

# Muscle Stimulation Waveform Timing Patterns for Upper and Lower Leg Muscle Groups to Increase Muscular Endurance in Functional Electrical Stimulation Pedaling Using a Forward Dynamic Model

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**Abstract**—Functional electrical stimulation (FES) of pedaling provides a means by which individuals with spinal cord injury can obtain cardiorespiratory exercise. However, the early onset of muscle fatigue is a limiting factor in the cardiorespiratory exercise obtained while pedaling an FES ergometer. One objective of this study was to determine muscle excitation timing patterns to increase muscle endurance in FES pedaling for three upper leg muscle groups and to compare these timing patterns to those used in a commercially available FES ergometer. The second objective was to determine excitation timing patterns for a lower leg muscle group in conjunction with the three upper leg muscle groups. The final objective was to determine the mechanical energy contributions of each of the muscle groups to drive the crank. To fulfill these objectives, we developed a forward dynamic simulation of FES pedaling to determine electrical stimulation ON and OFF times that minimize the muscle stress–time integral of the stimulated muscles. The computed electrical stimulation ON and OFF times differed from those utilized by a commercially available FES ergometer and resulted in 17% and 11% decrease in the muscle stress–time integral for the three upper leg muscle groups and four upper and lower leg muscle groups, respectively. Also, the duration of muscle activation by the hamstrings increased by 5% over a crank cycle for the computed stimulation ON and OFF times, and the mechanical energy generated by the hamstrings increased by 20%. The lower leg muscle group did not generate sufficient mechanical energy to reduce the energy contributions of the upper leg muscle groups. The computed stimulation ON and OFF times could prolong FES pedaling, and thereby provide improved cardiorespiratory and muscle training outcomes for individuals with spinal cord injury. Including the lower leg muscle group in FES pedaling could increase cardiorespiratory demand while not affecting the endurance of the muscles involved in the pedaling task.

**Index Terms**—Electrical stimulation, energy, functional electrical stimulation (FES), muscle, pedaling, recumbent, rehabilitation, simulation.

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## I. INTRODUCTION

FUNCTIONAL electrical stimulation (FES) provides a means to animate the leg muscles of individuals with spinal cord injury (SCI) for exercise activities such as pedaling. Previous research has indicated that FES pedaling by activating the quadriceps, hamstrings, and gluteal muscle groups is beneficial in that it leads to elevated cardiorespiratory activity [1]–[3], improved circulation [2], [4]–[8], reduced muscle atrophy [2], [9]–[11], and an improved sense of well-being [12]. However, a limiting factor in the benefit from FES pedaling is the short duration and low work rate, and thereby low levels of work achieved due to the onset of muscle fatigue [2], [13], [14].

The endurance of muscles in FES applications is affected by several factors, but the condition of the muscles (e.g., degree of atrophy, fiber-type composition) and the stimulation waveforms used to activate the muscles are of primary importance. Because the muscles themselves are not immediately alterable, efforts have been directed toward manipulating the stimulation waveforms to increase muscle endurance in electrical stimulation applications. Studies directed at identifying waveform characteristics (e.g., pulsewidth, number of pulses in a pulse train, frequency, and duty cycle) of individual muscle groups [2], [15]–[19] to optimize muscle recruitment through electrical stimulation have applicability to FES pedaling. However, because pedaling requires coordination of multiple stimulated muscle groups, muscle stimulation ON and OFF timing is also relevant.

Previous research directed at improving the efficacy of FES pedaling by altering electrical stimulation ON and OFF timing has taken two approaches: empirical and computer-model-based studies. Empirical studies have tested the effects that alterations to a set waveform (e.g., maximum intensity [20], [21], ON and OFF timing [20], [22], ramped versus no ramped onset and offset [23]) have on FES pedaling. While these studies have led to an increased understanding of how different stimulation parameters affect FES pedaling, the rationale for the parameter changes (e.g., the extent to which ON and OFF timing was changed) was not always apparent. In contrast, computer-model-based studies of FES pedaling have been used to determine electrical stimulation ON and OFF times that are designed to achieve a specific goal, for example, to maximize mechanical power output [24], [25].

However, a systematic and predictive analysis to determine electrical stimulation ON and OFF timing that is directed at increasing the duration and mechanical work associated with FES pedaling has not been performed.

A relationship exists between the endurance of a muscle and the muscle force–time integral, which reflects the interaction between force amplitude, duration of contraction, and rest interval between contractions [26], [27]. Additionally, it has been demonstrated that a reduction of the force–time integral for a single muscle group leads to an increase in the duration of the force generating capacity of the muscle group [26], [27]. There has also been a long held association between the reduction of the muscle stress–time integral and the increased endurance of multiple muscles working together to perform a gross motor task such as walking or pedaling [28]–[31]. However, there is no known study that has confirmed the relationship between the reduction of the muscle stress–time integral and increased endurance of several muscles working together to perform a gross motor task. To this end, the first objective of this study was to use a forward dynamic model to compute the electrical stimulation timing patterns that minimize the stress–time integral of the muscles used to pedal an FES ergometer. A secondary objective was to compare the muscle mechanical energies resulting from the computed electrical stimulation ON and OFF times with those associated with a commercially available FES ergometer to identify differences in the distribution of the total muscle mechanical energy required to pedal the ergometer.

Prior research using empirically derived stimulation timing patterns for the lower leg muscles has demonstrated that incorporating more muscle mass in the exercise activity provides metabolic and cardiopulmonary benefit [20], [21], yet none of the commercially available FES ergometers provides stimulation to the lower leg muscles. A theoretical analysis using a forward dynamic model determined that no mechanical energy gains could be obtained by incorporating the lower leg muscles in the exercise and releasing the ankle joint (existing FES ergometers fix the ankle joint in the neutral position) [25]. However, pedaling with a fixed ankle would reduce the number of DOFs and would permit the gastrocnemius and soleus to contract, thereby increasing the cardiorespiratory demand of the exercise and the training of these muscles. Further, the gastrocnemius could generate mechanical energy at the knee that contributes to the pedaling motion and reduces the energy demand on the upper leg muscles. Therefore, the second objective of this study was to determine the muscle ON and OFF timing for the upper and lower leg muscle groups that minimize the stress–time integral. A related secondary objective was to quantify the extent by which the addition of the lower leg muscles alters the mechanical energy demands of the upper leg muscles to pedal at a set work rate.

