Neural Prostheses for Walking Restoration

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Abstract—We review the state of the art of multi-channel electrical stimulation (ES) systems with surface electrodes for assistance in the standing and walking of paraplegics and hemiplegics. For the group of complete paraplegics, walking achieved with available ES systems is below their expectations, especially since these systems cannot compete with mobility provided to them by a wheelchair. However, standing and walking with ES systems are beneficial because they contribute to the improvement of physiological functions. For individuals who can stand with some arm support (e.g., paraplegics with incomplete injury and hemiplegies), the current ES systems are an effective augmentation of walking. We suggest that an ES system for walking of incomplete paraplegies and hemiplegies will be better accepted if the stimulation is regulated by a rule-based control, that is, a preprogrammed, sensor-triggered activation of several muscles resulting in normal walking. We present a method to obtain muscle activity profiles from simulation of a customized model of the patient that can be used for the synthesis of rules for control.

Index Terms—electrical stimulation, neural prosthesis, paraplegia, hemiplegia, standing, walking

I. INTRODUCTION

MOBILITY of individuals with complete paraplegia and tetraplegia consists solely of wheelchair propulsion. However, it is important for many reasons to make these individuals stand and walk. Standing upright changes a person’s view of the world, and the ability to transfer between two points in space that are not easily reachable by wheelchair can make a tremendous difference in quality of life. Neural prostheses (NP) which use Functional Electrical Stimulation (FES) can induce standing and walking; therefore, they can contribute to overall health by improving several important physiological functions.

Bipedal standing and walking are challenging tasks from a mechanical viewpoint, since the body is a multi-segment inverted pendulum with many joints and muscles (actuators). Control is achieved by a highly complex biological neural network with hierarchical structure receiving numerous sensory inputs and using both parallel and serial descending command links to muscles. Gravity is the main force acting on the system; therefore, the muscles need to generate forces which will work against gravity (as occurs in standing). After spinal cord injury, the flow of sensory and command signals is interrupted.

In complete paraplegics, all sensors and muscles below the lesion can not communicate with the central nervous system above the injury. The paralyzed muscles and nerves are connected to the central nervous system (spinal cord); hence, when a motor nerve is externally activated, the corresponding muscle will contract, and when a sensory nerve is externally activated, reflex response will follow. These externally elicited contractions and reflex responses could be combined into sequences that are similar to normal movements. Based on this, it has been demonstrated that stimulation of extensor muscles can extend knee and hip joints, and with appropriate support of the ankles, a paralyzed individual could stand while maintaining her/his balance with some support for the arms (e.g., parallel bars, standing frame, crutches, etc.). Muscles must also move body segments in a manner that will create shift of the center of mass in the desired direction in order to progress in movement such as walking. The control of walking is becoming even more challenging since the support area during bipedal walking is reduced to a fraction of one sole during a single stance phase of the walking, and the body is at most times out of balance. During normal walking, inertial forces contribute to stability (e.g., gyroscopic moment of rotating segments), but in walking paraplegics the velocities and accelerations are very small, and the inertial contributions are minimal. So far, the only possibility is to add arm supports and change from bipedal to quadrupedal walking.

This summary directly suggests that FES-based NP can be implemented in incomplete paraplegics and hemiplegics who have some preserved naturally controlled balance and leg movements. The NP in incomplete paraplegics and hemiplegics should be considered as a device that augments the preserved abilities and provides a minimal frame that allows bipedal walking. The basic principle for operation of a walking NP is the activation of motor systems by means of controlled bursts of electrical charge delivered to motor and sensory nerves, thereby activating paralyzed yet innervated muscles or triggering reflex responses leading to movement that, when timed appropriately, improve functional ability.

II. MULTICHANNEL ELECTRICAL STIMULATION VIA SURFACE ELECTRODES FOR STANDING AND WALKING

An NP should be considered as a bypass of the impaired motor mechanisms. Natural mechanisms for motor control are described by Popović and Sinkjaer [1]. The assignment of an effective NP becomes obvious: NP must provide synergistic action of many muscles; provide full control over each of the muscles by following the findings about the
size principle, recruitment order, and recruitment rate; and it should also include sensory feedback for both operation of the system and cognitive awareness of the action.

