Functional Electrical Stimulation

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CLOSED-LOOP CONTROL OF INDUCED MUSCLE CONTRACTIONS

I n North America, 400,000 people are living with spinal cord injury (SCI), and approximately 11,000 more such injuries occur each year [1]. SCI is caused by diseases that destroy the neurological tissue of the spinal cord or by trauma that compresses, stretches, or severs this tissue. SCI is often irreversible and can result in partial or total loss of sensory function, paralysis, or both to parts of the body below the level of the injury. For example, an injury to the spinal cord at the lower back usually affects the legs but not the arms. In some cases, SCI also disrupts the autonomic nervous system, which regulates visceral functions such as blood pressure, heart rate, body temperature, and digestive processes. SCI can cause secondary complications including pressure sores, muscle spasms, loss of muscle strength, cardiovascular disease, and osteoporosis. These SCI-related health problems can sometimes be improved by restoring motor functions such as grasping, standing, and walking. Restoring lost motor function can also markedly increase the quality of life enjoyed by individuals with SCI [2]. For these reasons, improving movement in individuals with SCI has been an active field of research for more than 40 years.

Aside from conventional physiotherapy and occupational therapy, the most commonly used technique for improving motor function in SCI individuals is functional electrical stimulation (FES), which uses short electrical pulses to generate contractions in paralyzed muscles. These contractions can be coordinated to

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actuate joints by stimulating one or more muscles that exert torques about the joint. The resulting joint angle can be controlled by modulating the intensity of stimulation delivered to the flexor and extensor muscles, which actuate the joint in opposite directions. A wide range of FES systems exist for individuals who have SCI, including grasping [3], [4], standing [5], [6], stationary rowing [7], and cycling [8], [9] as well as systems for reinforcing gait patterns during walking [10]–[12]. Figure 1 shows an FES grasping system in use.

Most FES systems that are used in clinical settings (outside of research laboratories) use either open-loop or finite-state control systems. Open-loop FES systems require continuous or repeated user input, which means that the user must devote his or her full attention to operating the FES device. Examples of open-loop controlled FES systems include the Compex Motion stimulator (Compex SA), the Free Hand system (Neurocontrol Corporation), the ParaStep system (Sigmedics, Inc.), and the H200 (BIONESS, Inc.). Finite-state FES systems execute a preset stimulation sequence in an open-loop fashion when a specific condition is met. This type of FES system can be used to improve the gait of stroke patients who struggle with drop foot, a common symptom in stroke patients. Drop foot is characterized by dragging the foot of the affected leg. A finite-state FES system for correcting foot drop uses a sensor to detect when the user's heel lifts off the ground and then stimulates the ankle flexor muscles, allowing the user to swing his or her foot and take a step. The Odstock dropped foot stimulator (Salisbury FES), the WalkAide foot drop stimulator (NeuroMotion, Inc.), and the L300 (BIONESS, Inc.) are finite-state FES systems that are used to correct foot drop. Finite-state control systems are technically closed-loop systems, since they include feedback to monitor the condition of interest. However, FES systems that use finite-state controllers typically do not correct for model errors or disturbances. For this reason, we group finite-state and open-loop FES systems together in this article.

Open-loop and finite-state FES systems are effective for correcting foot drop in stroke patients [13] and simple grasping tasks [14]. Also, commercial closed-loop FES cycling products are available, including the RT300 (Restorative Therapies, Inc.) and the Ergys 2 (Therapeutic Alliances, Inc.). These products use closed-loop control to keep the cycling cadence constant as the muscle begins to fatigue by increasing the amount of stimulation delivered to the muscle. However, many potential applications of FES technology for individuals with SCI require more sophisticated real-time control of the stimulation as well as closed-loop compensation for modeling errors and disturbances. These applications include FES systems for balancing during standing, torso control during sitting, and walking. Also, closed-loop FES systems require less user interaction than many of the currently available systems, thus facilitating tasks such as FES-assisted balance while standing. However, the response of muscles to electrical

Recommended Reading

To learn more about functional electrical stimulation and its application to spinal cord injuries, we recommend the following books:

- J.K. Chapin and K.A. Moxon, *Neural Prostheses for Restoration of Sensory and Motor Function*. Boca Raton, FL: CRC, 2001.
- D.B. Popovic and T. Sinkjaer, *Control of Movement for the Physically Disabled*. London, UK: Springer, 2000.
- L.L. Baker, C.L. Wederich, D.R. McNeal, C. Newsam, and R.L. Waters, *Neuromuscular Electrical Stimulation—A Practical Guide*, 4th ed. Downey, CA: Rancho Los Amigos National Rehabilitation Center, 2000.

stimulation, which is nonlinear, time varying, and coupled, is often accompanied by unpredictable perturbations in people who have SCI [15]. Also, the sensors that are required for feedback can make closed-loop FES systems cumbersome and time consuming to attach and remove. Consequently, closed-loop control strategies have not gained ground in clinical applications of FES technology.

The purpose of this article is to introduce the theory and techniques of FES and to present the main challenges to closed-loop FES control. We briefly discuss the physiology of skeletal muscle and then describe the methods used for FES. "Recommended Reading" provides a list of resources for more detailed information. We also present the special considerations involved in using FES for individuals who have SCI and then discuss the development of control systems for FES applications. Last, we survey
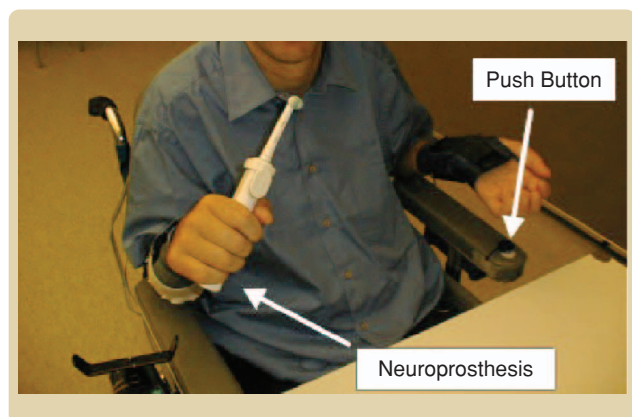