## II. METHODS

A modified version of the forward dynamic model of planar recumbent pedaling developed previously [32] formed the basis of the model used in our simulations. The modifications were made to account for the different rider–ergometer interface and

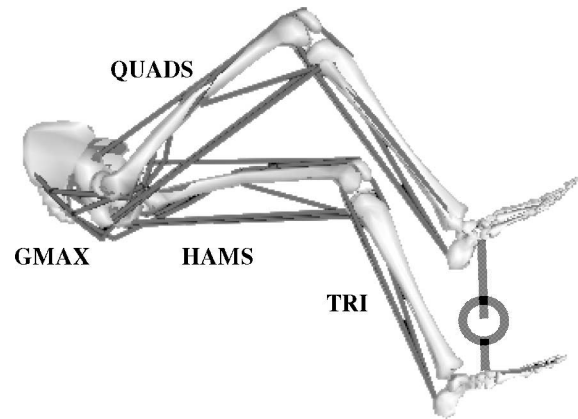


Fig. 1. Position of the seat in the recumbent position relative to the crank spindle. The beginning of the crank cycle ( $0^\circ$ ) was defined when the right crank arm was directed vertically upward (top dead center). The stimulated muscles of the upper and lower legs were combined into four muscle groups based on their function and cocontraction during FES pedaling include HAMS (semimembranosus, semitendinosus, and the long and short head of the biceps femoris), QUADS (the vastii and rectus femoris), GMAX (gluteus maximus), and TRI (soleus and medial and lateral gastrocnemius).

musculoskeletal activation and contraction dynamics associated with FES pedaling.

The forward dynamic model was comprised of two legs, two pedals, and a crank that was representative of the ergometer dynamics. Each leg was modeled as three segments (thigh, shank, and foot). The foot was rigidly attached to the pedal, and the two pedals were connected to a crank that rotated about an axis fixed in the inertial frame. The hip joint and pedal and crank spindles were modeled as revolute. The knee was modeled as a single-DOF planar joint with two translational motions that were constrained by the knee flexion angle [33]. The patella was constrained to follow a set path relative to the femur and defined as a function of knee flexion angle [34]. The ankle joint was fixed by means of a rigid boot as required by the FES ergometer. Each leg was treated as symmetric but  $180^\circ$  out of phase with the contralateral limb. The resulting kinematics of the rider–ergometer model had 1 DOF and could be expressed as a function of the crank angle [35]. A torque equivalent to the resistive and inertial torques produced by the ergometer for a 25-W work rate and 50 rev/min pedaling rate was applied about the rotational axis of the crank arm [36]. The effective inertia of the ergometer crank (e.g., the pedals, crank arms, chain rings, and flywheel) was computed using the manufacturer specifications of the flywheel and components [36]. The fixed axis of the crank was positioned relative to the pelvis to account for the recumbent pedaling position and to limit the knee angle to  $45^\circ$  of flexion (full extension equals  $0^\circ$ ) with the ankle in the neutral position (i.e., the foot and tibia form a  $90^\circ$  angle) [37] (ERGYS 2, Therapeutic Alliances, Inc., Dayton, OH, manufacturer recommendations; see Fig. 1).

The muscle origin and insertion points and the muscle moment arm lengths were based on the work of Delp *et al.* [34]. Via points along the musculotendon path for cases in which the muscle either wrapped around bone or was constrained by retinacula were also used [34]. A generic Hill-type

lumped-parameter model [38] formed the basis of the muscle contraction dynamics.

A forward dynamic model was generated using Dynamics Pipeline (MusculoGraphics, Inc., Santa Rosa, CA) from the equations of motion for the rider–ergometer system derived using SD/FAST (Parametric Technology Corporation, Needham, MA). Based on their function and cocontraction with surface electrical stimulation, ten muscles per leg were included into four muscle groups and were used to drive the model. The muscle groups and associated muscles were HAMS (semimembranosus, semitendinosus, biceps femoris long head and biceps femoris short head), QUADS (three-component vastus and rectus femoris), GMAX (gluteus maximus), and TRI (soleus and medial and lateral gastrocnemius) (see Fig. 1). Additional muscles not activated by electrical stimulation, namely adductor magnus, iliacus, and psoas, were included in the model to account for their passive properties in pedaling. The maximum isometric strength of each of the muscles was reduced to 25% of the corresponding value used previously for able-bodied pedaling [32], [39], [40] to account for muscle weakening associated with SCI and the magnitude of muscle force generated using surface electrical stimulation [16], [25], [41]–[44].

All of the muscles in a group received the same excitation signal. The excitation signal to each of the muscles was represented by an isosceles trapezoidal sloped pattern that defined the ON and OFF timing and amplitude of the muscle excitation [23]. The sloped regions each occurred over  $21^\circ$  of the crank cycle. The activation dynamics were represented by a first-order differential equation model [38] with a time constant of 0.07 s [35]. A 50-ms delay was incorporated into the model to account for the time delay between the electrical stimulation onset (and offset) and force initiation (and cessation) [17], [45]. The Hill muscle model constants  $a$  and  $b$  were set to  $0.35 \times$  the maximum isometric force and  $2.25 \text{ s}^{-1} \times$  the optimal muscle fiber length, respectively, to account for the force–velocity relationship of a muscle with larger percentage of type-II fast fibers [25], [46].

