REVIEW ARTICLES

Functional Electrical Stimulation Control of Standing and Stepping After Spinal Cord Injury: A Review of Technical Characteristics

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ABSTRACT

Objectives. To investigate the different approaches in the field of functional electrical stimulation (FES) control of gait and address fundamental perquisites to enable FES walking systems to become safer, more practical, and therefore clinically efficacious. **Design.** Systematic review was conducted from electronic data bases up to March 2008. Studies with innovative control strategies were highlighted for analysis, but all relevant literatures were described to deliver a broad viewpoint. **Study Selection.** FES studies applying 1) open and closed-loop controllers; 2) control algorithm techniques; or 3) feedback information to the control unit of neuromuscular stimulators via biological signals or artificial sensors. These studies were mostly associated to FES gait. **Results.** By far, more spinal cord-injured users have benefited from open-loop FES walking systems because they have had an easier and faster setup. However, because of their limitations over the control of knee extension, closed-loop control of gait may be a superior approach. The use of electromyogram to quantify quadriceps fatigue was not considered sufficiently appropriate to predict knee-buckle events; instead, the use of motion sensors for such purposes is recommended. Finite state controllers based on a set of deterministic rules to process feedback signals seemed more suitable to provide accurate command-and-control compared with dynamic or neural network controllers. **Conclusions.** Progress in the development of closed-loop FES walking systems has been impeded by their lack of practicality. In the near future, this obstacle could be overcome via implanted systems, especially if using controllers based on deterministic rule sets derived from motion sensor feedback.

KEY WORDS: *Control, functional electrical stimulation, review, standing, stepping.*

Introduction

The use of functional electrical stimulation (FES) in the lower extremities has been well acknowledged to improve the health of spinal cord-injured (SCI) individuals (1,2). The first attempt to restore upright mobility occurred in 1961 when Liberson et al. developed the first FES neuroprosthesis for the correction of drop foot in hemiparetic patients by stimulating their common peroneal nerve (CPN) through skin-surface electrodes (3). In subsequent years, the basis of FES standing and stepping for neurologically impaired patients was firmly established, but because of technological limitations, portable ambulatory neuroprostheses could not be realized until the early 1980s, when multichannel devices under microprocessor control became available (4–6).

Despite the significant technological progress during the last 15 years, current commercial FES systems for stimulation of the paralyzed lower extremities still operate in an "open-loop" manner. This means that the stimulator's controller receives no direct feedback about the actual state of the system (i.e., the temporospatial position of the lower limbs). In general, neuromuscular stimulation is triggered and adjusted by means of buttons that are mounted on crutches or a walking frame. For the user, this represents the requirement of pressing buttons on a frequent basis, what could be inopportune considering that the hands need a firm grip on the frame handles and that an error in a pressing command could result in a fall.

The development of FES walking systems that are able to overcome these limitations would be an important step forward, and might improve the user acceptance of these devices outside of the research environment. Moreover, a FES neuroprosthesis should also be portable, fail-safe, easy to set up and operate, and possess sufficient processing power to compute large quantities of data in real time (7,8). Ideally, the state of the art in this field would include a fully implanted FES walking system (9,10). Developments in other biomedical fields such as cochlear implants and cardiac pacing have demonstrated that these requirements can be met, and suggest that neuromuscular stimulators to restore standing and walking are not using current technology to fullest potential (11–13).

Characteristics in the Control Design of FES Walking Systems

Open-Loop Systems

Open-loop FES systems are the simplest walking-assistive devices. By pressing buttons on a rolling frame, elbow frame-walker or crutches, the user can trigger stimulation sequences for standing up, stepping forward, or sitting down, as well as to increase/decrease the stimulation intensity (12,14–16) (Fig. 1). The Parastep system (Sigmedics Inc., Fairborn, OH, USA) works in this fashion by applying fixed stimulation sequences with only one set of stimulation parameters (24 Hz, 150 μsec) (10). Brissot et al. observed that with this system, 13 paraplegics could ambulate 2–350 m after an average of 20 training sessions (17). Graupe and Kohn reported that among approximately 400 Parastep users, most could walk 6–9 m, but some could ambulate up to 800 m (18). With the same system, Winchester et al. also reported that their five experienced FES users stepped with velocities of 4.6–24.3 m/min (19). Such a high inter-subject variability in gait velocity was also reported by Klose et al., but the authors noted that the factors producing this variability were not easily identified (20). Although differences in a user's level/type of injury and cardiorespiratory fitness certainly affect function, habituation and latency of CPN responses undoubtedly influences the quality of stepping.