FIGURE 1 A functional electrical stimulation (FES) system in use. An individual with a quadriplegic spinal cord injury practices grasping and releasing objects using an FES system. Without the FES system, this individual cannot grasp or release objects. However, by pressing a button located on the wheelchair arm rest with his left hand, he can trigger the stimulator (not shown) to electrically stimulate the forearm muscles. The stimulation causes the muscles to contract in a grasping pattern, allowing him to pick up the toothbrush. The neuroprosthesis is incorporated into the brace that the individual wears on his right arm.

some of the work done to date in this area and discuss opportunities for future research. Our goal is to inspire researchers in the automatic control community to take up the challenge of developing closed-loop control algorithms for clinical FES applications.

SKELETAL MUSCLE PHYSIOLOGY

The principal functions of skeletal muscle are to maintain body position and produce movements. Skeletal muscle activity is controlled by the nervous system in neurologically intact individuals. Voluntary skeletal muscle contractions are initiated by electrochemical signals from the brain, which propagate through the nervous system to a motor neuron. Each motor neuron, together with the inner-

FIGURE 2 Summation of tension in motor units. The production of tension in skeletal muscle is accomplished by sequentially stimulating adjacent motor units, abbreviated "MU" in the figure. The stimulation is timed by the intact neurological system so that each motor unit contracts before the previously stimulated motor unit relaxes completely. The tension in the overall muscle is the sum of the tensions in the individual motor units. (Adapted from [22].)
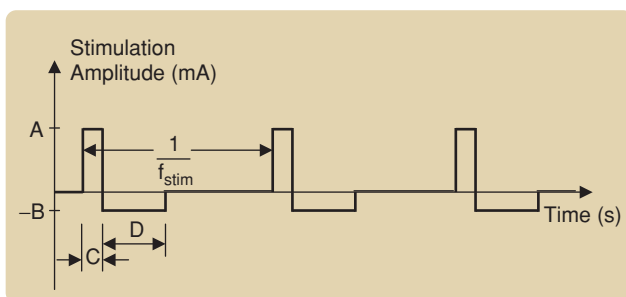
FIGURE 3 Stimulation pulse train. A typical stimulation waveform used for transcutaneous FES is a biphasic square-wave pulse train with a frequency of 20–40 Hz, an amplitude of 0–120 mA, and a pulse duration of 0–300 μs. A biphasic waveform is used because it induces charge transfer into the tissue and then immediately induces charge transfer out of the tissue. This pattern of charge transfer prevents galvanic processes that can cause tissue damage [18]. Notice that the amount of charge transferred into the tissue (given by the product AC) is the same as the charge transferred out of the tissue (given by the product BD).

vated muscle fibers, forms a motor unit. Muscles that are involved in gross movements, such as leg muscles, have few motor units, each of which includes a large number of muscle fibers. Muscles that are involved in fine movements, such as facial muscles, have many motor units, each of which includes a small number of muscle fibers. The body achieves gradations of muscular effort by recruiting varying numbers of motor units. For example, a task requiring maximum muscular effort uses almost all of the motor units in the muscle, resulting in a large portion of the muscle fibers being activated.

A single impulse in a motor neuron results in a fast, transient contraction of a single motor unit. Therefore, a motor neuron must deliver a train of impulses to its associated muscle fibers to maintain a constant contraction in the portion of the muscle served by the motor unit. The frequency of the pulses determines the intensity of the resulting muscle contraction. Typically, voluntary muscle contractions involve sustained, constant tension. The body achieves this constant tension, known as a tetanic contraction, by activating adjacent motor units at a frequency of 6–8 Hz in a sequential manner, so that one motor unit delivers a contractile impulse to its muscle fibers before the adjacent motor unit relaxes from the previous activation impulse [16], as shown in Figure 2. This method of sustaining a muscle contraction, known as asynchronous recruitment, allows the various motor units to share the work of maintaining a muscle contraction. Asynchronous recruitment ensures that the muscle fatigues slowly, since each motor unit is active only part of the time.

The main classes of muscle fibers are fast-twitch and slow-twitch fibers. Fast-twitch fibers respond quickly to a contractile impulse but also fatigue quickly. Slow-twitch fibers are more fatigue resistant than fast-twitch fibers but respond more slowly than fast-twitch fibers. The ratio between fast-twitch and slow-twitch fibers in a muscle depends on the function of the muscle. Also, the composition of a muscle can change over time depending on the type of contractions to which it is subjected [17]. For example, sprinters develop a higher proportion of fast-twitch fibers in their leg muscles than distance runners. This difference in muscle composition occurs because sprinters need muscles that respond quickly but don't need the muscles to work for an extended period of time, whereas distance runners need muscles that are fatigue resistant.

Muscle that does not receive regular exercise undergoes disuse atrophy and converts to a higher proportion of fast-twitch fibers than active muscle [17]. Many people with SCI have extensive disuse atrophy in their affected muscles [18]. Consequently, the affected muscles are weak and fatigue quickly and have become mostly fast-twitch fibers. However, disuse atrophy is often a reversible process since the affected muscles can be retrained with electrically stimulated weight-bearing exercise to increase their strength and fatigue resistance [18].
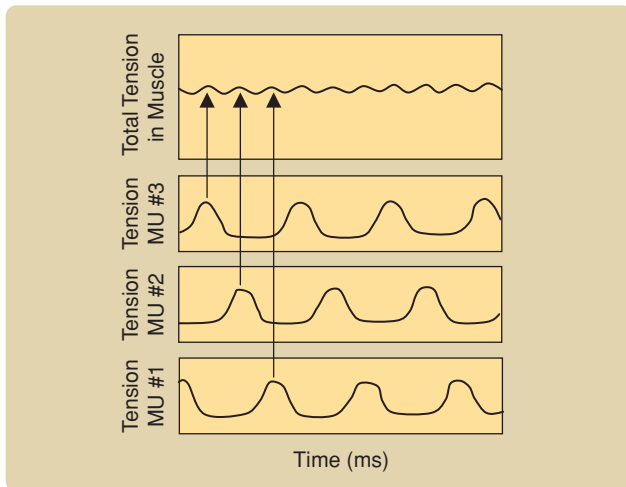
FUNCTIONAL ELECTRICAL STIMULATION