The forward dynamic model was used to compute the electrical stimulation ON and OFF times and amplitudes for the stimulated muscle groups that minimized the sum of the areas under the active component (i.e., no passive forces) of the muscle stress–time curve for each muscle (except soleus) and the difference between the areas under the active component of the muscle stress–time curves of all muscle pairs (except those pairs with soleus) over a crank cycle. The soleus muscle stress was not included in the cost calculation because the soleus muscle could not contribute to the pedaling motion due to the fixed ankle joint. The performance criterion  $J$  was as follows:

$$J = \sum_{i=1}^p \int_{t_{0i}}^{t_{fi}} \left( \frac{F_i}{A_i} \right) dt + \sum_{i=1}^p \sum_{j=1}^p \left| \int_{t_{0i}}^{t_{fi}} \left( \frac{F_i}{A_i} \right) dt - \int_{t_{0j}}^{t_{fj}} \left( \frac{F_j}{A_j} \right) dt \right|, \quad (1)$$

for  $i \neq j$  and only different combinations of  $i, j$

where  $F_i$  is the force of the  $i$ th muscle,  $A_i$  is the physiologic cross-sectional area of the  $i$ th muscle,  $t$  is the time,  $t_{0i}$  and  $t_{fi}$  are the ON and OFF times, respectively, of the  $i$ th muscle, and  $p$  is the number of activated muscles. The physiologic cross-sectional area was determined by normalizing the maximum isometric

strength of the muscle by the maximum active stress (i.e., specific tension/cross-sectional area). A maximum active stress of 250 kPa was used [9]. The optimal electrical stimulation ON and OFF times and amplitudes were obtained by converting the optimal control problem into a parameter optimization problem [47] and using a simulated annealing optimization algorithm [48] to compute the excitation parameters that both minimized the cost  $J$  and satisfied a time constraint requiring an average pedaling rate within 1 rev/min of the target 50 rev/min pedaling rate. The performance criterion (1) and time constraint were evaluated on the fourth cycle of the simulation to enable the simulation to reach steady state and become independent of the initial conditions [39].

To satisfy the first objective, the electrical stimulation ON and OFF timing and amplitudes for three of the four muscle groups (HAMS, QUADS, and GMAX) were computed (Stim3). Additionally, to determine the minimum muscle stress–time integral and associated muscle mechanical energies for the commercially available computer-controlled electrical stimulation ergometer (ERGYS 2, Therapeutic Alliances, Inc., Dayton, OH), the electrical stimulation ON and OFF angles utilized by the ERGYS 2 ergometer were programmed in the forward dynamic model to compute the stimulation amplitudes that yielded the minimum cost (i.e., minimum stress–time integral) for the predetermined ON and OFF angles (StimErg). For both simulations, the resulting net mechanical energy of each of the activated muscles was calculated by integrating the instantaneous muscle mechanical power over the crank cycle. To satisfy the second objective, electrical stimulation ON and OFF times and amplitudes for all four muscle groups were computed (Stim4). The net mechanical energy of each of the activated muscles that could contribute to the pedaling motion (i.e., all the muscles except soleus) was calculated to quantify the energy of the lower leg muscles in relation to the mechanical energy of the upper leg muscles.

### III. RESULTS

The electrical stimulation ON and OFF timing that minimized the stress–time integral for the HAMS, QUADS, and GMAX muscle groups (Stim3) differed from those utilized by the commercially available FES ergometer (StimErg). Stim3 ON and OFF timing shifted earlier in the crank cycle for the HAMS and GMAX and later in the crank cycle for the QUADS compared to StimErg (see Fig. 2). The difference between the Stim3 and StimErg ON timing was largest for the HAMS ( $17^\circ$ ) and smallest for the QUADS ( $6^\circ$ ) (see Table I). The difference between the Stim3 and StimErg OFF timing was largest for the GMAX ( $9^\circ$ ) and smallest for the QUADS ( $2^\circ$ ) (see Table I).

Stim3 resulted in a different distribution of net mechanical energy generated across the three muscle groups of the right leg than StimErg. There was a 20% increase (from 3.13 to 3.74 J) in the net mechanical energy generated by the HAMS with Stim3 as compared to StimErg (see Table II). Also, there was a 4% decrease (from 10.91 to 10.44 J) and a 2% decrease (from 2.73 to 2.66 J) in the net mechanical energy generated by the QUADS and GMAX, respectively, with Stim3 (see Table II).

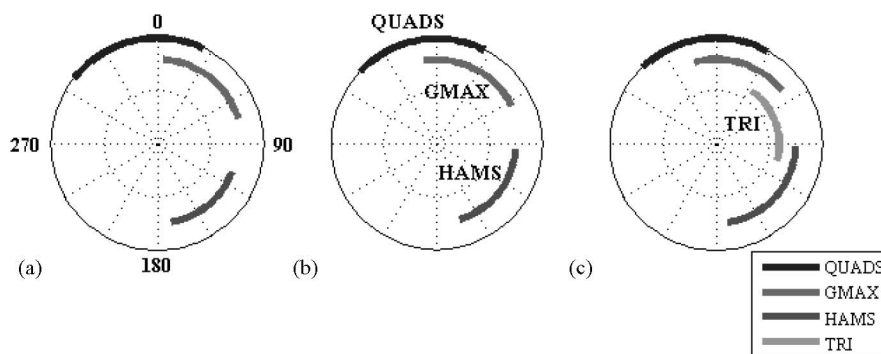


Fig. 2. Plot of muscle electrical stimulation ON and OFF timing as a function of crank angle for (a) the commercially available electrical stimulation ergometer (StimErg), and the minimized stress–time integral for (b) the three upper leg muscle groups stimulated (Stim3) and (c) both the three upper leg muscle groups and the one lower leg muscle group stimulated (Stim4). Top dead center indicates  $0^\circ$  and the beginning of the crank cycle. The ON and OFF timing angles are listed in Table I.