Granat et al. examined the decrease in hip flexion responses over a certain number of CPN stimuli (21). Interestingly, they described attenuation levels in hip flexion angular excursion from 40 to 100% (i.e., complete attenuation; $N = 8$). In the same study, the CPN reflex latencies across their patients varied by approximately half a second. Other similar studies performed by the same group found equivalent results (22,23), suggesting that the fixed stimulation sequence provided by the Parastep might not be the most appropriate approach to control the swinging leg during stepping.

With the intention of delivering a more refined open-loop control of the leg swing, the Vienna FES system (Ottobock Healthcare GmbH, Vienna, Austria) applied personalized stimulation sequences for each patient (24). Stimulating quadriceps and gluteus muscles and CPN, Bijak et al. reported that smooth stepping motion could be achieved in all patients after only a few training sessions. Efficacy results with 12 thoracic paraplegics revealed a promising regularity on step lengths (20–30 cm), but considerable differences on walking distances (4–60 m) (25).
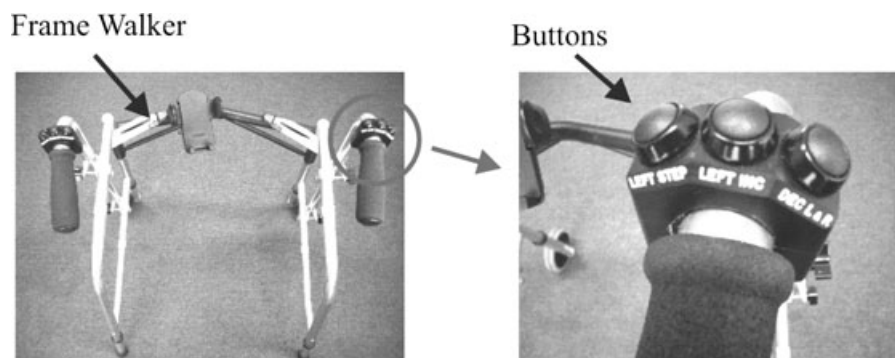

FIGURE 1. Walking frame with buttons to evoke functional electrical stimulation mobility tasks (Neopraxis Exostim).

However, in this study the subjects did not have the same period of training—seven subjects were deemed as experienced FES users and five as inexperienced.

To improve subjects' walking cadence, some authors introduced automatic triggering of stimulation sequences for stepping. Popovic et al. compared this approach at three different speeds against a hand-triggered operation (26). The range of speeds varied from 16.7 to 48.0 m/min. Although the faster walking speeds evoked physiologic costs up to 40% lower than the hand-triggered stepping, five out of six subjects still preferred the slower walking speed and one preferred the hand-triggered operation. Even though stepping at slower speeds was less efficient, the greater time to concatenate upper and lower body movements was postulated as the main reason for their choice.

Although promising results have been reported with open-loop FES walking systems, very few kinematic outcomes have been described in the literature (27). One major complication of this technique is the progressive fatigue of quadriceps muscles, which may result in dramatic knee buckling (28). Curiously, the occurrence of knee-buckle events impacting upon the safety and the quality of gait has not been previously reported. These limitations have led some researchers to consider "closed-loop" controllers—systems whereby feedback information is processed in real time and used to modify stimulation sequences so that a desired outcome may be achieved (29).

Closed-Loop Systems

Closed-loop control in FES walking has been used to generate controlled and target-directed movements (30). The current review considered two fundamental classes of feedback information comprising signals generated by biological processes and signals derived from artificial sensors.

Biological Feedback

To date, recordings of three different biopotentials have been the primary source for FES control of walking—muscle activity (electromyogram, EMG), brain activity (electroencephalogram, EEG), and nerve activity (electroneurogram, ENG) (31–33).

Electromyogram signals have been used for two different purposes: 1) to trigger stimulation sequences to initiate stepping; and 2) to monitor quadriceps fatigue on the stance leg.