Functional electrical stimulation involves artificially inducing a current in specific motor neurons to generate muscle contractions. The neuron receives a series of short electrical pulses that are delivered using electrodes [15]. These electrodes can be transcutaneous (placed on the skin surface), percutaneous (placed within a muscle), epimysial (placed on the surface of the muscle), or cuff (wrapped around the nerve that innervates the muscle of interest) [18]. The tension produced in electrically stimulated muscle depends on the intensity and frequency of stimulation. The stimulation intensity is a function of the total charge transferred to the muscle, which depends on the pulse amplitude, duration, and frequency as well as the shape of the pulse train. A typical stimulation waveform is shown in Figure 3. Figure 4 illustrates the differences between the production of tension in neurologically intact and SCI individuals. The resulting torque about the joint that is actuated by the muscle depends on the tension in the flexor and extensor muscles as well as factors such as the biomechanics of the joint.

The angle of a joint, or, alternatively, the torque produced about a joint, can be regulated by varying the tension produced in the flexor and extensor muscles of the joint. Consequently, the joint angle or joint torque can be controlled by modulating the pulse amplitude, pulse duration, or frequency of stimulation. Typically, either the pulse duration or the amplitude of stimulation is controlled. FES recruits motor units in a synchronous manner, unlike the asynchronous recruitment of motor units that occurs in the intact nervous system. Synchronous recruitment means that FES stimulates all of the motor units at the same time, instead of rotating through the motor units as is done by the nervous system. For this reason, achieving tetanic contractions with FES stimulation requires a much higher stimulation frequency (20–40 Hz) than the frequency required to achieve tetanic contractions with the asynchronous recruitment per-

formed by the intact nervous system (6–8 Hz). This higher stimulation frequency is the main cause of the increased rate of fatigue associated with stimulated muscle contractions as compared to contractions initiated by the central nervous system (CNS) [17].

Furthermore, FES is believed to recruit the fast-twitch fibers before the slow-twitch fibers. This order of fiber recruitment, known as nonphysiological recruitment, is the opposite of natural muscle-fiber recruitment order, in which the slow-twitch fibers are recruited first. Nonphysiological recruitment happens because the fast-twitch fibers are innervated by axons with a larger diameter than the diameter of the axons of the slow-twitch fibers. These large-diameter axons couple more of the electric field than the small-diameter axons of the slow-twitch fibers, so that fast-twitch fibers respond to FES at lower stimulation levels than slow-twitch fibers. Since fast-twitch fibers fatigue more
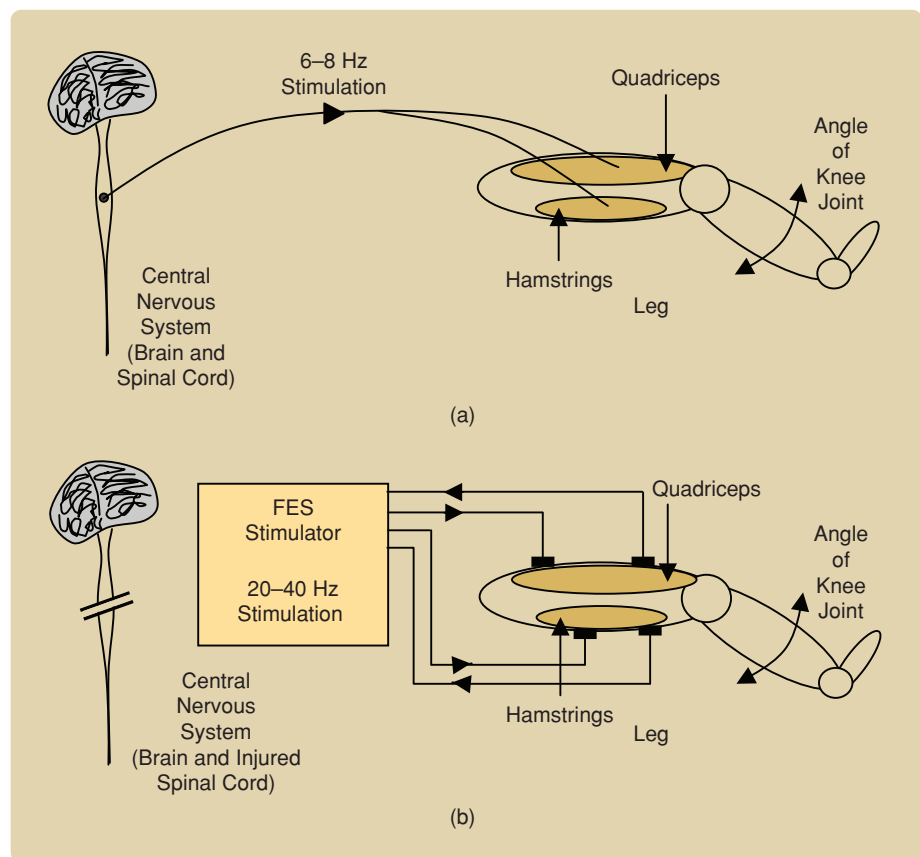

