previous studies have shown several attempts to overcome the dead points and produce smooth (i.e. with no abrupt change in crank velocity) FES cycling training for paraplegics. These have included improving the ergometer`s mechanical design, using different assisting mechanisms, and stimulating various combination of lower limbs` muscle groups [6-12].

Reference [6] developed a paracycle equipped with an auxiliary motor to assist the FES driven leg movement to pass the dead spots or to retard the cycling and impose braking if required. Three muscle groups were stimulated, namely quadriceps, hamstrings and gastrocnemius. Reference [7] designed a three-wheel mobile ergometer of two seats. One seat for a paraplegic and the other was to be used by a healthy person to assist the cycling and return home in case muscle fatigue occurred. Reference [8] developed an exercise tricycle equipped with an assisting auxiliary motor and used stimulation of femoral muscles. A tri-cycle of four-bar linkage assisting mechanism was suggested in [9] with a drawback of disabling the freewheeling of the tricycle. Reference [10] controlled the leg power and cycling cadence simultaneously with the assist of a motor and produced stimulation patterns to stimulate the quadriceps, hamstring, and gluteus muscle groups. An isokinetic leg exercise cycling ergometer to work with different speeds was developed in [11] making use of a motor and stimulated the quadriceps, hamstrings, and gluteus muscles.

One of the main limitations of the previously mentioned approaches is the use of a motor to assist the cycling. An electrical motor can be considered as a disadvantage, especially with mobile ergometers for locomotion, as it consumes energy and thus can only be used for limited distances. Moreover, a motorized system can pose potential risk of injury to individuals with SCI in case of high torque imposed on the paralyzed limbs [18]. Another limitation is the long preparation time for FES cycling due to the time consuming procedure of locating the electrodes on several muscle groups, as well as the experimentally adjustments of the stimulation parameters that might be uncomfortable for the disabled [12].

To overcome the above limitations, reference [12] showed the ability to perform coordinated and smooth FES cycling by stimulating the quadriceps muscle group with the aid of a passive energy storage device represented by a spring cable.

The only limitation of the approach proposed in [12] is that the spring, which is to take its energy from the leg during pushing phase and release its energy during the assist phase, adds additional load on the quadriceps, and thus the stimulation current on the muscle has to be increased to push the load and produce movement, consequently muscle fatigue would occur early.

Many studies have shown the use of flywheel in FES ergometers to provide resistive loads for individuals with strong muscles who are able to pedal under loads. The flywheel provides smoothness to the cycling and helps pass the dead points without jerking. Although the flywheel, as an energy storage device, is effective in assisting the cycling and reduces the energy expenditure [13], the drawback of using fix-geared flywheel is that untrained individuals` muscles are unable to generate sufficient force to give the first push to the flywheel.

In this study, a novel assisting mechanism is proposed that utilizes the advantages of using the flywheel (as a passive energy storage device) with an electrical clutch, and stimulating the quadriceps muscle group. The electrical clutch is used to perform the job of engaging and disengaging the flywheel to assist or resist the cycling when necessary. During FES cycling, if the leg`s speed exceeds the desired speed, the clutch will engage the flywheel with the shaft of the bicycle to absorb the excess in the energy produced by the leg, thus the flywheel is charged with kinetic energy and starts to rotate according to the amount of energy it has absorbed. When the leg is slower than the desired speed, the flywheel will be engaged to push and assist the leg movement and consequently the flywheel will discharge its energy. This mechanism is promoting delayed fatigue and consequently prolonged FES driven cycling.

II. System Description

A. Humanoid and Bicycle Model

Humanoid model was developed using Visual Nastran 4D (Vn4D) software with standard anthropometric human dimensions introduced by [14]. The length and mass of each body segment is expressed according to the overall height and weight of the humanoid model. The humanoid developed in this work is based on a human body of height 1.80m and 70kg in weight. The density and the centre of mass of each segment were obtained from the anthropometric data provided in [14]. The density of each segment was used to obtain the volume and consequently determine the segment width.

The bicycle model was also developed using Visual Nastran 4D (Vn4D) software. The dimensions of the bicycle (Pedal: 0.13 x 0.08 x 0.02m; Crankarm: 0.01 x 0.14 x 0.02m; Shaft: 0.01 x 0.15m) were obtained from a real cycling ergometer available in the laboratory of ACSE department, University of Sheffield.