The electrical stimulation is delivered to innervation pathways in the form of trains of pulses. The intensity of stimulation is directly related to the force generated. The regulation of the strength of a motor response is done through the number of active motor nerve fibers and the rate at which they trigger action potentials. In voluntary control, these two mechanisms are called recruitment and temporal summation, respectively. In a physiological contraction, the recruitment order is fixed; slow, fatigue-resistant motor units are more active at a lower voluntary effort than larger, fast, fatigable units. Most, if not all NP, simultaneously activate many motor units. In an electrically induced recruitment, the recruitment order is not known a priori, but depends upon the variables of position and geometry as well as of fiber size. An inverse order of electrically induced recruitment is typical when applying NP; the largest fibers are more easily excited as compared with small fibers. This implies that the recruitment has to be considered at all times in order to provide controlled and graded externally induced activation. The recruitment of nerve fibers with increasing stimulus pulse amplitude or duration is nonlinear. For this reason, a linear increase in muscle output force cannot be achieved by a linear change of the intensity of stimulation. The selection of the most effective parameter for regulation of recruitment has been studied [e.g., 2, 3], yet research is still needed to determine the optimal pattern of stimulation.

The second mechanism affecting the overall force developed by the muscle is temporal summation. The frequency at which forces are sufficiently smooth is known as fusion frequency. The point at which fusion is achieved depends upon the speed of contraction of the activated muscle fibers and, therefore, ultimately upon the level of recruitment. In most human extremity muscles, fusion occurs at less than 20 Hz. Increasing the stimulus frequency above the fusion frequency to the level of tetanus results in a further increase in force. Up to 40 or 50 percent of the maximum muscle force may be regulated by temporal summation from fusion to tetanus.

III. STANDING AND WALKING OF COMPLETE PARAPLEGICS

The prerequisite for walking is the ability to maintain a vertical posture, that is, to stand. The task of standing an individual with paralyzed legs upright is comprised of two important subtasks: 1) to provide sufficient strength and endurance of the leg extensor muscles to support the body weight against gravity, and 2) to control the body in a very unfavorable biomechanical position.

Kralj and Bajd [4] reviewed their original work where many paraplegic individuals participated in the muscle strengthening program leading to the ability to stand for short times. They reported [4] that the minimum joint torques at the knees should be in excess of 50 Nm. These joint torques can be achieved in most individuals with paraplegia; however, it is impossible to keep them at this level for periods longer than a few minutes due to nonphysiological activation (f > 20 Hz). In most cases, the joint torques, when stimulating quadriceps muscles in an individual with paraplegia at the beginning of the rehabilitation program, were only about 10 Nm. Chronic, daily stimulation of the quadriceps for periods of 30 minutes or more built up the strength to the desired level in a few weeks. However, the chronic stimulation of paralyzed quadriceps could not build the muscle fatigue resistance to the level where tetanic contraction of the muscle generated by electrical stimulation would maintain the force for more than a few minutes. There is no currently known technique that will extend this period.

The possible method to overcome this problem is to train paraplegic individuals to maintain a so-called “C-posture” (hips in front of the knee joints), thereby loading the knee joints in the direction of hyperextension [4]. At this position, the stimulation of the knee extensors can be turned off; however, the paraplegic will be able to maintain vertical posture due to gravity and ligaments holding the knee in the slight hyperextension. With the appropriate sensory system, it might be possible to recognize the point at which gravity starts flexing the knees due to the shift of the hips behind the knee joints. This information is then used to turn on the stimulation that will activate the quadriceps This has been investigated by Andrews et al. [5, 6], who reported that paraplegic individuals were standing for periods longer than 30 minutes, in some cases approaching one hour. It is assumed that the paraplegics are capable of controlling the trunk in the upright posture. In the work of Andrews, an additional brace was used to decrease the loading of the ligaments by preventing hyperextension. This approach was originally introduced by Popović et al. [7] in a so-called Hybrid Orthosis and then refined into a Hybrid Assistive System [8-10]. The combination of a light modular orthosis was not sufficiently practical, and although the results were promising, it was not accepted for further clinical and home use. Today, there are several developments of orthotic devices that could be integrated into a hybrid system, but none have been perfected to the level that would make them widely acceptable for home usage [11, 12].