FIGURE 4 Production of tension in neurologically intact and spinal cord injured individuals. The angle of a joint, or the torque about that joint, can be regulated by varying the tension produced in the flexor and extensor muscles that actuate the joint. For the knee joint, the flexor muscles are the hamstrings group, while the extensor muscles are the quadriceps group. The hamstrings flex the knee to a bent position, while the quadriceps extend the knee and straighten the leg. (a) The intact neurological system produces tetanic contractions, which are characterized by sustained, constant tension, by stimulating each motor unit at a frequency of 6–8 Hz. Adjacent motor units are stimulated sequentially so that the overall muscle produces a tetanic contraction. If the muscle tension produced by the tetanic contraction is sufficiently high, the knee angle changes, as shown. (b) A functional electrical stimulation (FES) system can produce tetanic contractions in a spinal cord injured subject. However, the system must stimulate at 20–40 Hz to achieve this result because the individual motor units cannot be stimulated sequentially with FES.
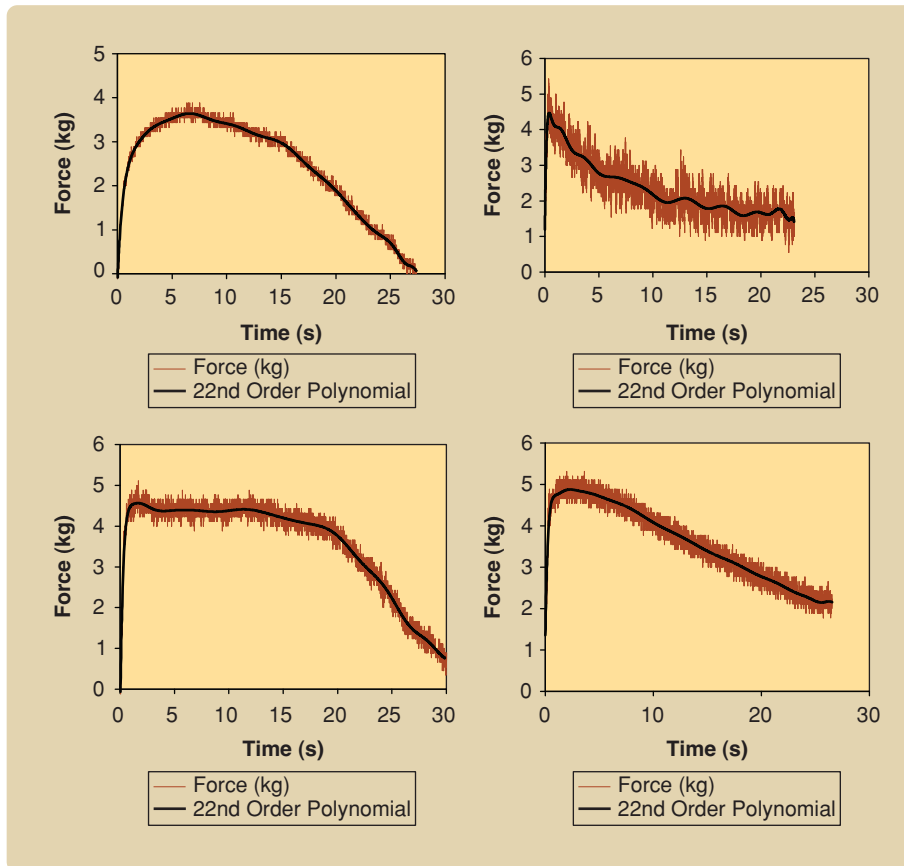
FIGURE 5 Force exerted by quadriceps versus time. These plots show the force exerted by the quadriceps muscle group when subjected to electrical stimulation. The data were collected from an individual who had sustained a spinal cord injury resulting in complete paralysis of the lower extremities [56]. Four different force decay profiles are shown, representing the wide range of force profiles that can be seen as the muscle fatigues in response to stimulation.

fatigue patterns that are typically seen in SCI subjects during FES sessions.

SCI AND FES

SCI can result in partial or total loss of motor function to the parts of the body below the level of the injury. The secondary complications associated with SCI and that strongly affect control system design are spasticity and hyperactive feedback loops in the CNS. Both of these phenomena affect closed-loop FES control when stimulating paralyzed muscles.

Spasticity, which is a common phenomenon in individuals with SCI, is characterized by varying degrees of increased muscle tone and hyperactive reflexes [18]. In the absence of control signals from the brain, the paralyzed muscles can develop a tendency to maximally contract in response to a wide range of muscular or sensory stimuli. A paralyzed muscle does not exhibit voluntary contractions because it receives no signals from the brain. However, the muscle can still contract in response to other stimuli, such as reflexive responses to pain. The presence of spasticity must be taken into account when designing FES control systems, since the controller must be able to compensate for the increased muscle tone of spastic muscles.

Spinal reflexes are feedback loops in the CNS [16]. Some spinal reflexes are simple reflexes that involve a single synapse between two motor neurons, whereas others are complex, multisynaptic reflexes. An example of a simple spinal reflex is the flexion withdrawal reflex, which causes a limb to rapidly recoil toward the body when a pain stimulus occurs at the extremity. Complex CNS feedback loops can involve neural circuits and oscillators formed by the connections between neurons in the spinal cord.

Spinal reflexes that originate in the spinal cord below the level of the injury often remain intact in people who have SCI [23]. These reflexes might be activated during FES-induced muscle contractions, causing strong perturbations. For example, walking is controlled in part by a neuronal circuit located in the spinal cord called a central pattern generator (CPG) [16]. When a paralyzed individual uses FES for walking, a CPG might generate a walking

quickly than slow-twitch fibers, the nonphysiological order of recruitment that occurs with FES also contributes to the increased rate of fatigue seen with artificial stimulation, as compared to natural stimulation. Recent work challenges the idea that FES causes nonphysiological recruitment of muscle fibers [19]. This research may yield new approaches to reducing fatigue in electrically stimulated muscle.

Currently, there is no way to avoid the synchronous motor-unit stimulation or nonphysiological recruitment that occurs with FES when transcutaneous electrodes are used. Several selective nerve-blocking techniques that aim to provide more natural motor-unit stimulation are available [20], [21]. However, these techniques require nerve cuff electrodes, which must be surgically implanted in the user. Regardless of the type of electrode used, it is possible to increase the fatigue resistance of electrically stimulated muscle by intensive muscle training using FES [14]. This training causes an increase in the strength and volume of the muscles, which increases the fatigue resistance of the muscle.