A standard ballbearings friction with rotational coefficient (0.0015) and effective radius (0.01m) was added to the model to simulate a more realistic system and obtain reasonable results. A standard flywheel of 0.2m radius and 3.5kg weight was added to the bicycle with an on/off gear, to simulate the behaviour of an electrical clutch, to engage and disengage the flywheel with the shaft of the bicycle when required. The complete humanoid-bicycle model using Vn4D software is shown in Fig.1.

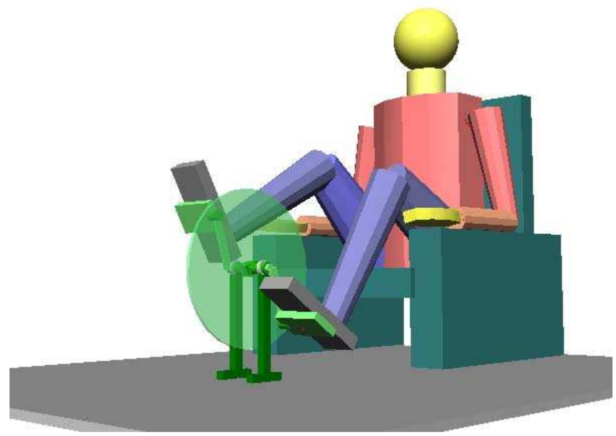

Figure 1. *Humanoid-bicycle with flywheel model*

B. Muscle model

Several attempts of modelling the behaviour of human muscles that are responsible for producing movements can be seen in the literature [15-17]. Although physiologically based muscle models, as in [16], are more accurate than others, several parameters need to be chosen and optimized to obtain acceptable muscle response, and this increases the complexity of implementing such models. For this reason, in this study, it is preferred to use the model featured in [17], which is simple to implement and is accurate enough as it has been derived from data obtained experimentally from paraplegics and healthy subjects. The model describes the relationship between electrical stimulus and knee joint torque produced by the quadriceps muscle group. To obtain the muscle model, a mathematical function describing the dynamic equilibrium of the moments acting on the knee joint was taken into consideration. In addition to the gravitational and inertial characteristics of the anatomical segments, the damping and stiffness properties of the knee were calculated. A passive pendulum test of the lower limb, to estimate the unknown parameters of viscous-elastic parameters, was performed. The knee movement after stimulating the quadriceps muscle and the produced muscle torque was recorded. Using autoregressive with exogenous inputs (ARX) model, with a parametric identification approach, to estimate a single pole transfer function, that describes the relationship between the electrical stimulus and the generated active muscle torque, is thus obtained. This is given as in (1):

$$H(s)=\frac{G}{1+s\tau} \qquad (1)$$

where: τ is the time constant of the pole and G represents the static gain

In this work, an average value of knee joint`s viscous coefficient for paraplegic subjects (0.287 N.m.s./rad) is added to the knee joint of the humanoid model. Also the static gain (0.04 Nm/μs) and the time constant (0.45 sec) values are used as provided in [17]. The stimulation frequency is fixed while the pulse width of the stimulus is kept variable to control the muscle torque.

C. Force Drop (Fatigue) Monitor

A monitor to measure the drop in force during the cycling training period is derived from experimentally-obtained data. The experiment, an isometric test, was achieved with the aid of a paraplegic subject of incomplete spinal cord lesion T2-T3. The test was performed by applying one stimulus per ten seconds (3 sec on and 7 sec off), with different pulse width (200μs to 400μs) while keeping other parameters fixed (current 40mA, frequency 30Hz) [19]. Making use of this data, a relationship that combines the pulse width and the number of stimulus with the resulted muscle force was derived and used in this work to monitor the force drop in leg muscles during FES cycling.

III. CONTROL STRATEGIES

To perform smooth FES driven cycling, i.e. coordinated leg cycling movement with no jerking, for a desired cycling speed, a closed-loop control mechanism is required. The torque generated by a muscle, as a response to FES signal, can be controlled by varying the pulse width of the stimulus. Knee angle reference for 35 rpm speed is used in this work. The actual knee trajectory, using a position sensor in the humanoid-bicycle model, is measured. The knee trajectory is used as a feedback and compared with the knee angle reference signal. According to the error signal, the controllers adjust the pulse width of the stimulus to control muscle torque and consequently control leg pedalling movement to follow the reference. It is worth mentioning that continuous and successive stimulation of a muscle can cause rapid muscle fatigue; therefore, for prolonged FES cycling, the muscle should be stimulated for short periods of time and give enough time to the muscle to rest before the next stimulation is due.

