A forum for emerging systems and control technologies.

DYNAMIC SYSTEMS & CONTROL

SEPTEMBER 2016 VOL. 4 NO. 3

HEALTH CARE: SYSTEMS FOR REHABILITATION AND THE ELDERLY

ASME

CYCLING INDUCED BY FUNCTIONAL ELECTRICAL STIMULATION: A CONTROL SYSTEMS PERSPECTIVE

BY WARREN DIXON
PROFESSOR
DEPARTMENT OF MECHANICAL AND AEROSPACE ENGINEERING
UNIVERSITY OF FLORIDA

MATTHEW BELLMAN
FOUNDER & CHIEF TECHNOLOGY OFFICER
MYOLYN, LLC

An electric field can be applied to muscle to yield a contraction, generally termed neuromuscular electrical stimulation (NMES). When applied to yield functional tasks, it is more specifically termed functional electrical stimulation (FES). FES is commonly prescribed as a treatment for various neurological disorders (e.g., stroke, spinal cord injury, traumatic brain injury, Parkinson's disease, etc.). Such disorders affect millions of Americans, resulting in costs that exceed $100 billion per year. Common impairments caused by neurological disorders lead to limited physical activity due to muscle weakness, paralysis, and/or loss of limb coordination. In turn, limited physical activity increases the risk of negative secondary health effects such as obesity, poor self-image, diabetes, cardiovascular disease, and other chronic conditions. While current clinical practice typically uses open-loop FES, many researchers are currently focused on closed-loop, computer-controlled FES. In the development of such closed-loop FES algorithms, the musculoskeletal system is modeled as a mechanism comprising links and actuators that are activated by controls and robotics methods executed by a computer interface to the person (i.e., a cybernetic system).

Automated FES methods hold the potential to maximize therapeutic outcomes by self-adjusting to the particular individual, facilitating at-home therapy and enabling effective therapy from less experienced clinicians. Yet, the development of automated FES devices is complicated by the uncertain nonlinear musculoskeletal response to stimulation, including disturbances such as fatigue that are difficult to measure or model. Unfortunately, therapeutic dosage (i.e., intensity and number of contractions) is limited by the onset of fatigue and poor muscle response during fatigue. Specifically, fatigue results in diminished force production and increased delay in the muscle response, leading to inaccurate limb positioning and low musculoskeletal loading. As a result, methods to yield efficient muscle force/torque production (including shared force production by motorized systems such as robotic exoskeletons) and stimulation methods that mitigate fatigue (e.g., asynchronous and other multi-channel stimula-

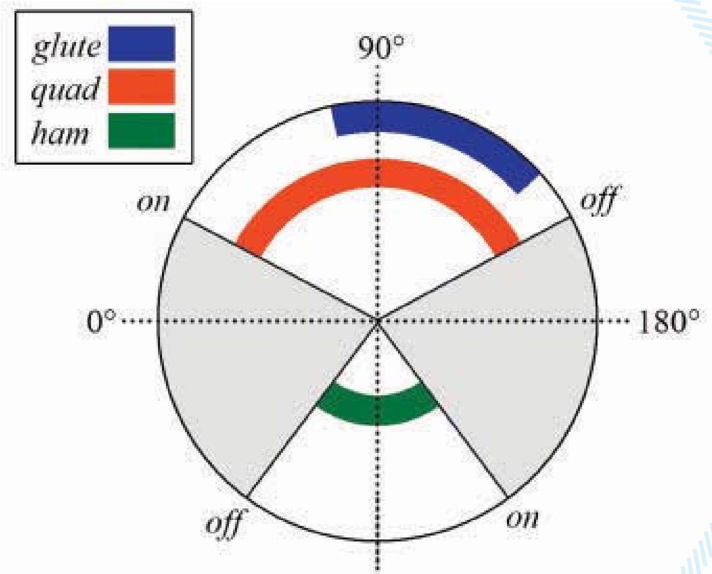

FIGURE 1 Example stimulation pattern as a function of the crank position for the gluteal, quadriceps, and hamstring muscle groups on a single leg. Grey shaded regions indicate sections of the crank cycle where it is inefficient for a selected muscle group to produce torque. Electrical stimulation is turned on and delivered to respective muscles in specified regions and turned off in regions where it is inefficient to produce torque.

tion methods) are the mainstream focus in general FES literature.

A common mode of activity-based therapy utilized clinically is stationary FES-cycling, since it is a low-impact, stable, repetitive exercise that involves coordinated limb motion. FES (often coupled with an electric motor to help turn the cycle crank) is a key enabler for rehabilitative cycling therapies because individuals with neurological disorders often lack sufficient strength and coordination for symmetric, volitional pedaling, making it difficult to achieve and maintain a sufficient workload for target heart rate thresholds and other desired training effects.