When an electrically stimulated muscle begins to fatigue, its response changes nonlinearly [22]. Eventually, the muscle can no longer produce a contraction. Figure 5 shows four
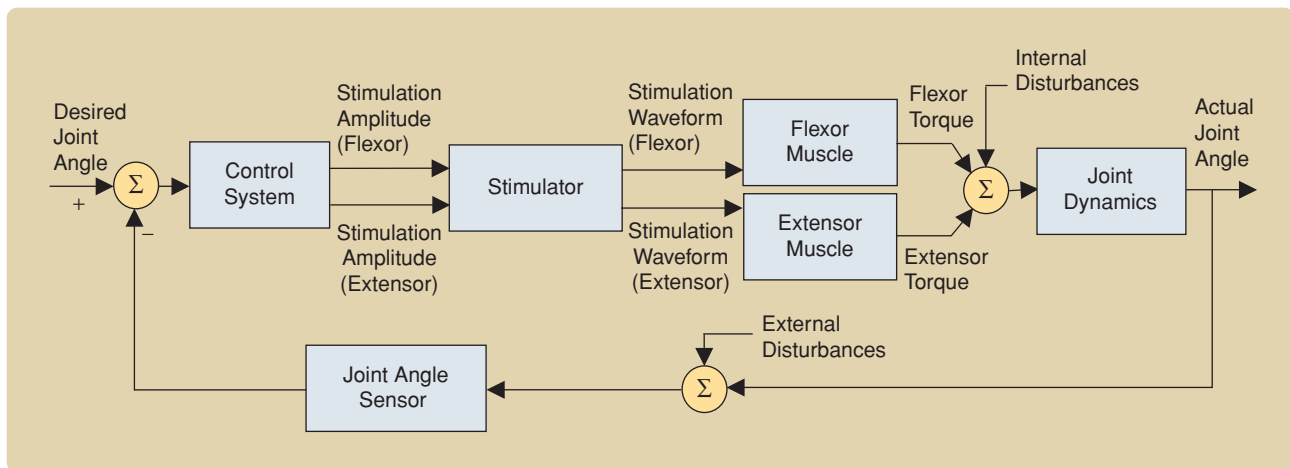
FIGURE 6 Functional electrical stimulation (FES) control system. This generic closed-loop FES system regulates joint angle by controlling the stimulation delivered to the flexor and extensor muscle groups. The input to the controller is the error between the desired and actual joint angles, while the controller's output is the required stimulation amplitude for each muscle group. The stimulator delivers a biphasic waveform with the required amplitude to each muscle group. A delay of 10–50 ms occurs between the onset of stimulation and the production of tension in the muscle [25], [26]. This delay is due to the biochemical processes that transduce electrical stimulation to muscle tension. When the stimulation causes sufficient tension in the muscle to overcome any opposing torques acting on the joint, the resulting change in joint angle is detected by the sensor in the feedback path. Although internal disturbances due to muscle spasms and spinal reflexes are shown as additional torques acting on the joint, these inputs are often treated alternatively as exogenous control signals. Also, external disturbances, such as the limb being bumped, might perturb the joint angle, as shown in the figure.

pattern [24] that conflicts with the pattern of muscle contractions induced by the FES walking system. CPGs and other spinal reflexes affect closed-loop FES control because these phenomena effectively act as exogenous control signals sent to the paralyzed muscles in parallel with the FES control signal. Since these exogenous control signals can disrupt the desired joint movement, the FES control system must compensate for these unintended control signals.

DEVELOPING CONTROL SYSTEMS FOR INDIVIDUALS WITH SCI

Interactions among parts of the body under voluntary control and parts of the body affected by the injury must be considered when designing FES systems for individuals with SCI. For example, an FES system that stimulates the lower extremities to maintain balance while standing must take into account voluntary movements of the user's arms and torso that can affect balance. However, since every FES system presents a different set of system-level considerations, we restrict our discussion to joint-level control.

Controlling joint angles or torques by regulating the amount of stimulation delivered to the muscles is a challenging problem. The response of stimulated muscle is nonlinear and time varying due to fatigue, nonphysiological recruitment of muscle fibers, and changing muscle composition due to regular FES use. For individuals who have SCI, spinal reflexes and perturbations due to spastic muscle contractions add another layer of complexity to the response of muscles to FES.

Regulating joint angles or torques with FES also involves controlling a highly coupled system because each

joint is actuated by at least two muscle groups, flexor muscles and extensor muscles. Furthermore, many muscles are biarticular, meaning that they actuate two joints simultaneously. For example, contractions of the quadriceps muscle can change both the knee and hip joint angles. Moreover, the maximum force that can be exerted by a muscle is a function of its length and the rate of change of its length [16], both of which can vary with joint angle.

Also, there is a time delay of 10–50 ms between stimulation and the onset of a muscle contraction, depending on the stimulation parameters and hardware, as well as the muscle being stimulated [25], [26]. This delay is in addition to the processing and transmission delays involved in the electrical stimulation system, which are typically on the order of 1 ms. Figure 6 shows a block diagram of a generic closed-loop FES system.

Models of the Response of Electrically Stimulated Muscle

Various models of the response of skeletal muscle to electrical stimulation are available [27]–[29]. These muscle models are divided into two types. The first type consists of physiological models, which attempt to model the physiological structure and behavior of muscle. Physiological models tend to be accurate, complex, and specific to a particular subject. Also, the values of some of the anatomical and physiological parameters of these models can be difficult to obtain. The second type of muscle model consists of empirical or black-box models. Although these models attempt to reproduce the input-output behavior of real muscle, their structure does not necessarily reflect the physiology of muscle.