Initiation of stepping is sometimes derived from EMG analysis of the contraction patterns of muscles above the lesion level (Fig. 2a). Graupe reported that via this strategy, patients could perform from 50 to 100 steps, with an error rate of less than 1% (34). The author proposed that during
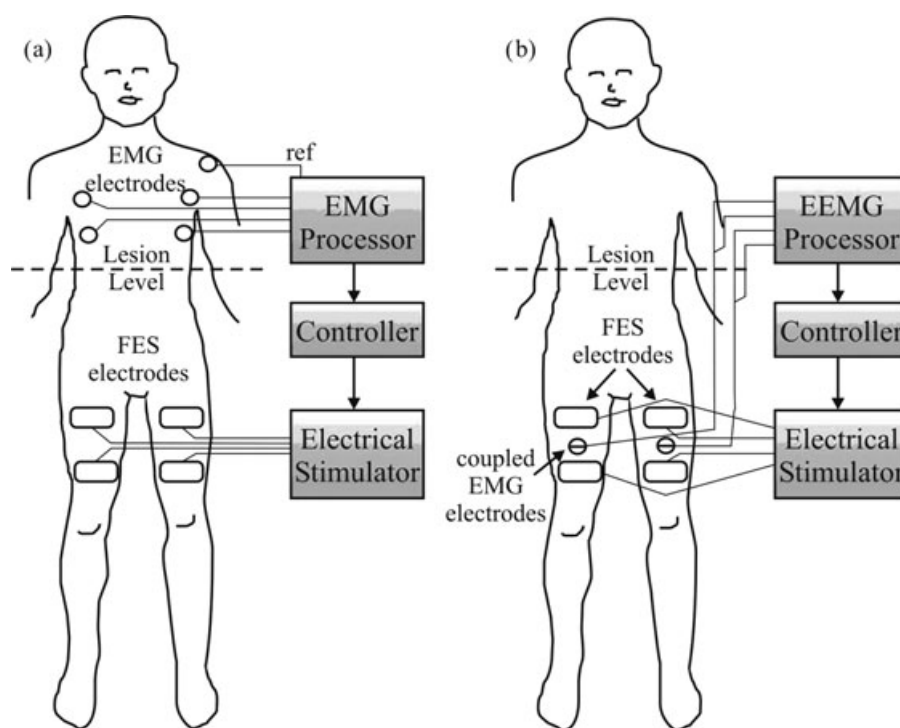

FIGURE 2. Schematics of (a) above-lesion and (b) below-lesion electromyogram (EMG) feedback (redrawn and adapted from (34)). FES, functional electrical stimulation.
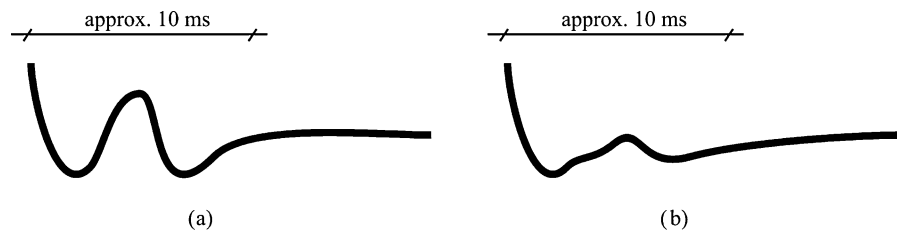
FIGURE 3. M-wave generated from a single electrical stimulation pulse from the quadriceps of a paraplegic subject while standing. (a) M-wave signal immediately after subject stood-up; (b) M-wave signal just before knee collapses due to quadriceps fatigue (redrawn and adapted from (34)).

the misactivation of a wrong step, no falls were observed because the subjects had the safety support of a walking frame. Even so, the use of this technique hardly seems justifiable in relation to the simpler and safer use of pressing buttons to initiate gait events.

Alternatively, the use of EMG signals recorded from the quadriceps can be used to assess fatigue during FES standing (Fig. 2b). These signals are often referred to as electrically evoked EMG (EEMG), because they resemble a strong single-unit action potential synchronized by the electrical stimulation (M-wave). Mizrahi et al. observed a progressive decrease of M-wave amplitude during sustained isometric contractions of the quadriceps (35). Although changes in the M-wave's shape could be used to predict knee buckling, the absolute level of fatigue before knee buckling has not been quantified (34), representing a weakness of this approach. Figure 3 portrays two typical EEMG signals recorded from the quadriceps muscle before and during the progression of muscle fatigue.