FES-induced cycling methods began to emerge in the 1980s. Early FES-cycling studies used open-loop or proportional-derivative feedback control of the stimulation intensity to achieve a desired cycling cadence. Over the past several decades, researchers began using tools from the control systems community to improve FES-cycling performance, including: system identification and pole placement methods for linearized models, robust control methods such as sliding mode control, and intelligent control methods such as neural networks and fuzzy logic.

All of these previous FES-cycling control studies alternated stimulation across different muscle groups according to a predefined stimulation pattern. The stimulation pattern defines the segments of the crank cycle over which each muscle group is stimulated to achieve the desired cycling motion. **Figure 1** depicts an example stimulation pattern from [1] wherein the quadriceps, hamstrings, and gluteal muscle groups are stimulated. The stimulation pattern plays an essential role in segregating the control input across different muscle groups and can vary according to the muscle groups involved and whether a motor is included. Various strategies have been developed to determine the stimulation pattern including: manual determination based on observation, offline numerical optimization, analysis of the person's kinematic relationships, or electromyography (EMG) of able-bodied cyclists.

FES-CYCLING AS A SWITCHED SYSTEM

It is well known that switching between different closed-loop subsystems can destabilize the system, even when those subsystems are exponentially stable (cf., [2],[3]). To induce cycling, FES is applied to different sets of muscles of the left and right legs according to a switching signal dictated by the stimulation pattern, which depends on the crank angular position. Hence, FES-induced cycling is a switched control system with autonomous, state-dependent switching. As illustrated in **Figure 1** there are regions of the crank cycle where it is kinematically inefficient to produce torque (voluntarily or by FES). In these regions, stimulation is not applied, because this can lead to increased muscle fatigue. When muscle groups are activated by a computer-controlled method in the unshaded regions of **Figure 1**, exponential convergence of the cadence tracking error can be achieved [1]; however, when the crank enters the shaded region and no stimulation is applied, the cadence error system can become unstable. Specifically, if the limbs do not have sufficient momentum, when stimulation is removed (and voluntary efforts by individuals are diminished and operating in an inefficient region) the cycle may come to rest in a region where it cannot escape, leading to unbounded cadence tracking error. This problem is exacerbated by the fact that the dynamics of the FES-cycling system are nonlinear, time-varying, and uncertain, so that the system's state trajectories are unknown *a priori* and difficult to predict. With the exception of recent results by the authors (cf. [1], [4, 5, 6]), FES-cycling results have not considered these practical stability issues. Investigating FES-cycling in the light of switched systems theory may yield control strategies that improve FES-cycling performance, thereby increasing the safety and effectiveness of FES-cycling.

In [1], switched systems methods are used to switch between different muscle groups for the right and left legs. Specifically, a stimulation pattern for the gluteal, quadriceps femoris, and hamstrings muscle groups was analytically derived from the kinematics of the cycle-rider system based on the ability of the rider's hip and knee joints to produce a forward torque about the cycle crank. A robust sliding mode control input was then developed based on the derived stimulation pattern with the goal for the rider to pedal (induced by FES) at a desired cadence. A Lyapunov-based analysis was then used to show that the cadence tracking error is exponentially stable in the controlled (unshaded) regions of the crank cycle and could be upper bounded by exponential growth in the uncontrolled (shaded) regions. By developing a ratio between the rate of decrease and rate of increase of the cadence errors in the different regions, sufficient conditions were constructed that indicate how long the crank needs to spend in each region (i.e., dwell-time conditions). The developed conditions are a function of bounds on the uncertain dynamics and the desired cadence. Essentially, from the bounds on the dynamics, the cadence tracking error, and the desired cadence, sufficient conditions can be developed to ensure that there is enough momentum to carry the limbs into the next region of controlled pedaling. The overall error system decays to a residual steady-state error (ultimately bounded stability result).

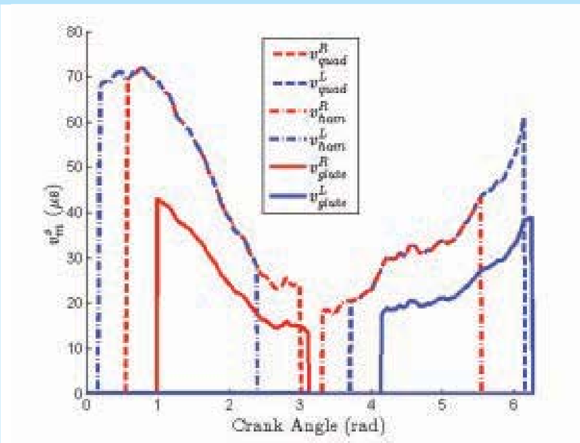

FIGURE 2 Stimulation pulse width for each muscle group for a healthy normal person as a function of the crank cycle.