A. Scenario I

In this part, cycling performance is tested without the use of the flywheel or any other assisting device and by stimulating single muscle group, the quadriceps. By stimulating the quadriceps, only extension, i.e. pushing, torque can be generated. To control the cycling cadence, i.e. pedalling speed, it is required to provide an opposite torque to resist the movement in case the leg speed exceeds the required cadence. This torque can only be provided by stimulating the quadriceps of the opposite leg [12]. The push and resist duration was specified according to the crank angle, as shown in Fig. 2.

The control block diagram is shown in Fig. 3. The difference between the knee reference and the real knee trajectory is used as the error signal for the controllers to adjust the pulse width of the stimulus. The PD controllers values (PD1 and PD4: Kp=3.6, Kd=2.1, PD2 and PD3: Kp=4, Kd=1.8) were obtained heuristically. The push and resist periods in relation to crank angle is shown in Fig. 4. The right leg tracking the reference, the generated right and left muscle torque, and smoothness of the cycling are shown in Fig. 5, 6, 7 and 8 respectively.

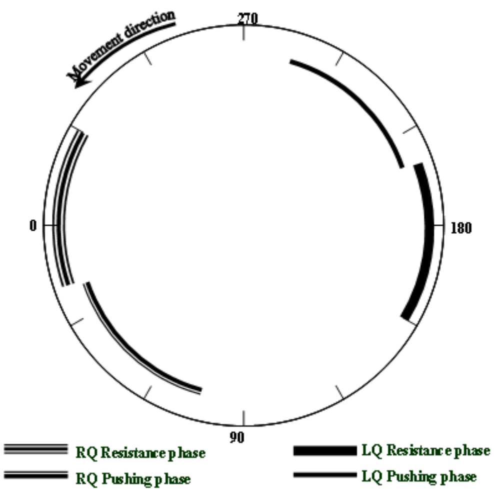

Figure 2. *Push and resist phases according to crank angle [12].*

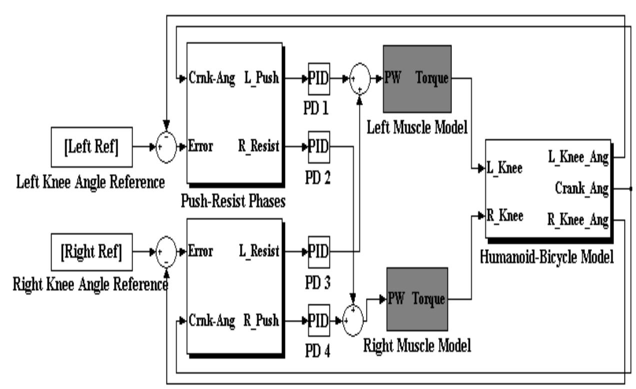

Figure 3. *The control block diagram*

Although the control strategy produced acceptable tracking performance (Fig. 5) and cycling with no jerking (Fig. 8), it can be noticed that each muscle is stimulated twice (Fig. 6 and Fig. 7), one for push and one for resist phase, per cycle. Practically, two successive muscle stimulations lead to rapid muscle fatigue as the muscle re-contracts before getting rest from first contraction. As a result, this scenario leads to rapid muscle fatigue and consequently termination of the training session after few cycles.