The use of EMG signals is an inexpensive and readily available technology. Nevertheless, its applicability in a "real-world" setting seems quite limited, because its deployment requires multiple pairs of EMG electrodes. Also, the user may have difficulties in reproducing the position of the EMG electrodes, resulting in a possible system malfunction (36).

In an attempt to overcome the limitations of EMG, some researchers have investigated the use of signals recorded from the brain as a trigger for predefined FES walking tasks. From EEG signals, movement-related potentials (MRPs) have been extracted and used as inputs to FES control. Sinkjaer et al. reported that the late components of MRPs coded the rates of torque development rather than the levels of torque achieved (37). This type of feedback information was considered more complicated as a control variable and the authors suggested that a better topographical resolution might be achieved using subcortical electrodes. However, this more invasive approach might only be possible after the development of more clinically acceptable technologies.

Another invasive means of feedback trialed in the lower extremities has been referred to as "natural sensors." Comprising implanted nerve cuff electrodes, ENG signals elicited from changes in contact forces of the associated limb/extremity are collected from sensory nerves. So far, this technique has been extensively tested in animal research models. Results have shown that detailed static and dynamic data from the forces applied on the footpad could be obtained (38–40). Hansen et al. reported that the detection of heel-strike and toe-off could be made without errors even after 392 days of testing in one hemiplegic subject (41). Their system also reliably detected the differences between walking and standing. By using feedback signals from nerve cuff electrodes, Hoffer et al. described the first implantation of the Neurostep FES system to correct foot drop (42). These authors' results demonstrated that after six months of training, a 70-year-old hemiparetic patient could walk 250 m. Moreover, Kerr and Hoffer assessed the feasibility of a laboratory-based system to control paraplegic posture during standing (43). Data about the centre of pressure of six able-bodied subjects were collected during sway and correlated to the corresponding innervation fields in the foot sole. With their system implanted in a pig, data could be collected from the pig's tibial nerve 94% of the time when similar pressures were applied on the pig's footpad.

Although artificial sensors are delivering reliable and reproducible results, we believe that the additional donning and doffing are contradictory to their widespread use. The development of safe and reliable fully implanted "natural" sensors is therefore considered a very important step toward greater acceptance of future FES walking systems. The abilities to resist the "hazardous" environment of the human body and to withstand sterilization before implantation are extremely relevant in this respect (44). Equally important is how changes in signal properties will be treated in the long term (39). In this regard, the use of implanted electronic sensors rather than use of physiologic signals for feedback may be a more practical option.

Artificial Feedback

Accurate temporospatial information about the lower limbs is critically important for the performance of any closed-loop controlled FES system. In the past, different technologies have been used to measure lower body kinematics and kinetics or to detect phases of the gait cycle. These include electrogoniometers (45), potentiometers (46), Hall effect transducers, optical fibers, polymer piezoresistive materials, piezoelectric transducers, and IC capacitive sensors (47). Despite their ability to perform measurements more or less precisely, most of these devices have been found to be time-consuming to set up and have generated data of limited reliability (47–51).

A renewed interest in artificial sensors was shown by many researchers when patient-mountable motion sensors became easily accessible (52). These new sensors—often referred to as "motion sensors"—mainly consist of inclinometers (or tilt sensors) (53), accelerometers (54–58), gyroscopes (59–61), or a combination of these (52,62–66). Williamson and Andrews tested the accuracy of a rate gyroscope-accelerometer sensor against a reference goniometer (64). Results during standing trials revealed mean differences of 2.1–2.4° in knee angles for two able-bodied subjects and one paraplegic subject standing with FES assistance. A similar sensor configuration was developed by Simcox et al. (65). The performance of their sensors was verified in multiple able-bodied sit-to-stand and walking cycles. The sensors' angular data were significantly correlated with the data from a 3D motion analysis system ($r = 0.90$–0.99), with root mean square (RMS) errors of less than 5° in the stepping cycle motion. This sensor was very compact ($63 \times 35 \times 10$ mm), which is an important factor in enabling more practical usage and even the future prospect of miniaturization and implantation (55). In a review article, Veltink suggested that wireless communication between sensors and controllers, and automatic in-use calibration of sensors were desirable characteristics in any closed-loop system (67).