The two most commonly used muscle models are the Hammerstein [30] and Hill models [31]. The Hammerstein model, shown in Figure 7, is an empirical muscle model, which consists of a static nonlinearity representing muscle-fiber recruitment, followed by a linear dynamic model. The linear part of the model represents the contraction dynamics of the muscle in response to electrical stimulation. Although the Hammerstein model is accurate under constant stimulation conditions, it is not accurate across the entire range of typical stimulation parameters [30].

The Hill muscle model, which is also an empirical model, attempts to model the response of electrically stimulated muscle using mechanical elements such as dashpots and stiffnesses, as shown in Figure 8. The Hill model in Figure 8 does not account for the effects of muscle length or rate of change of muscle length on force production, and thus its accuracy is reduced for electrically stimulated muscle contractions that result in movement [18]. More complex Hill models that include muscle length are given in [25]. Neither the Hammerstein nor Hill models account for fatigue. This neglect is a problem when using these models in FES control design, because the force output of muscle decreases with fatigue [18], [32].

Recently, Riener's physiological model has gained acceptance in the FES community [29], [33]–[36]. This model describes the dynamics of the knee in response to electrically stimulated muscle contractions and accounts for the nonlinear static and dynamic properties of 13 skeletal muscles as well as tendon tissues. Muscle fatigue and passive muscle velocity are included in Riener's model, along with the nonlinear dynamics of the body segments. Riener's model is based on experimentally obtained data and was tested with five untrained paraplegic patients [29]. As noted in [29], the model's complexity is its main limitation, because the identification process required to tune the model for a particular subject is time consuming.
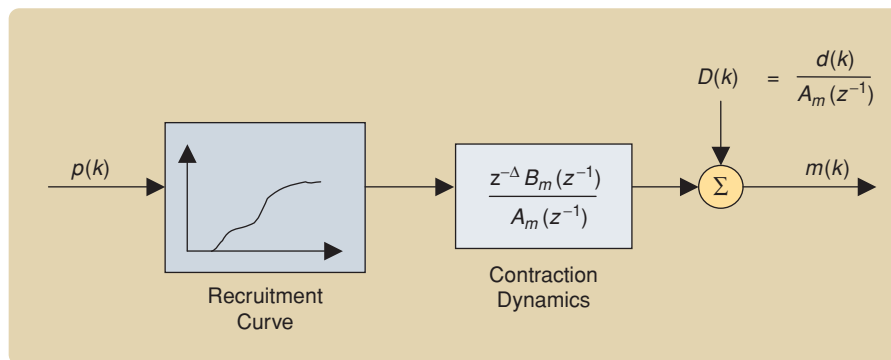

FIGURE 7 Hammerstein muscle model. This model (adapted from [30]) relates the duration of the stimulation pulse to the moment generated by the muscle. The recruitment curve is a nonlinear function that describes the proportion of muscle fibers recruited by stimulation of a particular intensity. The Hammerstein model assumes that the stimulation frequency and amplitude are constant, and that the pulse duration varies. Both the recruitment curve and linear dynamic model, which describes the contraction dynamics of the muscle, must be identified individually for each muscle. In this diagram, $p(k)$ is the stimulation pulse width, $m(k)$ is the muscle moment, $D(k)$ is a disturbance signal, $d(k)$ is zero-mean white noise, z^{-1} is the delay operator, Δ is an integer time delay greater than or equal to one, and $B_m(z^{-1})$ and $A_m(z^{-1})$ are polynomials in the delay operator.
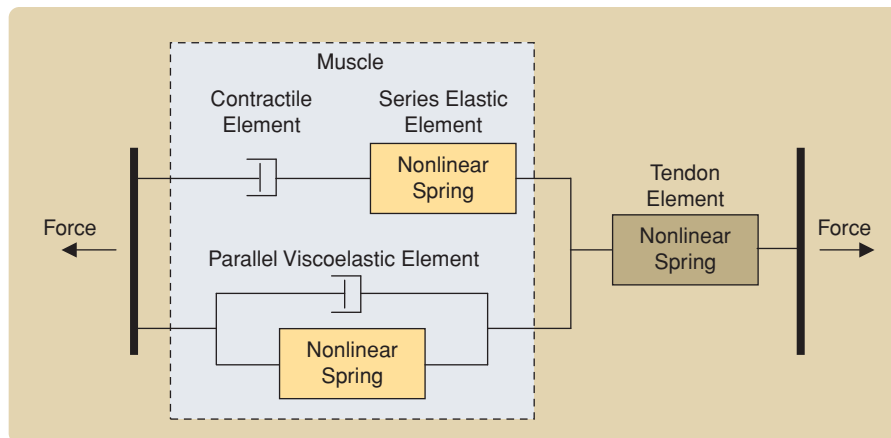

FIGURE 8 Hill muscle model. This model (adapted from [18]) uses mechanical elements to represent the viscoelastic properties of the electrically stimulated musculotendon system. The muscle model consists of an elastic element representing the tendon along with a combination of series and parallel viscous and elastic elements representing the muscle. All of the springs in this model are nonlinear elastic elements.

Considerations when Testing Closed-Loop FES Control Systems

Testing closed-loop FES control systems involve several considerations that are not a concern when testing controllers for more traditional plants. Most importantly, the FES controller must be initially tested in isolation from voluntary muscle contractions. This practice ensures that any muscle contraction is the result of action taken by the controller, not by the experimental subject. This isolation can be achieved by recruiting subjects who have a complete SCI. Individuals with complete SCI have no voluntary motor function below the level of injury and therefore cannot contract a paralyzed muscle during controller testing. Once the efficacy of a control method has been established in this way, the controller can be applied to FES systems for individuals who have other types of SCI.