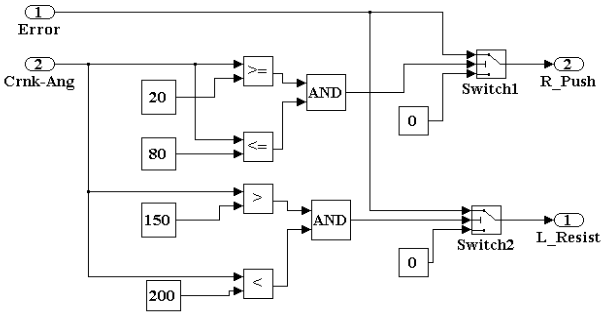

Figure 4. *Push and resist phase of right leg*

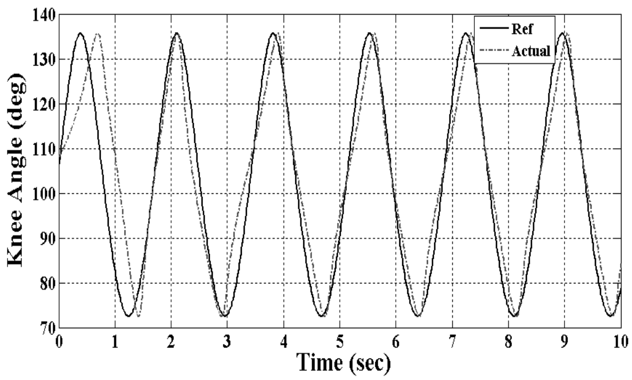

Figure 5. *Right leg tracking the reference*

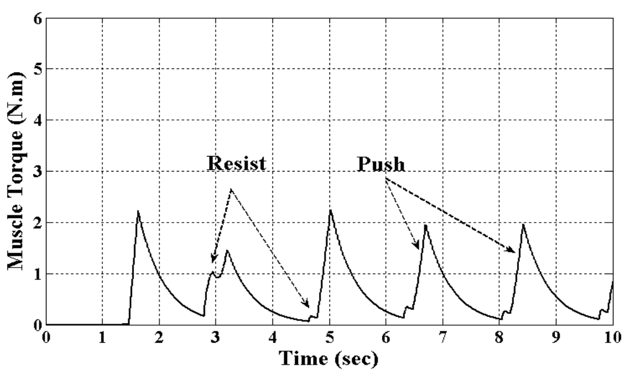

Figure 6. *Right muscle torque*

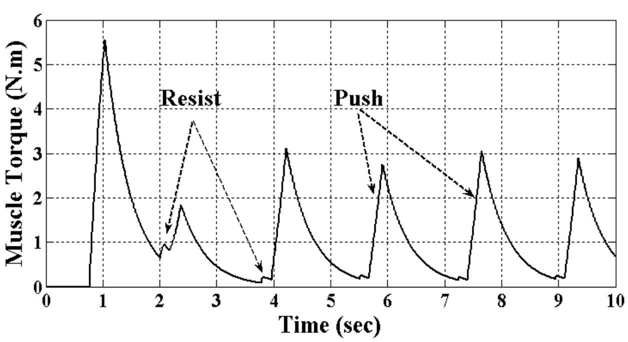

Figure 7. *Left muscle torque*

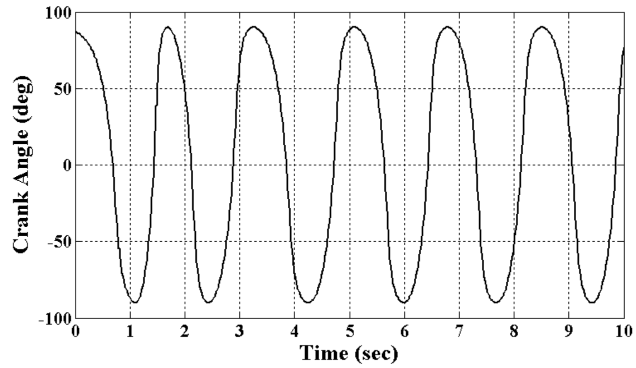

Figure 8. *Cycling smoothness*

B. Scenario II

In this part, to reduce the probability of muscle fatigue that might result from successive stimulation to the muscle (as in Scenario I), an energy storage device, i.e. a flywheel, with an electrically activated clutch mechanism are used to assist the cycling. The mechanism is used to assist and retard the cycling when necessary and to replace muscle stimulation in resist phase that used in *Scenario I*. The electrical clutch engages the flywheel with the shaft to absorb the excess in the energy produced by the leg and to store it as kinetic energy in the flywheel. Also the flywheel, loaded with energy, is engaged by the clutch to support the cycling in case the energy of the leg was not enough to pedal. The control block diagram is shown in Fig. 9.