The second subtask of providing control of standing was almost exclusively approached by the use of upper extremities for support over a walker or crutches. This simplifies the control because the support area is greatly increased (four points of support), allowing the paraplegic individual to balance by using his/her use arms. Several biomechanical studies provided evidence that controlling the stiffness of ankle joints [13, 14], which is doable by stimulation of ankle flexors (tibialis anterior) and extensors (soleus and gastrocnemius), can greatly contribute to the balance in the sagittal plane. In parallel, control of balance in the coronal plane [15] must include the activation of hip abductors and adductors, which is very difficult to achieve with surface stimulation. Both tasks could be reduced and eventually resolved with implantable systems, but there is currently no practical system that provides sufficient balance control.

Almost 30 years ago, Kralj et al. [16] demonstrated that a relatively simple transcutaneous FES system could assist individuals with thoracic spinal cord injury in standing and stepping. The pioneering electrical stimulation-based walking orthosis, which followed the standing research, was a four-channel system [4]. In this orthosis, bilateral stimulation of the quadriceps muscle locks both knee joints, which results in standing when combined with voluntary hip extension. Switching on the stimulation of the common peroneal nerve in parallel with switching off the quadriceps...
stimulation on the ipsilateral leg produces the withdrawal, flexion reflex that lifts the ipsilateral foot from the ground. When the leg is lifted from the ground, a paraplegic subject leans the body forward by upper body movement and arm support; hence, when the reflex is switched off, the ipsilateral leg touches the ground at a new position in front of the contralateral foot. The sequence consists of: lifting the leg, moving the body, extending the knee, and touching the ground in a replica of the swing phase of a walking cycle. The same sequence of events would then be repeated with the contralateral leg.

This functional electrical stimulation system has progressed into a commercial NP with six channels named Parastep™ [17, 18]. The two additional stimulation channels were introduced to improve posture by activating hip extensors during stance. The Parastep system (Sigmedics Inc., Northfield, IL, USA) is typically applied in combination with a plastic ankle foot orthosis (AFO), and it is the only FES ambulation system actually approved by the Food and Drug Administration (FDA), USA. The device is a six-channel stimulator, controlled by a microprocessor and powered by a battery. The stimulation unit delivers monophasic rectangular pulses at a frequency of f = 24 Hz, with pulse duration of T = 150 µs, and current control output that can be varied manually from 0 to 300 mA. A pair of electrodes needs to be positioned, based on the manufacturer recommendations, on the quadriceps in order to maintain the standing posture. A pair of electrodes is placed laterally at the knee to produce a flexion withdrawal reflex of the leg, triggering the stepping action. Despite the suggestion of Graupe and Kohn [18], the electrodes are often not used on the gluteus maximus because patients, even with high lesion levels, are able to bear the body mass on the upper limb and to adequately control trunk equilibrium. Moreover, the stimulation of the gluteus maximus was quite uncomfortable for subjects and was a source of instability during gait. The positioning of the electrodes on the gluteus maximus is also difficult when it has to be done by the patient him- or herself. However, patients are able to put the electrodes on both the quadriceps and the peroneal nerve on their own. The donning time is about ten minutes. The stimulator controls are mounted on the standard walker and consist of push-button switches, allowing for walking, standing up and sitting down.

The walking pattern generated by the systems described above is an exercise and should be termed ambulation rather than walking. The energy consumption is several times higher than the aerobic capacity. The speed of progression is only about 0.2 m/s compared with about 1 m/s, which is typical for healthy individuals. The maximum distance that can be covered is limited to only 20 to 30 meters. Muscle fatigue is considerable and limits the standing interval. If the duration of the stance phase is decreased and the level of activation of the extensor muscles is decreased, then the speed of progression and the path that can be covered could be substantially increased because the muscle fatigue would be greatly reduced.