TABLE I  
ELECTRICAL STIMULATION MUSCLE EXCITATION ON AND OFF TIMES  
(BY CRANK ANGLE)

		StimErg (deg)	Stim3 (deg)	Stim4 (deg)
TRI	on			33
	off			106
HAMS	on	111	94	92
	off	170	163	171
QUADS	on	308	314	316
	off	25	27	28
GMAX	on	4	351	345
	off	72	63	52

TABLE II  
NET MECHANICAL ENERGY GENERATED BY THE MUSCLE GROUPS FROM  
SIMULATIONS DRIVEN BY ELECTRICAL STIMULATION MUSCLE EXCITATION  
ON AND OFF TIMES (BY CRANK ANGLE)

Muscle Group	StimErg (J)	Stim3 (J)	Stim4 (J)
TRI*			-0.27
HAMS	3.13	3.74	3.74
QUADS	10.91	10.44	10.73
GMAX	2.73	2.66	2.73

\*Note that the mechanical energy for the TRI muscle group corresponds to the mechanical energy of the gastrocnemius muscle because the soleus contracts isometrically due to the fixed ankle joint.

The performance criterion cost  $J$  associated with Stim3 (0.29 Pa·s) was 17% lower than that of StimErg (0.35 Pa·s).

The electrical stimulation ON and OFF timing for two (HAMS and GMAX) of the three muscle groups common to both Stim3 and Stim4 was different. HAMS OFF timing occurred  $8^\circ$  later in the crank cycle with Stim4 than Stim3 (see Table I and Fig. 2). The ON and OFF timing of the GMAX with Stim4 preceded those of Stim3 by  $6^\circ$  and  $11^\circ$ , respectively (see Table I and Fig. 2).

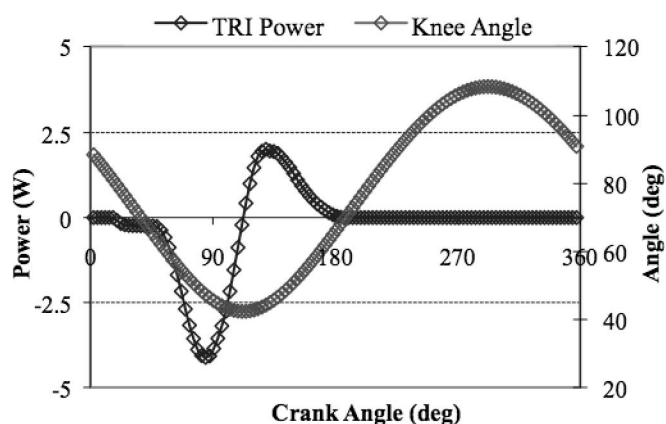


Fig. 3. Muscle power generated by the TRI muscle group using the Stim4 electrical stimulation ON and OFF timing and amplitudes and the corresponding knee angle as a function of crank angle. A knee joint angle equal to  $0^\circ$  is equivalent to the corresponding joint angle when in the anatomical position.

The ON and OFF timing of the TRI muscle group bridged the ON and OFF timing of the GMAX and HAMS muscle groups.

The addition of the lower leg muscle group resulted in small increases in the net mechanical energy generated by the QUADS and GMAX of the right leg, 3% and 2%, respectively, compared to Stim3. For both Stim3 and Stim4, the QUADS generated about four times more work than GMAX.

The TRI muscle group performed negative work, and then positive work as the knee transitioned from extension to flexion (see Fig. 3). As a result, the net mechanical energy from TRI was nearly zero (see Table II). The performance criterion cost for Stim4 (0.35 Pa·s) was 2.6% lower than that of Stim3, when adjusted for the number of muscles included in the cost function (i.e., the cost associated with the first term of (1) was normalized to the number of muscles activated and the cost associated with the second term of (1) was normalized to the number of two-muscle combinations).

#### IV. DISCUSSION

Because FES pedaling provides benefit to individuals with SCI and because a limiting factor in the benefit of FES pedaling is the low work performed, the objectives of this study were:

1) to determine the electrical stimulation ON and OFF timing for the upper leg muscles that minimize the stress–time integral; 2) to quantify the muscle mechanical energies associated with the ON and OFF timing that minimize the stress–time integral and compare them to those associated with an existing FES ergometer; 3) to determine the electrical stimulation ON and OFF timing for the upper and lower leg muscles that minimize the stress–time integral; and 4) to quantify the changes in muscle mechanical energies of the upper leg muscles when the lower leg muscles are also activated. One key finding from these objectives was that for the upper leg muscle groups, the cost function was lower for the electrical stimulation ON and OFF timing that minimized the stress–time integral than for the existing FES ergometer. Also, for the upper leg muscle groups, another key finding was that the electrical stimulation ON and OFF timing that minimized our performance criterion resulted in a 20% increase in HAMS mechanical energy compared to the ON and OFF timing used by the existing FES ergometer. A final key finding was that the electrical stimulation timing for the upper and lower leg muscle groups resulted in virtually no net mechanical energy by the TRI muscle group.

Before addressing the importance of these findings, a discussion of the underlying assumptions and limitations of the model is warranted. The tendon slack length parameter values used in the model were the same as those used previously for able-bodied cyclists [32], [39], [40]. While the potential for tendon lengths to shorten due to muscle disuse exists, tendon shortening does not always occur. The maximum isometric force of the muscles was set at 25% of the values used previously for able-bodied cyclists [32], [39], [40] to account for muscle atrophy and spatial limitations of stimulating muscle with surface electrodes. The choice of percentage in force reduction should not have influenced the results because the value is within the range of values reported in the literature for electrical stimulation studies of paralyzed muscle (e.g., [32], [39], and [40]). Furthermore, none of the computed electrical stimulation amplitudes associated with the ON and OFF times were saturated (i.e., reached the maximum value of 1.0).

An assumption in the model was that the activation dynamics time constant and the delay between electrical stimulation onset (and offset) and force initiation (and cessation) do not change. However, it is known that there are differences in the time constant and delay for isometric contractions at different joint angles [45]. Because the values of these time constants are dynamic and vary with muscle–tendon length to an extent that is currently not known, we used values of the time constant and delay that reflected the average values for the isometric case over the range of joint angles observed in our simulation. Additionally, it was assumed that there would be no unanticipated muscle activity (i.e., spasms) associated with the electrical stimulation.