As in Scenario I, the difference between the knee reference and the actual knee trajectory is used as input to the controllers to adjust the pulse width of the stimulus. The two PD controllers parameter values used in this scenario (PD1 and PD2: Kp=3.6, Kd=2.1) were the same as PD1 and PD4 in Scenario I. The flywheel velocity was compared with the derivative of the actual knee angle, i.e. angular velocity of the leg, to decide whether to engage the flywheel or not. For example, if the leg is faster than the required speed and the flywheel`s speed is less than the speed of the leg, the clutch will engage the flywheel to absorb the energy, store it as kinetic energy, and produce damping effect on the movement. While if the leg speed is less than the required speed and the flywheel`s speed is higher than that of the leg, the flywheel will be engaged to assist and speed-up the cycling by discharging its kinetic energy into the system. The engagement and disengagement of the flywheel by the clutch is shown in Fig. 13.

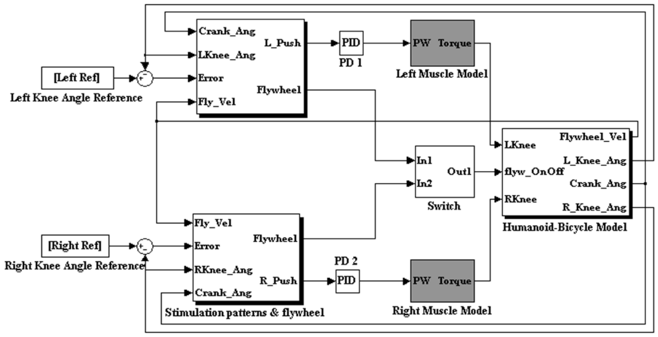

Figure 9. *Control block diagram using flywheel and electrical clutch assisting mechanism*

Although the control strategy produced smooth cycling (Fig. 14) it can be seen from Fig. 10 that there was a slight delay in the tracking at the first cycle. This is due to the fact that the flywheel assisting mechanism was activated after the first cycle to make use of the rotational momentum caused by the gravitational force on the leg. Even though, the mechanism with the controller was successful in following the reference in subsequent cycles.

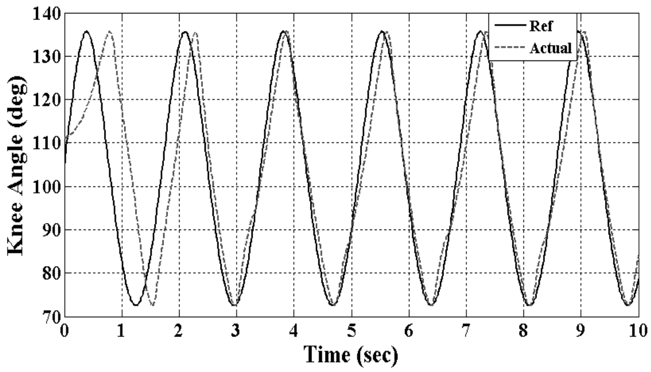

Figure 10. *Right leg tracking the reference*

It can be noticed from Fig. 11 and Fig. 12 that the muscles were stimulated only once per cycle during pushing phase and there were no successive stimulations as in Scenario I. From Fig. 15, it is clear that the fatigue in leg muscles appears faster in Scenario I, as the force drops faster than that in Scenario II. The new mechanism has decreased the fatigue rate by 15-18% (Fig. 16) in comparison with that of Scenario I, which means that the new mechanism has delayed the occurrence of muscle fatigue and enhanced the legs during cycling.