Typically, subjects with SCI have to be trained before controller testing begins to increase the fatigue resistance of the paralyzed muscles. This training results in muscles that more closely approximate the muscles of a chronic FES user and therefore provide a more realistic testbed for a clinical FES controller than untrained muscle [37]. Furthermore, after being injured, the majority of SCI patients enter a state of spinal shock that can last several weeks. After the spinal shock subsides, patients undergo a period of recovery during which some voluntary functions might spontaneously return. Therefore, individuals must be neurologically stable before they are recruited as experimental subjects for initial controller testing to determine whether the subjects do in fact have a complete SCI. Typically, individuals are not neurologically stable until at least 12 months post injury.

The performance of closed-loop FES controllers must be tested using realistic joint-angle or joint-torque trajectories as well as perturbations that are representative of muscle spasms and reflexive muscle contractions. Standard performance measures must be reported, such as rise time, settling time, overshoot, steady-state error, lag, root-mean-squared (RMS) error, stability margin, and sensitivity analysis. These testing issues are sometimes neglected in the literature on closed-loop FES control, making it difficult to compare different controllers [38].

EXISTING STRATEGIES FOR CLOSED-LOOP CONTROL
Many strategies are available for closed-loop control of electrically stimulated muscle contractions. We describe several of these controllers to illustrate some of the challenges that must be addressed when designing control systems for closed-loop FES systems. We also present examples of FES controllers that can provide researchers with a starting point for future work. This section is intended to give the reader an overview of some of the issues involved in designing FES control systems. For additional details, see [39]–[43].

A controller for FES-based unsupported standing in a paraplegic subject is described in [44]. The objective of the controller is to maintain a hip angle of $0°$. The control algorithm consists of a proportional-integral differential (PID) controller in series with a nonlinear function that relates PID output (that is, the muscle force required to maintain the desired hip angle) to the duration of the stimulation pulses. Controller testing is done using two complete SCI subjects who participated in unspecified FES training for more than

six months prior to controller testing. The performance of the controller is evaluated by applying a disturbance that causes the subject to bend at the hip and recording how quickly the controller rejects the disturbance. The subject is also permitted to hold a fixed horizontal bar for support. The results of [44] show that the controller provides a 41% reduction in RMS error and a 52% reduction in steady-state error compared to open-loop control. Since the subjects' safety is of paramount importance, it can be difficult to design a safe study protocol that does not cloud the results of the experiment. For example, it is difficult to determine whether the results in [44] are due only to the action of the controller or whether the subject is voluntarily stabilizing himself or herself with the support bar.

An FES system for unsupported standing that maintains balance by stimulating the ankle flexor and extensor muscles to regulate ankle moment is described in [6] and [45]. The approach of [45] uses an H_∞ controller to regulate the moment about the ankles. H_∞ control guarantees stability when the nominal models of the plant and the uncertainties in the system are accurate and is able to compensate for perturbations that are included in the plant model. The results of [45] show that the H_∞ controller maintains stability during a series of ankle-moment tracking and disturbance-rejection tests in neurologically intact subjects. Researchers often test FES controllers on neurologically intact subjects since it can be difficult to recruit a sufficient number of subjects with SCI. However, neurologically intact subjects might skew the results of the experiment by subconsciously contracting their muscles. Therefore, it cannot be assumed that the results generated in this type of study are consistent with results for individuals with SCI. This problem is addressed in [45] by distracting the subjects during the experiment to reduce the likelihood of subconscious muscle contractions. The preliminary work in [45] is complemented by [6], which reports testing the H_∞ controller for regulating ankle moment with SCI subjects, as part of a nested control system for standing balance.

A neuro-PID controller for regulating knee angle in [46] uses an artificial neural network to map the nonlinear relationship between the desired knee angle and the required stimulation parameters and also uses a PID controller in a negative feedback loop to compensate for tracking errors caused by disturbances and modeling errors. The neural network is trained using a conjugate gradient algorithm, and the PID controller is tuned using the Ziegler-Nichols method. The controller is tested with one untrained SCI

subject, for which the neuro-PID controller achieves an RMS error of 5° when tracking a 1-Hz sinusoidal trajectory with amplitude 60° over four trials of unspecified length, with 60 s between trials. It can be difficult to collect a wide range of performance measures when testing FES controllers with human subjects since the subjects often have a limited amount of time to devote to participating in experiments.

Examples of Closed-Loop FES Controllers for Regulating Knee Angle

We now provide an overview of recent control techniques for regulating knee angle using closed-loop FES control. Although these techniques have not been proven in clinical FES applications, this work can provide a basis for further research, while giving a sense of the breadth of control techniques that have been applied to FES.

Four controllers for regulating knee angle are tested in [47]. These controllers, which are depicted in Figure 9, are
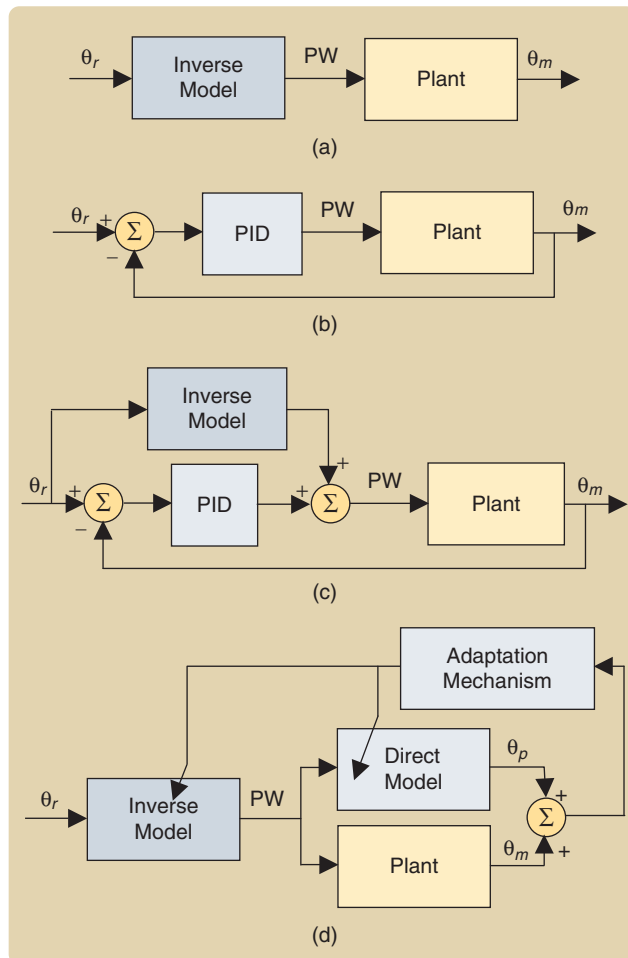