Another assumption was related to the results generated using the StimErg electrical stimulation patterns. StimErg ON and OFF timing was based on EMG studies of the corresponding muscles of nondisabled individuals as they pedaled the ergometer. Therefore, the ON and OFF timing utilized by the commercially available ergometer may not minimize the muscle stress–time integral. However, by setting the ON and OFF timing equal to that

of the ERGYS 2 and using the forward dynamic model and optimization routine to determine the stimulation amplitudes that minimized the cost function, the model-generated results represent a “best-case” scenario for minimal stress–time integral for the StimErg ON and OFF timing patterns and were, therefore, worthy of examination.

The minimization of the stress–time integral of activated muscle as a means to increase muscle mechanical work in FES pedaling warrants discussion. Because a relationship exists between muscle endurance and the muscle force–time integral [27], minimization of the force–time integral (for a single muscle) or the stress–time integral (for a group of muscles working together) is a logical approach to increase the endurance and the resulting work generated during FES pedaling. Additionally, the muscle stress–time integral was utilized in this study instead of the more common muscle stress-squared or stress-cubed time integral [28]–[31]. Whereas raising stress to the second or third power would have yielded a more equal distribution of stress across muscles [31], it also would have had a negative impact on the optimization results. Raising stress to the second or third power would have weighted muscle stress more than time in the objective function, and thereby resulted in electrical stimulation patterns that activated the muscles over a greater portion of the crank cycle (i.e., larger duty cycle). Longer duty cycles decrease muscle endurance in surface electrical stimulation activities [16], [19], [49]. Instead, the second term in our cost function served to distribute the stress–time integral across the muscles more equally.

The muscle stress–time integral used in conjunction with the forward dynamic simulation resulted in duty cycles that are advantageous for electrical stimulation applications. The Stim3 ON and OFF times result in duty cycles for each of the three muscle groups of either 19% (HAMS) or 20% (QUADS and GMAX). These duty cycles are meaningful for two reasons. First, the duty cycles are similar for each of the muscle groups. The similarity in duty cycles indicates that each of the muscle groups was equally taxed with respect to excitation time. Second, previous studies directed at the relationship between duty cycle and endurance in muscle activated by electrical stimulation in a time period similar to that observed in our pedaling study indicate that a duty cycle of 20% resulted in greater endurance than longer duty cycles [16], [19], [50]. These observations provide added confidence that the computed Stim3 electrical stimulation ON and OFF times would lead to increased endurance in FES pedaling.

The electrical stimulation ON and OFF times comprising Stim3 minimize the muscle stress–time integral of the individual muscles and the differences in the muscle stress–time integrals of all activated muscle pairs. The differences between Stim3 and StimErg are large enough to indicate that the ON and OFF timing for StimErg does not minimize our cost function as defined by (1). However, as indicated in a pilot study that we have performed, the differences in ON and OFF times between Stim3 and StimErg are not large enough to prevent an individual with SCI to pedal an FES ergometer using the Stim3 ON and OFF times.

The differences in ON and OFF times between Stim3 and StimErg electrical stimulation patterns gave rise to different

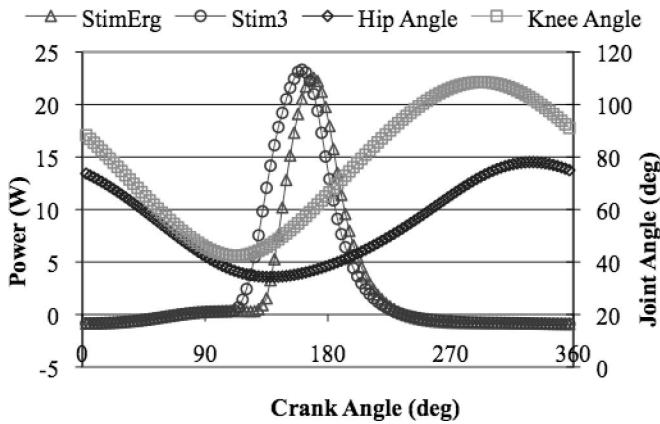


Fig. 4. Muscle power generated by the HAMS using the StimErg and Stim3 electrical stimulation ON and OFF timing and amplitudes and the corresponding hip and knee angles as a function of crank angle. A hip joint angle and knee joint angle equal to  $0^\circ$  is equivalent to the corresponding joint angles when in the anatomical position.

mechanical energies generated by the muscle groups. Based on the relationship between HAMS origins and insertions and the hip and knee trajectories in fixed-ankle pedaling, HAMS was capable of generating mechanical energy as soon as the knee transitioned from extension to flexion (see Fig. 4). The earlier onset of the HAMS excitation associated with Stim3 corresponded with the knee transition from extension to flexion and resulted in the HAMS capacity to generate more mechanical energy to contribute to pedaling compared to StimErg (see Fig. 4). The magnitude of variation between the mechanical energies generated by the QUADS and GMAX using Stim3 and StimErg, respectively, was not as meaningful as that for HAMS. The increase in mechanical energy generated by the HAMS compensated for the decreased mechanical energy generated by the QUADS and GMAX (see Table II).

The addition of the lower leg muscle group (TRI) resulted in minor shifts of the ON and OFF times of the upper leg muscle groups. Whereas the QUADS ON and OFF times were similar to Stim3, the GMAX ON and OFF times shifted earlier in the crank cycle and the HAMS ON time shifted earlier and the OFF time shifted later in the crank cycle. The TRI muscle group served to bridge the cessation of the GMAX excitation and the beginning of the HAMS excitation (see Table I and Fig. 2). The result that the sequential excitation of the TRI before the HAMS differed from the coactivation of the TRI and HAMS observed in our simulations of fixed-ankle FES pedaling is fundamentally different than the EMG timing of these muscles when able-bodied subjects pedal with a released ankle [32], [51]. In normal recumbent pedaling with a released ankle, the muscles in the TRI group are coactivated with the HAMS, generate little mechanical energy, serve to stabilize the ankle joint, and transfer mechanical energy between the limbs and the crank [32]. However, in FES pedaling with a fixed ankle, the TRI muscles do not need to stabilize the ankle and only the gastrocnemius muscle of the TRI muscle group can generate force to flex the knee over a range of the crank cycle.