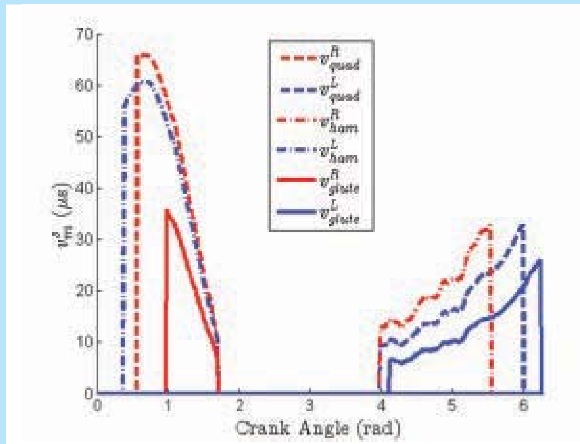

FIGURE 3 Stimulation pulse width for each muscle group as a function of the crank cycle for an individual with decreased strength in the right side due to Parkinson's disease.

Since different individuals have a different response to stimulation, experiments were performed in [1] on four able-bodied individuals and one individual with Parkinson's disease. The cycling protocol was reviewed and approved by the University of Florida Institutional Review Board (IRB), and before participating each person was required to meet the inclusion criteria and sign an informed consent. The electrical stimulation control input was delivered to the muscle groups through biphasic, symmetric, rectangular pulses applied to self-adhesive, cutaneous electrodes. The stimulation frequency was fixed at 60 Hz (higher frequencies yield more intense and continuous contractions, but also lead to increased fatigue) and the stimulation intensity was controlled by fixing the pulse amplitude for each muscle group (i.e., between 50-110 mA) and controlling the pulse width according to a sliding mode controller. As indicated in [1], when the healthy normal volunteers were able to look at the desired trajectory error to guide their cycling efforts, they were able to achieve a desired cadence of 50 revolutions per minute (RPM) with an average of -0.14 ± 1.40 RPMs of steady-state error using only volitional effort and no FES contribution. When they were not able to see the desired trajectory and were asked to sit passively while the sliding mode controller coordinated their limbs to produce cycling, the steady-state error was 5.94 ± 1.76 RPMs on average. An example control input across each muscle group of the healthy normal participants is depicted in **Figure 2**.

The subject with Parkinson's disease exhibited mild bilateral motor impairment with evident tremor, where his right side exhibited greater impairment. The impact of his impairment on his cycling performance was that, when his right leg was supposed to pedal, his cadence decreased significantly. This individual was not able to maintain cadence tracking using FES alone (the stimulation intensities to achieve cycling exceeded his tolerance), so comparisons were made with his completely voluntary ability versus his voluntary ability augmented with FES (i.e., FES was used in conjunction with his voluntary effort to compensate for deficiencies in his right side). His average tracking error was 0.43±4.06 RPMs using only voluntary inputs and 0.17±3.11 RPMs when his voluntary inputs were augmented with FES control. **Figure 3** illustrates the differences in stimulation when his right leg was activated versus his left leg (compare to the more symmetric control input for the healthy normal subject).

MOTORIZED FES-CYCLING AS A SWITCHED SYSTEM

Given the existence of regions in the crank cycle where it is inefficient to produce torque, a motor can be included as another torque source. See the Sidebar "Motorized Cycling Testbed Construction" for further details on an example cycle testbed. From a review of research results and commercial products, the use of a motor to augment FES cycling is common, but when the motor is activated and in what capacity varies significantly. In [4], the motor input was only used in the regions where the torque production was inefficient and FES was turned off (i.e., the shaded regions in **Figure 1**). From a rehabilitation perspective, only using the motor in the inefficient regions maximizes the effort required by the person (in contrast to other designs where the motor continuously provides assistance) and helps to reduce fatigue since a person's muscles are only activated when it is kinematically efficient to do so, which can vary depending on a person's physiology and ability to produce torque. From a control systems perspective, the advantage of using a motor is that it eliminates the uncontrolled regions, simplifying control design and analysis strategies. That is, compared to results such as [1], switching with a motor in the loop only involves switching between stable subsystems. FES control of the muscles yields cadence tracking in torque efficient regions while the motor yields cadence tracking when it is inefficient for the limbs to produce torque. A sliding mode controller is developed in [4] for the muscle control inputs and the motor. Since each closed-loop error system is stable, a single Lyapunov function can be established to show global exponential tracking for the overall switched system, without the need to develop sufficient dwell-time conditions. Experiments were performed on five healthy normal individuals in [4], where they were not informed of the desired trajectory and were asked to relax and let the FES and motor controller manage the cycling cadence. In these experiments, only the hamstrings and quadriceps were stimulated. The average error across all five subjects for