A different type of artificial sensor, considered a substitute for ENG feedback in relation to foot-floor contact, is the use of force sensing resistors (FSRs). Deployed into shoes, FSRs detect stance, toe-off, and heel-strike, and can be used to automatically trigger the stimulation sequences for stepping (50). Based on feedback data from force transducers in the foot sole, de Castro and Cliquet employed skin-surface electro-tactile stimulation on the shoulders as a "closed-loop" system to control gait. This type of stimulation, which gave the sensation of a pen sliding on the skin, improved their subjects' upright posture because of the new perception of toe-off and heel-strike (68). Pappas et al. combined FSRs with a gyroscope measuring the angular velocity of the foot (61). This approach enabled the detection of swing phase of gait. The rate of success of their system was above 96% for subjects with impaired walking. Although FSRs have shown promise, a number of practical problems are not often mentioned in the literature such as thresholds having to be set for each subject, malfunctioning due to the foot moving within the shoes and device short-life (69,70).

Closed-loop feedback will benefit FES systems only if its control unit has an appropriate control algorithm. The following section reviews the main algorithm techniques that have been used in FES control of walking and at the same time discusses some of their advantages and drawbacks.

Control Algorithm Methods

In any FES walking system, the control unit uses an algorithm to process various input signals, such as button presses or feedback signals, with the aim of generating a predefined responses of the system's output. The output variable could, in a simple case, be an adapted stimulation pulse width or raised stimulation voltage, or in a more complex situation, a complete stimulation sequence. Dynamic controllers were one of the first algorithm techniques applied during FES walking.

Dynamic Controllers

In the past, dynamic controllers could be implemented with passive electronic components such as capacitors and resistors (Fig. 4). For this reason, they were used in closed-loop control as early as 1974 (71). For example, Veltink et al. tested both open-loop and closed-loop controllers based on three variables in cat muscles: non-linear muscle dynamics action, angle-torque, and angular velocity-torque relationships (72). The open-loop controller comprised a non-linear compensator, based on a modified version of Hill's muscle model, while the closed-loop controller used a proportional-integral-derivative (PID) controller. The authors achieved best results when the non-linear compensator was combined with the PID controller, because the PID controller compensated for modeling errors of the non-linear compensator. The range of RMS errors of the best controller was of $1.0 \pm 0.3\%$ ($N = 7$). During steady-state stance, one important aspect addressed by the authors was the easy saturation of the controller due to maximal stimulation being required to evoke small changes in joint torques (73).

Dynamic controllers are, by nature of their design, limited in the rate of transition toward the desired output signal. For this reason, the tracking error responses increase as the velocity of the desired movement quickens (74). Quintern et al. tested a PID controller in complete SCI patients to control knee-joint angles and torques. Good results were obtained for isometric conditions, but the controller could not efficiently control a freely swinging shank. The use of a compensator combined with this controller provided better accuracy of movement and no RMS errors were detected (75).
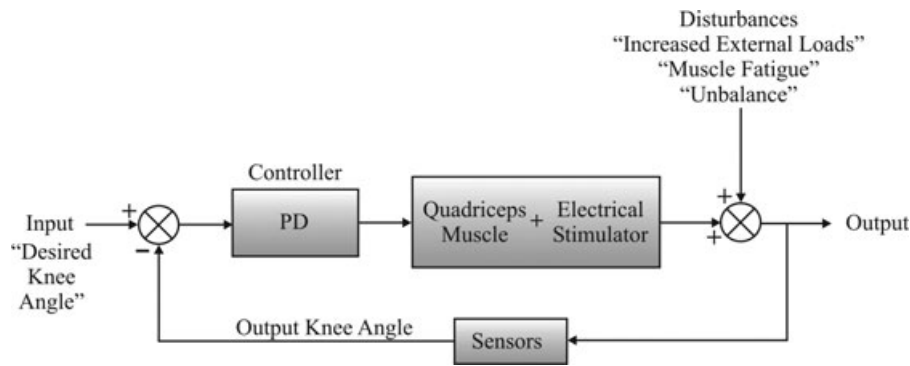
FIGURE 4. Schematic representation of proportional-derivative (PD) controller. In this example, knee angles are controlled via quadriceps muscle stimulation (redrawn and adapted from (95)).
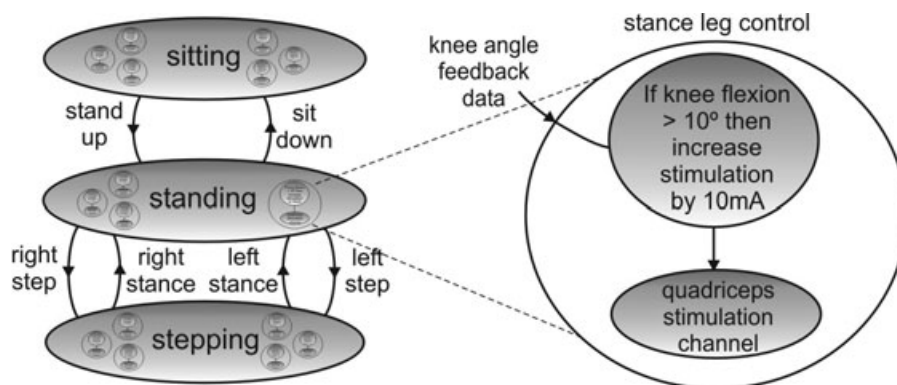