Stimulating the lower leg muscles in conjunction with the upper leg muscles would be expected to increase the cardiorespiratory demand of FES pedaling. Based on measures of leg muscle volumes and their corresponding masses [52], [53], reductions to stimulated muscle mass due to muscle hypertrophy and surface electrical stimulation [16], [25], [41]–[44], estimates of the rate of oxygen consumption per kilogram of skeletal muscle [54], and measures of the rate of oxygen consumption during FES pedaling [2], [22], including the lower leg muscles in FES pedaling with Stim4, should increase the rate of oxygen consumption by 3%–10%.

The results of this study provide new information on the use of forward dynamic modeling combined with nonlinear optimization as a means to determine muscle excitation timing to fulfill a complicated task in a coordinated manner. Previous studies utilizing forward dynamic simulations of FES pedaling have sought to determine the seating parameters that would enable a larger population to benefit from the exercise [35] and determine electrical stimulation timing patterns that maximize mechanical power output [24], [25]. While the goal of achieving maximum power output addresses the low work rate of current FES pedaling, it could lead to high muscle forces and low endurance. Instead, minimization of the factors that contribute to a decrease in the endurance of muscles performing a gross motor task could prolong the duration of the task, and thereby provide increased muscle metabolic responses and muscle training. The reduction in the performance criterion cost associated with Stim3 and Stim4 compared to StimErg provides support for experimental studies to determine whether the computed muscle excitation timing patterns increase the duration of FES pedaling and lead to improved FES pedaling outcomes.

In summary, this study used forward dynamic simulations to determine the electrical stimulation ON and OFF timing for three upper leg muscle groups and for three upper leg muscle groups and one lower leg muscle group that minimize the stress–time integral in FES pedaling. The electrical stimulation ON and OFF timing that minimized the stress–time integral for the three upper leg muscle groups differed from that utilized by the commercially available FES ergometer. Because the magnitude of the stress–time integral is an indicator of muscle endurance in gross motor tasks, spinal cord injured subjects might be expected to pedal with increased work production. The addition of the lower leg muscle group did not reduce the mechanical energy demands of the upper leg muscle groups during pedaling. Accordingly, including the lower leg muscles in FES pedaling would be expected to increase the muscle metabolic response to the exercise while not affecting the endurance of the muscles involved in the pedaling task. A useful next step would be to test the new ON and OFF timing patterns to validate these expectations experimentally.

## REFERENCES

- [1] T. J. Barstow, A. M. Scremin, D. L. Mutton, C. F. Kunkel, T. G. Cagle, and B. J. Whipp, "Changes in gas exchange kinetics with training in patients with spinal cord injury," *Med. Sci. Sports Exerc.*, vol. 28, no. 10, pp. 1221–1228, 1996.
- [2] S. P. Hooker, S. F. Ficoni, M. M. Rodgers, R. M. Glaser, T. Mathews, A. G. Suryaprasad, and S. C. Gupta, "Physiologic effects of electrical