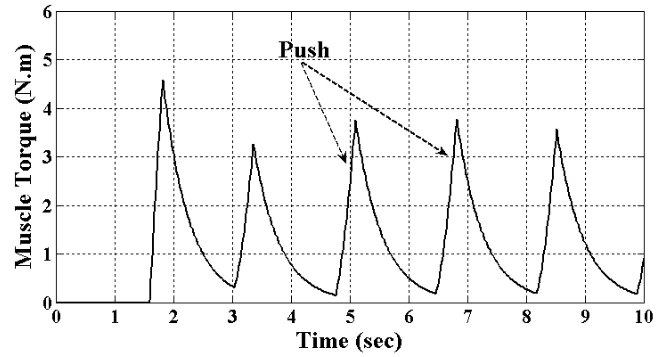

Figure 11. *Right muscle torque*

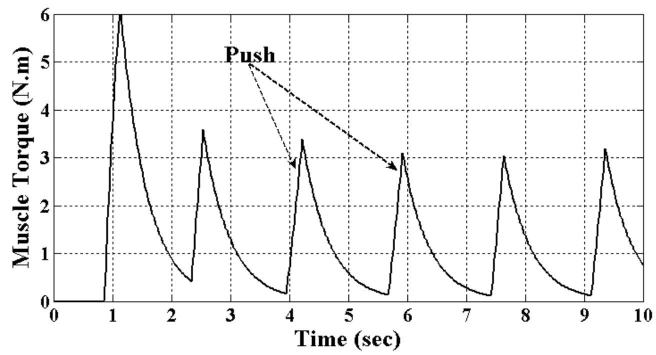

Figure 12. *Left muscle torque*

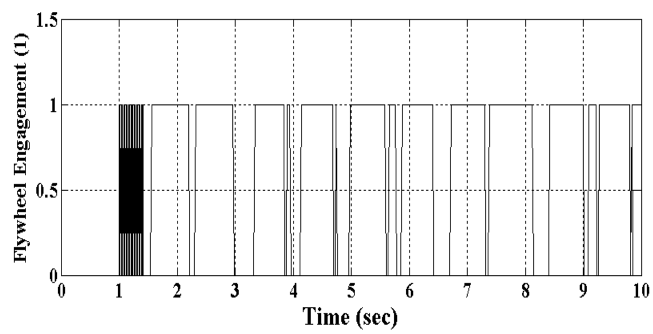

Figure 13. *Flywheel engagement and disengagement periods*

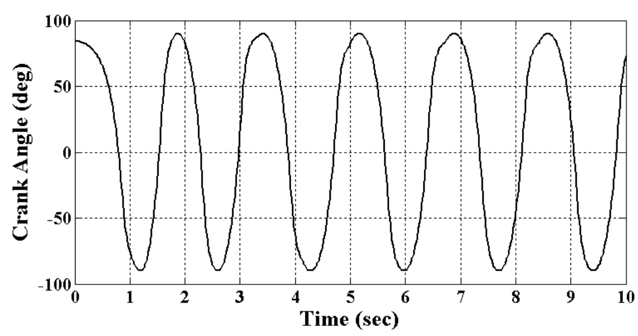

Figure 14. *Cycling smoothness*

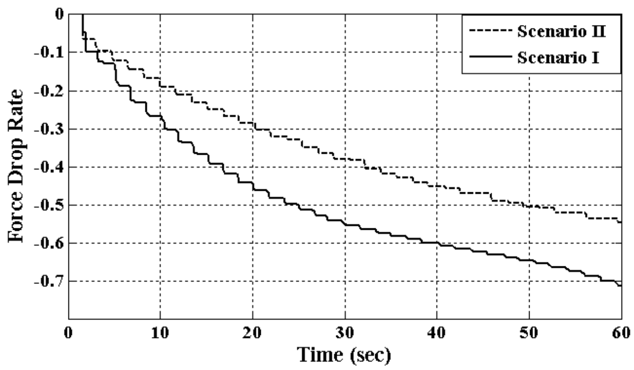

Figure 15. *Force drop rate in right muscle*

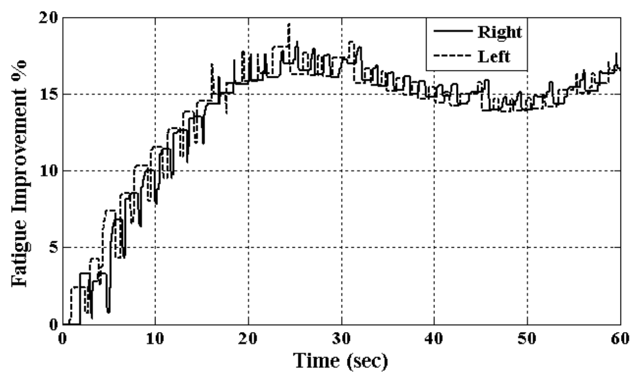

Figure 16. *Fatigue improvement percentage for Scenario II in comparison with Scenario I*

IV. CONCLUSION

Coordinated FES driven cycling can be performed by stimulating the quadriceps muscle group with the assistance of a flywheel and electrical clutch mechanism. With the new mechanism, the quadriceps muscle group will be stimulated once per cycle and will have enough time to rest before the next stimulus is due, and hence muscle fatigue will be delayed. As a result, this mechanism is promoting prolonged FES cycling by stimulating single muscle group.

REFERENCES

[1] J. C. Baldi, R. Jackson, R. Moraille, and W. J. Mysiw, "Muscle atrophy is prevented in patients with acute spinal cord injury using functional electrical stimulation," Spinal Cord, vol. 36, p. 463, 1998.