FIGURE 5. Schematics of a finite state controller model for functional electrical stimulation standing and stepping. The left panel represents the three states of the model—sitting, standing, and stepping. The right panel represents a one stance leg control rule within the standing state. Based on knee angle feedback, stimulation amplitude over quadriceps is increased by 10 mA when knee become flexed more than 10°.

Although relatively good results have been achieved with PID controllers, they have proved cumbersome to setup, requiring a "trial and error" during fine-tuning (76). This problem might have been predicted because inherent properties of the musculoskeletal system, such as latency, are more difficult to characterize than a dynamic controller applied to a machine, such as a robotic arm, for example. More recently, some of the limitations of dynamic controllers have been overcome by rapid development of microcontrollers and personal computer designs. This has permitted the implementation of a wide number of rules into one program or piece of equipment (12,77–79). This new programming flexibility has enabled more complex control strategies to be implemented faster and at an affordable cost.

Finite State Controllers (FSCs)

Finite state or rule-based controllers first appeared after the late 1980s and were subsequently deployed in a number of FES studies (80–83). As an FSC employs sequential logic and control functions based on finite functional states (84), a relatively simple, accurate design of the control algorithm is facilitated. Figure 5 portrays a representation of a typical FSC model for FES-evoked walking.

Davis et al. developed an FSC-based anti-knee-buckling strategy whereby stimulation amplitude was increased every time a knee-buckle over 10° was detected (45). Two T10 ASIA-A subjects using an implanted stimulator could stand between 30 and 60 min. Two sensorimotor complete, low-thoracic spinal lesion patients employed an implanted stimulator to stand for 30–60 min. Mulder et al. investigated a similar, but transcutaneous, FES approach whereby knee flexion of less than 5° triggered step increments up to the maximal amplitude of their stimulator (evoking knee locking), followed by a ramping down in stimulation until the knee unlocking was detected (85). When this strategy was assessed on two individuals with paraplegia, standing duration improved by three to five times compared with the

duration achieved with continuous maximal amplitude stimulation. Braz et al. used an analogous transcutaneous FES approach, noting similar results, but the stimulation amplitude changes deployed by their control algorithm were less abrupt (86).

In a review presented by Sweeney et al. (87), FSC systems were described as an effective and intuitive control strategy; however, their application was limited by the requirement for external sensors and electrodes. The authors proposed an integration of natural sensors and the use of implanted multichannel stimulators would be the next evolutionary step toward a genuinely usable FES-gait neuroprostheses for SCI individuals.

Finite state controllers using ENG recordings to control FES-evoked stepping has also been trialed using cats, whereby thresholds to switch states were based on fixed percentages of the peak ENG signals for a particular movement. The FSC design of Strange and Hoffer was tested on 14 adult felines with an average correct prediction of gait phases of approximately 95% (39). In both referred studies, the authors stated that although the use of simple rules were promising, the complexity in FES applications required more complex rules and an enhanced mapping of the neural and muscle activity recordings. Kostov et al. also pointed out some limitations in the rule-based FSC design, arguing that a high level of expertise in FES biomechanics was not available at the current time (84). In this scenario, FSCs with control rules created by techniques of artificial intelligence were developed.