ABOUT THE AUTHORS

Matthew Bellman has been a member of the Nonlinear Controls and Robotics Group since May 2009, and has since received both his bachelor's and master's degrees, magna cum laude, in mechanical engineering with a minor in biomechanics from the University of Florida. Matthew recently completed his doctoral degree in mechanical engineering as a National Defense Science and Engineering (NDSEG) Fellow, with a focus on the theoretical development of robust, adaptive control systems for applications involving functional electrical stimulation, specifically those involving rehabilitation and mobility of the lower extremities. His work has been recognized by the 2015 American Automatic Control Council (AACC) O. Hugo Schuck (Best Paper) Award

Prof. Warren Dixon received his Ph.D. in 2000 in the Department of Electrical and Computer Engineering from Clemson University. He was selected as an Eugene P. Wigner Fellow at Oak Ridge National Laboratory (ORNL). In 2004, he joined the University of Florida in the Mechanical and Aerospace Engineering Department. His main research interest has been the development and application of Lyapunov-based control techniques for uncertain nonlinear systems. He has published 3 books, an edited collection, 12 chapters, and over 120 journal and 220 conference papers. His work has been recognized by the 2015 & 2009 American Automatic Control Council (AACC) O. Hugo Schuck (Best Paper) Award, the 2013 Fred Ellersick Award for Best Overall MILCOM Paper, a 2012-2013 University of Florida College of Engineering Doctoral Dissertation Mentoring Award, the 2011 American Society of Mechanical Engineers (ASME) Dynamics Systems and Control Division Outstanding Young Investigator Award, the 2006 IEEE Robotics and Automation Society (RAS) Early Academic Career Award, an NSF CAREER Award (2006-2011), the 2004 Department of Energy Outstanding Mentor Award, and the 2001 ORNL Early Career Award for Engineering Achievement. He is an ASME and IEEE Fellow, an IEEE Control Systems Society (CSS) Distinguished Lecturer, and served as the Director of Operations for the Executive Committee of the IEEE CSS Board of Governors (2012-2015). He currently serves as a member of the U.S. Air Force Science Advisory Board. He is currently or was formerly an associate editor for *ASME Journal of Journal of Dynamic Systems, Measurement and Control, Automatica, IEEE Control Systems Magazine, IEEE Transactions on Systems Man and Cybernetics: Part B Cybernetics, and the International Journal of Robust and Nonlinear Control.*

REFERENCES

1 M. J. Bellman, T. -H. Cheng, R. J. Downey, C. J. Hass, W. E. Dixon, "Switched Control of Cadence During Stationary Cycling Induced by Functional Electrical Stimulation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, to appear.

2 D. Liberzon, *Switching in Systems and Control*, Birkhauser, 2003.

3 R. Goebel, R. G. Sanfelice, A. R. Teel, "Hybrid Dynamical Systems," *IEEE Control Systems Magazine*, Vol. 29, No. 2, pp. 28-93, April 2009.

4 M. J. Bellman, R. J. Downey, A. Parikh, W. E. Dixon, "Automatic Control of Cycling Induced by Functional Electrical Stimulation with Electric Motor Assistance," *IEEE Trans. Autom. Science Eng.*, to appear.

5 M. J. Bellman, T.-H. Cheng, R. J. Downey, W. E. Dixon, "Stationary Cycling Induced by Switched Functional Electrical Stimulation Control" *Proc. Am. Control Conf.*, 2014, pp. 4802-4809.

6 H. Kawai, M. J. Bellman, R. J. Downey, W. E. Dixon, "Tracking Control for FES-Cycling based on Force Direction Efficiency with Antagonistic Bi-Articular Muscles," *Proc. Am. Control Conf.*, 2014, pp. 5484-5489.

7 K. J. Hunt, B. Stone, N.-O. Negård, 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," *IEEE Trans. Neural Syst. Rehabil. Eng.*, Vol. 12, No. 1, pp. 89–101, March 2004.

8 A. Farhoud and A. Erfanian, "Fully automatic control of paraplegic FES pedaling using higher-order sliding mode and fuzzy logic control," *IEEE Trans. Neural Syst. Rehabil. Eng.*, Vol. 22, No. 3, pp. 533–542, 2014.

a desired cadence of 50 RPM was 0.00±2.91 RPM. The stimulation/motor control input for one cycle by an example individual is depicted in **Figure 4**.