FIGURE 9 Control systems evaluated in [47]. (a) Open-loop controller. (b) Closed-loop PID controller. (c) Combination feedforward-feedback controller. (d) Adaptive controller. θ_r is the reference knee angle, θ_m is the measured knee angle, and θ_p is the knee angle predicted by the direct model. PW denotes the total pulse width delivered to the plant. (Adapted from [47].)

based on a nonlinear, physiological model of the knee's response to electrical stimulation of the quadriceps muscle. The control methods are (a) an open-loop controller that uses the inverse knee model as a compensator, (b) a closed-loop PID controller, (c) a feedforward-feedback controller that combines the inverse knee model with a PID controller, and (d) an adaptive controller that uses the inverse knee model to deliver a stimulation signal to both the plant and the direct model, so that the direct knee model functions as an observer of the plant. For (d), the error between the knee angle predicted by the direct model and the actual knee angle is minimized by varying two model parameters using an adaptive algorithm.

Two trained SCI subjects are used to obtain the results in [47]. The RMS errors for each controller when tracking a sinusoid are 1) 11.7°, 2) 6.0°, 3) 4.6°, and 4) less than 10° after 2 min of adaptation. The average lag for the same tracking task is reported as 1) 0.18 s, 2) 0.29 s, and 3) 0.18 s, where lag is not reported for the adaptive controller because the lag changes during the adaptation process. The results of [47] show that the combined feedforward-feedback controller performs best. However, it is noted in [47] that the inverse model is imperfect because it neglects noninvertible model components, such as time delays and saturation effects, which can degrade the performance of the feedforward-feedback controller.

A gain-scheduling controller for knee joint control is described in [36]. This controller uses a nonlinear autoregressive exogenous (NARX) polynomial model of the knee joint, which is identified using data from an untrained SCI subject. The NARX knee model is linearized at several operating points, and a linear quadratic controller is designed for each point. The knee angle is used as the scheduling variable for the controller, meaning that the knee angle is used to select the parameters of the linear quadratic controller. The gain-scheduling controller is tested in simulation using Riener's knee model [29]. The results of [36] show accurate tracking performance for the gain-scheduling controller with zero steady-state error and RMS error, zero overshoot, a rise time of less than 1 s, and lag of less than 1 s when tracking a 0.067-Hz sinusoid with an amplitude of 35°. The controller is able to reject knee angle disturbances of ±10° in about 2.5 s.

Because the knee model used for testing in [36] does not exhibit fatigue or spasticity, the results of testing the gain-scheduling controller could be verified with a real test subject. It might also be necessary to choose an alternative scheduling variable, since knee angle alone does not capture the nonlinearities of the system, as recommended in [48] when selecting a scheduling variable for gain-scheduling control.

A sliding mode controller for regulating knee angle is reported in [34]. Sliding mode control is a nonlinear control design technique that guarantees stability, provides tracking performance, and is robust to parameter variations. This technique uses a control law that causes the

state of the system to converge to a chosen sliding manifold in finite time and then evolve along the manifold toward a goal state [49]. In [34], the states of the system are the position and velocity errors, while the sliding manifold is a line that passes through the origin. The sliding mode control law causes the position and velocity errors to converge toward and then travel along the line toward the goal state, which is zero position error and zero velocity error. Riener's knee model is used as the basis for controller development in [34]. The sliding mode controller is tested using two paraplegic subjects. It is shown in [34] that the sliding mode controller remains stable in all experiments, and achieves an RMS error of $3.5°$ when tracking a step trajectory of $20°$.

CONCLUSIONS

Individuals who live with SCI seek FES systems that provide standing balance, torso control, and walking, as well as systems to prevent pressure sores while sitting. These requests are echoed by physicians and therapists who work with SCI individuals. FES systems that operate automatically are especially in demand so that the user can concentrate on other activities while using the system. These applications require closed-loop control to compensate for errors due to model inaccuracies and disturbances.

Designing a control system that tracks a joint trajectory or torque profile by regulating the electrical stimulation delivered to the muscles is a challenging problem. The response of muscles to electrical stimulation is nonlinear, time varying, and coupled, and is subject to strong perturbations due to muscle spasms and CNS feedback loops. Furthermore, rigorous testing methods must be used to verify the performance of these systems. In particular, experiments must be conducted using people who have complete SCI. This choice of experimental subjects ensures that any muscle contractions are due to actions taken by the control system and are not due to subconscious muscle contractions. Also, the subjects must be trained prior to testing so that their response to stimulation is representative of a chronic FES user.

It is important to evaluate closed-loop FES control systems using standard time- and frequency-domain performance metrics to facilitate the discussion of results between research groups. Moreover, uniform reporting of the performance of control methods expedites the process of developing clinically useful controllers by concentrating research efforts on promising control designs.

In this article, we have focused on FES applications that benefit individuals who have SCI. However, this technology is also used for rehabilitation after stroke [3], [50], [51] and traumatic brain injury [52]–[55] and can potentially be useful for managing the effects of other neuromuscular diseases and conditions.

We believe that the development of effective closed-loop control systems for FES applications is a vital step toward creating the next generation of FES systems, which have the potential to improve the lives of people with a wide range of physical challenges. We also believe that there is an opportunity for synergies to develop between the control systems and FES research communities with the purpose of developing closed-loop control systems for these new FES applications. We hope that this article inspires researchers in the control systems community to become involved in addressing this problem.

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