Artificial Neural Networks (ANNs)

Because of their ability to generalize, to learn from experience, and to modify themselves, neural networks seem well-suited for FES control of stepping (88). Otherwise known as machine learning, the ANN's structure can be understood as a block of interconnected pseudo-neurons. The pattern of these interconnections determines the behavior of the network. By adapting the strength of the interconnections, the network can learn and the input–output properties are changed (89). Traditional ANNs apply binary values (0 or 1) for the weights of these connections, while more recent fuzzy neural networks apply multiple answers or a spectrum of values (90). This could be valuable in FES systems wherein the network response might declare that a muscle is not active or inactive, but partially active. Three-layered networks are the most commonly used network architecture (Fig. 6), since 3 is the minimum number of layers for a network to perform consistent approximations (91).

ANN-based controllers have been tested on many specific problems in FES walking. Heller et al. applied an ANN to reconstruct muscle activation patterns from kinematic and EMG data during able-bodied walking (92). Their ANN showed promise in predicting the EMG patterns of walking
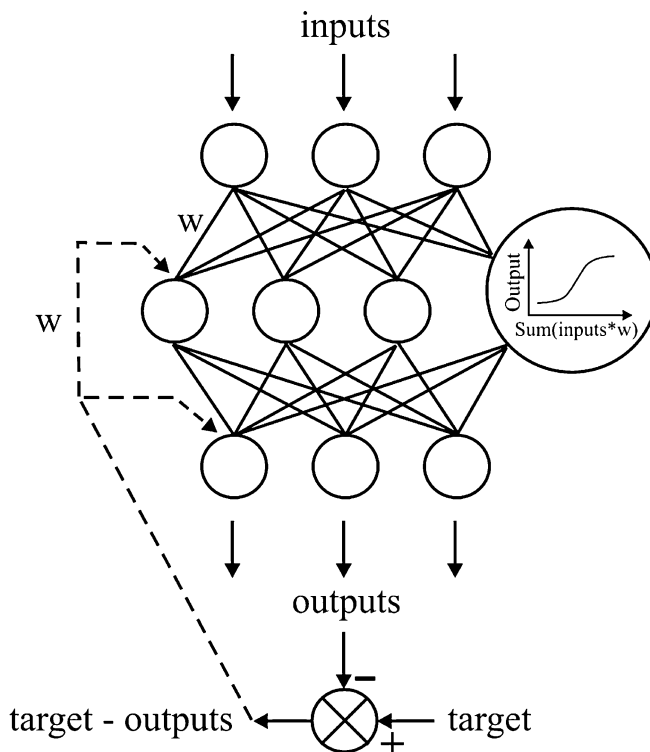

FIGURE 6. Schematic architecture of a three-layered neural network. Each neuron generates output based on the sum of the product inputs × weights (input * w) for a sigmoid activation function. Dashed lines indicate the influence of a learning algorithm based on a desired target response (redrawn and adapted from (91)).

at different speeds, but the network could not provide any insight about the biomechanical behavior of the system. Winslow et al. reported on the use of neural networks to predict increases of FES amplitude based on EMG feedback in an attempt to maintain a defined knee angle (93). Their data revealed an accuracy of greater than 78% when predicting a change in knee angle. Abbas and Triolo employed an ANN to generate muscle contractions to follow cyclic torque trajectories (94). The author's experiments showed average errors from 4.7% to 12.5% across each of the legs of four SCI subjects. To detect patient's current phase of gait, Skelly and Chizeck developed a fuzzy model based on feedback from FSR signals and phase of stimulation sequences (50). Tested on three SCI subjects, the range of errors also varied considerably across each of their subject's legs—from zero to 7.8% in detecting heel-strike and 3.2% to 12% to perceive toe-off.

Riess and Abbas analyzed the performance of a proportional-derivative (PD) against an ANN controller for the generation of cyclic movements (95). Results with both systems on two SCI subjects revealed that the ANN controller provided a better track of the desired knee angles (PD's RMS error = 45%; ANN's RMS error ≤10% after 30 cycles).

Similar results were found by Chang et al. when using a PID controller (96). In their study, a hybrid approach using a neuro-PID controller was also tested, providing slightly better results. An important observation about these ANN studies is that the very high RMS errors of the dynamic controllers might have been caused by improper setting of the controller variables in favor of their ANNs.

Surprisingly, although most of the studies in this critical review included human trials, none could be effectively applied in a clinical environment. Some reasons for poor clinical utility include: poor learning activation mechanisms and large computational time (91); realistic perturbations were not included in the models (90); inaccuracy or minimal output errors within the range of 5–10% (84,94,95,97,98); no efficient process to reduce network complexity (99) or to avoid modeling errors (96). As a result of these limitations, the outcomes of ANN-based controllers have not progressed much over the last 10 years, and in our opinion have some way to go before "real-world" deployment.