- stimulation leg cycle exercise training in spinal cord injured persons," *Arch. Phys. Med. Rehabil.*, vol. 73, no. 5, pp. 470–476, 1992.
- [3] T. Mohr, J. L. Andersen, F. Biering-Srensen, H. Galbo, J. Bangsbo, A. Wagner, and M. Kjaer, "Long-term adaptation to electrically induced cycle training in severe spinal cord injured individuals," *Spinal Cord*, vol. 35, no. 1, pp. 1–16, 1997.
  - [4] H. L. Gerrits, A. de Haan, A. J. Sargeant, H. van Langen, and M. T. Hopman, "Peripheral vascular changes after electrically stimulated cycle training in people with spinal cord injury," *Arch. Phys. Med. Rehabil.*, vol. 82, no. 6, pp. 832–839, 2001.
  - [5] S. P. Hooker, S. F. Figoni, R. M. Glaser, M. M. Rodgers, B. N. Ezenwa, and P. D. Faghri, "Physiologic responses to prolonged electrically stimulated leg-cycle exercise in the spinal cord injured," *Arch. Phys. Med. Rehabil.*, vol. 71, no. 11, pp. 863–869, 1990.
  - [6] S. P. Hooker, S. F. Figoni, M. M. Rodgers, R. M. Glaser, T. Mathews, A. G. Suryaprasad, and S. C. Gupta, "Metabolic and hemodynamic responses to concurrent voluntary arm crank and electrical stimulation leg cycle exercise in quadriplegics," *J. Rehabil. Res. Dev.*, vol. 29, no. 3, pp. 1–11, 1992.
  - [7] M. S. Nash, S. Bilsker, A. E. Marcillo, S. M. Isaac, L. A. Botelho, K. J. Klose, B. A. Green, M. T. Rountree, and J. D. Shea, "Reversal of adaptive left ventricular atrophy following electrically-stimulated exercise training in human tetraplegics," *Paraplegia*, vol. 29, no. 9, pp. 590–599, 1991.
  - [8] S. F. Pollack, K. Axen, N. Spielholz, N. Levin, F. Haas, and K. T. Ragnarsson, "Aerobic training effects of electrically induced lower extremity exercises in spinal cord injured people," *Arch. Phys. Med. Rehabil.*, vol. 70, pp. 214–219, 1989.
  - [9] R. L. Lieber, *Skeletal Muscle Structure and Function: Implications for Rehabilitation and Sports Medicine*. Baltimore, MD: Williams & Wilkins, 1992.
  - [10] M. M. Rodgers, R. M. Glaser, S. F. Figoni, S. P. Hooker, B. N. Ezenwa, S. R. Collins, T. Mathews, A. G. Suryaprasad, and S. C. Gupta, "Musculoskeletal responses of spinal cord injured individuals to functional neuromuscular stimulation-induced knee extension exercise training," *J. Rehabil. Res. Dev.*, vol. 28, no. 4, pp. 19–26, 1991.
  - [11] L. A. Bremner, K. E. Sloan, R. E. Day, E. R. Scull, and T. Ackland, "A clinical exercise system for paraplegics using functional electrical stimulation," *Paraplegia*, vol. 30, no. 9, pp. 647–655, 1992.
  - [12] M. L. Sipski, J. A. Delisa, and S. Schweer, "Functional electrical stimulation bicycle ergometry: Patient perceptions," *Amer. J. Phys. Med. Rehabil.*, vol. 68, no. 3, pp. 147–149, 1989.
  - [13] P. C. Eser, N. de N. Donaldson, H. Knecht, and E. Stussi, "Influence of different stimulation frequencies on power output and fatigue during FES-cycling in recently injured SCI people," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 11, no. 3, pp. 236–240, Sep. 2003.
  - [14] D. L. Mutton, A. M. E. Scremin, T. J. Barstow, M. D. Scott, C. F. Kunkel, and T. G. Cagle, "Physiologic responses during functional electrical stimulation leg cycling and hybrid exercise in spinal cord injured subjects," *Arch. Phys. Med. Rehabil.*, vol. 78, no. 7, pp. 712–718, 1997.
  - [15] Z. Z. Karu, W. K. Durfee, and A. M. Barzilai, "Reducing muscle fatigue in FES applications by stimulating with N-let pulse trains," *IEEE Trans. Biomed. Eng.*, vol. 42, no. 8, pp. 809–817, 1995.
  - [16] H. B. Boom, A. J. Mulder, and P. H. Veltink, "Fatigue during functional neuromuscular stimulation," *Progr. Brain Res.*, vol. 97, pp. 409–418, 1993.
  - [17] S. G. Carroll, R. J. Triolo, H. J. Chizeck, R. Kobetic, and E. B. Marsolais, "Tetanic responses of electrically stimulated paralyzed muscle at varying interpulse intervals," *IEEE Trans. Biomed. Eng.*, vol. 36, no. 7, pp. 644–653, 1989.
  - [18] S. A. Binder-Macleod, E. E. Halden, and K. A. Jungles, "Effects of stimulation intensity on the physiological responses of human motor units," *Med. Sci. Sports Exerc.*, vol. 27, no. 4, pp. 556–565, 1995.
  - [19] J. Ding, A. S. Wexler, and S. A. Binder-Macleod, "A predictive fatigue model—II: Predicting the effect of resting times on fatigue," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 10, no. 1, pp. 59–67, 2002.
  - [20] T. W. J. Janssen, R. M. Glaser, J. W. Almeyda, D. D. Pringle, and T. Mathews, "Improving FES-leg cycle ergometer performance in individuals who have plateaued during long-term training," in *Proc. RESNA 1996—Exploring New Horizons: Pioneering the 21st Century*, Washington, DC: Resna Press, pp. 288–290.
  - [21] R. M. Glaser, W. P. Couch, T. W. J. Janssen, J. W. Almeyda, D. D. Pringle, S. R. Collins, and T. Mathews, "A development system to enhance FES leg cycle ergometer technology," in *Proc. RESNA 1996—Exploring New Horizons: Pioneering the 21st Century*, Washington, DC: Resna Press, pp. 279–281.
  - [22] K. J. Hunt, C. Ferrario, S. Grant, B. Stone, A. N. McLean, M. H. Fraser, and D. B. Allan, "Comparison of stimulation patterns for FES-cycling using measures of oxygen cost and stimulation cost," *Med. Eng. Phys.*, vol. 28, no. 7, pp. 710–718, 2006.
  - [23] T. W. Janssen, M. Bakker, A. Wyngaert, K. H. Gerrits, and A. de Haan, "Effects of stimulation pattern on electrical stimulation-induced leg cycling performance," *J. Rehabil. Res. Dev.*, vol. 41, no. 6A, pp. 787–796, 2004.
  - [24] M. Gfohler and P. Lugner, "Cycling by means of functional electrical stimulation," *IEEE Trans. Rehabil. Eng.*, vol. 8, no. 2, pp. 233–243, 2000.
  - [25] A. J. van Soest, M. Gfohler, and L. J. Casius, "Consequences of ankle joint fixation on FES cycling power output: A simulation study," *Med. Sci. Sports Exerc.*, vol. 37, no. 5, pp. 797–806, 2005.
  - [26] T. Dolmage and E. Cafarelli, "Rate of fatigue during repeated submaximal contractions of human quadriceps muscle," *Can. J. Physiol. Pharmacol.*, vol. 69, no. 10, pp. 1410–1415, 1991.
  - [27] Q. T. Tran, D. Docherty, and D. Behm, "The effects of varying time under tension and volume load on acute neuromuscular responses," *Eur. J. Appl. Physiol.*, vol. 98, no. 4, pp. 402–410, 2006.
  - [28] F. C. Anderson and M. G. Pandey, "Static and dynamic optimization solutions for gait are practically equivalent," *J. Biomech.*, vol. 34, no. 2, pp. 153–161, 2001.
  - [29] R. D. Crowninshield and R. A. Brand, "A physiologically based criterion of muscle force prediction in locomotion," *J. Biomech.*, vol. 14, no. 11, pp. 793–801, 1981.
  - [30] R. R. Neptune and M. L. Hull, "A theoretical analysis of preferred pedaling rate selection in endurance cycling," *J. Biomech.*, vol. 32, no. 4, pp. 409–415, 1999.
  - [31] B. I. Prilutsky and R. J. Gregory, "Analysis of muscle coordination strategies in cycling," *IEEE Trans. Rehabil. Eng.*, vol. 8, no. 3, pp. 362–370, 2000.
  - [32] N. A. Hakansson and M. L. Hull, "Influence of pedaling rate on muscle mechanical energy in low power recumbent pedaling using forward dynamic simulations," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 15, no. 4, pp. 509–516, 2007.
  - [33] G. T. Yamaguchi and F. E. Zajac, "A planar model of the knee joint to characterize the knee extensor mechanism," *J. Biomech.*, vol. 22, no. 1, pp. 1–10, 1989.
  - [34] S. L. Delp, J. P. Loan, M. G. Hoy, F. E. Zajac, E. L. Topp, and J. M. Rosen, "An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures," *IEEE Trans. Biomed. Eng.*, vol. 37, no. 8, pp. 757–767, 1990.
  - [35] L. M. Schutte, M. M. Rodgers, F. E. Zajac, and R. M. Glaser, "Improving the efficacy of electrical stimulation-induced leg cycle ergometry: An analysis based on a dynamic musculoskeletal model," *IEEE Trans. Rehabil. Eng.*, vol. 1, no. 2, pp. 109–125, 1993.
  - [36] B. J. Fregly, F. E. Zajac, and C. A. Dairaghi, "Bicycle drive system dynamics: Theory and experimental validation," *J. Biomech. Eng.*, vol. 122, no. 4, pp. 446–452, 2000.
  - [37] R. D. Trumbower and P. D. Faghri, "Kinematic analyses of semireclined leg cycling in able-bodied and spinal cord injured individuals," *Spinal Cord*, vol. 43, no. 9, pp. 543–549, 2005.
  - [38] F. E. Zajac, "Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control," *Crit. Rev. Biomed. Eng.*, vol. 17, no. 4, pp. 359–411, 1989.
  - [39] R. R. Neptune and M. L. Hull, "Evaluation of performance criteria for simulation of submaximal steady-state cycling using a forward dynamic model," *J. Biomech. Eng.*, vol. 120, no. 3, pp. 334–341, 1998.
  - [40] C. C. Raasch, F. E. Zajac, B. Ma, and W. S. Levine, "Muscle coordination of maximum-speed pedaling," *J. Biomech.*, vol. 30, no. 6, pp. 595–602, 1997.
  - [41] H. L. Gerrits, A. De Haan, M. T. E. Hopman, L. H. V. Van der Woude, D. A. Jones, and A. J. Sargeant, "Contractile properties of the quadriceps muscle in individuals with spinal cord injury," *Muscle Nerve*, vol. 22, no. 9, pp. 1249–1256, 1999.
  - [42] E. A. Hilleagass and G. A. Dudley, "Surface electrical stimulation of skeletal muscle after spinal cord injury," *Spinal Cord*, vol. 37, no. 4, pp. 251–257, 1999.
  - [43] E. Rabischong and F. Ohanna, "Effects of functional electrical-stimulation (FES) on evoked muscular output in paraplegic quadriceps muscle," *Paraplegia*, vol. 30, no. 7, pp. 467–473, 1992.
  - [44] M. Levy, J. Mizrahi, and Z. Susak, "Recruitment, force and fatigue characteristics of quadriceps muscles of paraplegics isometrically activated by