CONCLUSIONS AND ONGOING WORK

FES plays an important role in the rehabilitation of individuals with neurological disorders that exhibit muscle dysfunction. The potential impact to society and the daily lives of individuals with certain neurological disorders provides significant motivation to examine the challenges associated with FES-induced activities. In particular, FES-induced cycling is a common activity-based rehabilitation therapy because it is a safe, repetitive, and low impact exercise. However, as illustrated by the aforementioned studies, there exist significant challenges to the development of FES controllers for the uncertain, switched, nonlinear dynamic system.

While some success has been achieved by using various off-the-shelf control solutions, significant promise exists for new developments to emerge from the control and robotics communities, where constructive

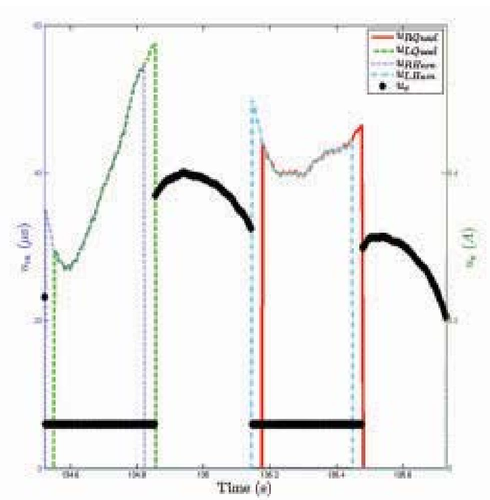

FIGURE 4 Stimulation pulse width (denoted by u_m in microseconds) and motor control (denoted by u_e in amps) inputs for the quadriceps and hamstring muscle groups and an electric motor for an example healthy normal individual.

methods could be developed and analyzed for the problems with FES-induced activities. In particular, the control systems and robotics communities have the potential to make significant inroads in FES-cycling, where relatively few constructive closed-loop controllers have been published. To date, only robust control tools have been used to develop a stability analysis for the switched system. However, motivation exists to develop and analyze adaptive and learning controllers which may exhibit lower frequency content and/or lower magnitude control intensities. Such developments may lead to reduced muscle fatigue, thereby extending the rehabilitative treatment, and to better measures of therapeutic outcomes by means of system identification strategies. When switching between stable and unstable regions (i.e., without a motor in the loop), significant challenges remain in developing sufficient dwell-time conditions to ensure stability for adaptive switched systems. Specifically, since adaptive systems typically yield asymptotic convergence, the development of dwell-time conditions is an open challenge. The inclusion of a motor enables switching between stable systems and eliminates the need for the development of sufficient dwell-time conditions. Hence, the development of adaptive switched controllers for motorized FES-cycling systems may have a closer horizon. The inclusion of a motor also expands the possible control objectives that can be pursued. For example, physical therapists would like to prescribe both a desired power output and cadence for individuals participating in cycling therapies. Based on results such as [7,8], the motor could be tasked with maintaining cadence control, allowing the FES inputs to yield desired torques. Such development is still in the early stages, and various adaptive and learning tools can potentially be developed to advance such goals. ■

FIGURE 5 Stationary motorized FES cycle testbed.

MOTORIZED CYCLING TESTBED CONSTRUCTION

The latest motorized FES cycle used at the University of Florida Nonlinear Control and Robotics Laboratory is a modified, commercially available recumbent tricycle (TerraTrike Rover) depicted in Figure 5. A 250 Watt, brushed, 24 VDC electric motor was mounted to the frame and coupled to the drive chain. Orthotic boots incorporated with custom pedals are used to fix the rider's feet to the pedals, prevent dorsiflexion and plantarflexion of the ankles and maintain sagittal alignment of the lower legs. An optical encoder is coupled to the cycle crank via spur gears to measure the crank position. Current control of the cycle's motor was enabled by a linear amplifier interfacing with data acquisition hardware (Quanser Q8-USB), which also measured the encoder signal. To ensure safe operation, an emergency stop is included and, when people are seated on the tricycle, the tricycle's seat position is adjusted for each subject's comfort while ensuring that full extension of the knees is not possible (to prevent hyperextension). Electric stimulation is provided by a commercial, current-controlled, eight-channel stimulator (RehaStim, Hasomed) interfaced with a personal computer and cutaneous electrodes (Axelgaard Manufacturing Co. Ltd.). One pedal is attached to a cycling power meter (SRM) that wirelessly transmits the torque produced at the crank for experiments that involve power tracking control objectives.