Discussion

Until recent times, open-loop FES walking systems have been the major system design promising benefits to the SCI population. Its acceptance in the clinical environment is undoubtedly related to its relatively simple donning and doffing. Although there are many studies reporting positive performance using open-loop systems in terms of walking distances, very few have quantified the quality of gait via biomechanical analysis. As a result, high inter-subject variability in stepping motion induced by either fixed or customized stimulation sequences has not been carefully investigated. Open-loop systems also require substantial user awareness and because of that, they are not fail-safe in terms of falls prevention, severe knee-buckle, and hip "jack-knifing." These negative sequelae of open-loop control of gait might be lessened by the real-time adaptability of the stimulation sequences using closed-loop control.

Similar to the open-loop designs, closed-loop systems also possess considerable drawbacks. Previous studies with SCI subjects have been limited to very small sample sizes and generally have not presented functional performance on walking distances or standing times. Although this could be justified by the nature of its more complex setup, the efficacy of many closed-loop controllers has not been often quantified, and the presentation of the results has been limited to the discussion of technical or engineering issues.

Closed-loop control via EEG has been demonstrated as a promising FES technique, although a better mapping of the motor cortex is needed (37). Improved technology is also required to permit safe surgical implantation of cortical electrodes (44). On the other hand, for EMG-based feedback (and notwithstanding the fact that information on fatigue would be highly desirable in almost all FES applica-

tions), there is still an unclear correlation between changes in the EEMG signals associated with the progression of fatigue and the occurrence of knee-buckle (34). This is also applicable to the ENG feedback which does not provide one crucial piece of information—the accurate spatial location of the lower limbs. ENG recordings from an innervated foot have demonstrated good results in the detection of toe-off and heel-strike (37); however, the use of FSRs provided shorter delays in the detection of those events (50). Both methods would be usable to provide command-and-control as a substitute to button pressing to initiate stepping. However, it is important to consider that safety, balance, and the ability to synchronize upper and lower body movements are appreciated by neuroprosthesis users (8,26). Because of the trunk instability after mid-thoracic SCI, the use of a walking frame or crutches remains necessary in any case. For this reason, mounted buttons to evoke stepping on these walking aids is clearly warranted, even for closed-loop systems. However, by using miniaturized motion sensors, the two major tasks of the control algorithm could focus on the control of knee extension during stance and the control of the swinging leg during stepping. Although this would improve the quality of gait, need for sensors would certainly compromise the practicality of any neuroprosthesis system. This might be less of a problem in an implanted device comprising sensors and stimulator.

The control of FES gait by means of dynamic controllers has demonstrated limited reliability. Although these controllers were able to sustain isometric muscle contractions, they have proven inaccurate during dynamic movements. In this regard ANNs have provided better results, but because of their quasi-deterministic behavior previous authors expressed reservations about their real-world performance (100). The use of these sophisticated techniques may be technologically interesting and theoretically promising, but the simplification of these models required by the complexity of the neuromuscular system after SCI trauma has obscured their usefulness (101).

Completely relying on deterministic control rules, FSCs permitted the development of a more accurate and robust algorithm design. Because of their modularity, tasks such as the in-use calibration of sensors or the determination of ENG thresholds could be treated independently, providing greater accuracy to the system, and therefore safety. Also, the processing of deterministic rules enables a faster response time when compared with ANNs, allowing the inclusion of a much greater number of control rules. Currently, FSC-driven neuroprosthesis demonstrates promise for advancement of FES-evoked standing and walking.

Conclusion

Although FES devices to restore upright mobility have undergone dramatic changes since their beginnings in the 1960s, they have not yet reached a technological potential

sufficient to benefit a great number of FES users worldwide. This will only be possible with the development of a reliable system that is simple to set up and to operate and that, more importantly, enable a satisfactory quality of gait.

To achieve this, the use of externally mounted miniaturized motion sensors and a FSC based on deterministic rules seems to be the most suitable approach at present. In the near future, its implantation should overcome the major practical obstacles often described in the donning and doffing of FES walking systems.

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Conflict of Interest

Authors have reported no conflicts of interest.

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