[2] G. M. Davis, N. A. Hamzaid, and C. Fornusek, "Cardiorespiratory, metabolic, and biomechanical responses during functional electrical stimulation leg exercise: health and fitness benefits," Artificial Organs, vol. 32, pp. 625-629, 2008.

[3] T. W. J. Janssen, R. M. Glaser, and D. B. Shuster, “Clinical efficacy of electrical stimulation exercise training: Effects on health, fitness, and function,” Top Spinal Cord Inj. Rehab., vol. 3, pp. 33–49, 1998.

[4] R. Kobetic, R. J. Triolo, J. P. Uhlir, C. Bieri, M. Wibowo, G. Polando, E. B. Marsolais, J. A. Davis Jr, K. A. Ferguson, and M. Sharma, "Implanted functional electrical stimulation system for mobility in paraplegia: a follow-up case report," Rehabilitation Engineering, IEEE Transactions on, vol. 7, pp. 390-398, 1999.

[5] J. Rasmussen, S. T. Christensen, M. Gföhler, M. Damsgaard, and T. Angeli, "Design optimization of a pedaling mechanism for paraplegics," Structural and Multidisciplinary Optimization, vol. 26, pp. 132-138, 2004.

[6] D. Pons, C. Vaughan, and G. Jaros, "Cycling device powered by the electrically stimulated muscles of paraplegics," Medical and Biological Engineering and Computing, vol. 27, pp. 1-7, 1989.

[7] J. Petrofsky and J. Smith, "Three-wheel cycle ergometer for use by men and women with paralysis," Medical and Biological Engineering and Computing, vol. 30, pp. 364-369, 1992.

[8] M. Gföhler, M. Loicht, and P. Lugner, "Exercise tricycle for paraplegics," Medical and Biological Engineering and Computing, vol. 36, pp. 118-121, 1998.

[9] T. Angeli, M. Gfohler, T. Eberharter, L. Rinder, "Tricycle for paraplegics using functional electrostimulation," Medical and Biological Engineering and Computing, vol 37, pp. 326-327, 1999.

[10] K. J. Hunt, B. Stone, N. O. Negard, T. Schauer, M. H. Fraser, A. J. Cathcart, C. Ferrario, S. A. Ward, and S. Grant, "Control strategies for integration of electric motor assist and functional electrical stimulation in paraplegic cycling: utility for exercise testing and mobile cycling," Neural Systems and Rehabilitation Engineering, IEEE Transactions on, vol. 12, pp. 89-101, 2004.

[11] C. Fornusek, G. M. Davis, P. J. Sinclair, and B. Milthorpe, "Development of an isokinetic functional electrical stimulation cycle ergometer," Neuromodulation: Technology at the Neural Interface, vol. 7, pp. 56-64, 2004.

[12] R. Massoud, “Intelligent control techniques for spring-assisted FES-cycling” PhD Thesis, Department of Automatic Control and System Engineering, University of Sheffield, United Kingdom, 2007.

[13] J. Szecsi, P. Krause, S. Krafczyk, T. Brandt, and A. Straube, "Functional output improvement in FES cycling by means of forced smooth pedalling," Medicine & Science in Sports & Exercise, vol. 39, p. 764, 2007.

[14] D. A. Winter, Biomechanics and motor control of human movement: John Wiley & Sons Inc, 2009.

[15] A. Hill, "The heat of shortening and the dynamic constants of muscle," Proceedings of the Royal Society of London. Series B, Biological Sciences, vol. 126, pp. 136-195, 1938.

[16] R. Riener and J. Quintern, "A physiologically based model of muscle activation verified by electrical stimulation," Bioelectrochemistry and bioenergetics, vol. 43, pp. 257-264, 1997.

[17] M. Ferrarin and A. Pedotti, "The relationship between electrical stimulus and joint torque: a dynamic model," Rehabilitation Engineering, IEEE Transactions on, vol. 8, pp. 342-352, 2000.

[18] R. Tong, Biomechatronics in medicine and health care, Singapore: Pan Stanford, 2011, pp.100.

[19] R. Jailani, “Analysis and control of FES-Assisted paraplegic walking with wheel walker” PhD Thesis, Department of Automatic Control and System Engineering, University of Sheffield, United Kingdom, 2011.