- surface functional electrical stimulation," *J. Biomed. Eng.*, vol. 12, no. 2, pp. 150–156, 1990.
- [45] P. J. Sinclair, R. M. Smith, and G. M. Davis, "The effect of joint angle on the timing of muscle contractions elicited by neuromuscular electrical stimulation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 12, no. 2, pp. 303–306, 2004.
- [46] P. J. Sinclair, G. M. Davis, and R. A. Smith, "Musculo-skeletal modelling of NMES-evoked knee extension in spinal cord injury," *J. Biomech.*, vol. 39, no. 3, pp. 483–492, 2006.
- [47] M. G. Pandy, F. C. Anderson, and D. G. Hull, "A parameter optimization approach for the optimal control of large-scale musculoskeletal systems," *J. Biomech. Eng.*, vol. 114, no. 4, pp. 450–460, 1992.
- [48] W. L. Goffe, G. D. Ferrier, and J. Rogers, "Global optimization of statistical functions with simulated annealing," *J. Econometrics*, vol. 60, pp. 65–99, 1994.
- [49] E. D. Zonnevjlle, N. N. Somia, R. W. Stremel, C. J. Maldonado, P. M. Werker, M. Kon, and J. H. Barker, "Sequential segmental neuromuscular stimulation: An effective approach to enhance fatigue resistance," *Plast. Reconstr. Surg.*, vol. 105, no. 2, pp. 667–673, 2000.
- [50] M. B. Reid, G. J. Grubwieser, D. S. Stokic, S. M. Koch, and A. A. Leis, "Development and reversal of fatigue in human tibialis anterior," *Muscle Nerve*, vol. 16, no. 11, pp. 1239–1245, 1993.
- [51] N. A. Hakansson and M. L. Hull, "Functional roles of the leg muscles when pedaling in the recumbent versus the upright position," *J. Biomech. Eng.*, vol. 127, no. 2, pp. 301–310, 2005.
- [52] J. A. Friederich and R. A. Brand, "Muscle-fiber architecture in the human lower-limb," *J. Biomech.*, vol. 23, no. 1, pp. 91–95, 1990.
- [53] T. L. Wickiewicz, R. R. Roy, P. L. Powell, and V. R. Edgerton, "Muscle architecture of the human lower-limb," *Clin. Orthop. Relat. Res.*, no. 179, pp. 275–283, 1983.
- [54] J. P. LeMaitre, S. Harris, J. Hannan, K. A. A. Fox, and M. A. Denvir, "Maximum oxygen uptake corrected for skeletal muscle mass accurately predicts functional improvements following exercise training in chronic heart failure," *Eur. J. Heart Failure*, vol. 8, no. 3, pp. 243–248, 2006.



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