Seven paraplegic FES-experienced users tested a newly developed eight channel stimulation system designed by the Vienna research group [19]. The goal was to discover the influence of various stimulation parameters on gait quality. An additional task was to improve usability and simplify the use of the system. Commercially available hydrogel electrodes were attached to the gluteus and quadriceps muscles and to the peroneal nerve. The later version of the stimulator included electrodes over the hip adductors. All patients were positive about the handling of the stimulation system, and the wireless remote control especially improved the simplicity of system use. Initial results demonstrated the importance of an amplitude ramp during stimulation onset (0.1-0.4 s), resulting in a smooth and more natural movement. For an adequate step length and walking speed, the timing of the quadriceps and peroneal nerve stimulation at the end of the swing phase was crucial. Experienced patients with a higher walking speed required a short swing phase and an overlap of quadriceps decay and peroneal nerve onset (0.2-0.4 s). Activation of the adductor muscles reduced hip abduction and led to a better knee trajectory during standing and better leg movement during the swing phase. Patients performed between 15 and 25 steps per minute, still with a short step length (20 cm to 30 cm). The walking distance until exhaustion or muscle fatigue occurred between 4 and 60 m. The analysis of the used stimulation sequences resulted in guidelines for a fast and effective parameter optimization procedure.

Graupe [20], one of the pioneers in the design of new and promising methods for walking of paraplegics, recently reviewed the achievements in this field. The study considered the use of the Parastep stimulation system. The review suggests that the average walking distance found in several research studies is 440 m when major muscle...
reinforcement and preconditioning of cardiovascular and respiratory systems preceded gait training. Medical, metabolic, and psychological outcomes indicate benefits of FES walking, including a 60% increase in blood flow to lower extremities. Myofiber tissues of patients with upper motor neuron paralysis compare well with those of normal tissue even many years post-injury.

The standing and walking of patients were also evaluated from the point of view of physiological changes in the body. Following an initial training period to strengthen the lower extremities, approximately three months of FES walking training with the Parastep-1 system has been shown to provide substantial benefits to various physiological systems. For example, lower extremity muscle mass is significantly increased, with up to a 50% increase in thigh muscle volume [21]. Interestingly, peripheral vasculature in those paralyzed limbs, including the cross-sectional size of and flow through the common femoral artery, increased proportionally with muscular hypertrophy [22]. Additionally, after training there was an enhanced hyperemic response to ischemia, suggesting positive adaptations in the size and function of more peripheral vascular structures. Central cardiovascular training effects benefit from the use of non-paralyzed muscles [23]. Following thirteen weeks of training, there was a significant improvement in performance on a volitional arm crank stress test. The peak power output reached, time to fatigue, and peak VO2 were significantly increased, while heart rate was significantly lower at rest and at matched levels of power output. Thus, the improved performance with the arms, which were not primarily stressed with transcutaneous FES walking, indicates that positive central cardiovascular effects occurred. Some studies reported a decrease in spasticity in participants with incomplete spinal cord injury (SCI,) whereas other studies reported no change. Bone mineral density has also been considered in FES studies, which failed to find any change in bone mineral density following three–four months of FES gait training. Several studies [4] evaluated lower limb muscle strength following FES gait training by measuring knee torque or by assessing the changes in thigh girth. The results clearly show increases in muscle strength and/or muscle girth. In one study, the knee joints were evaluated by means of MRI, and data demonstrated improvements in pre-existing pathologies. Effusions and cartilage matrix glycoproteins in synovial fluid either remained stable or improved, confirming that no deleterious effects occurred from implementing an FES gait, as well as suggesting possible benefits to joint health.

IV. NEURAL PROSTHESES FOR WALKING RESTORATION IN INCOMPLETE PARAPLEGICS AND HEMIPLEGICS

Incomplete paraplegics and hemiplegics, in contrast to complete paraplegics, are able to stand with some support and have some motor function in their legs. These differences eliminate the need for prolonged stimulation of knee extensors that ultimately lead to fatigue, and they also reduce the problem of balance control. Stein et al. [24] presented the use of a drop-foot stimulator that is efficient in hemiplegics who can walk with external support of only the paretic foot. Here we address the walking ability in individuals who do have other motor impairment that requires more than just the correction of drop-foot. Current neuroscience research provides evidence that the assistance in walking will not only be orthotic, but will also lead to long-term carry-over effects that will provide the ability to walk without an NP [25]. Clinical assessments of the regained function after treatments where incomplete paraplegics and hemiplegics walked with the assistance of robots such as Lokomat® [26] or Advanced Gait Trainer® [27] suggest that inducing normal walking retrained the central nervous system and provides much more efficient use of both nonparetic and paretic extremities. This is a definite reason to apply a practical NP based on electrical stimulation that could induce normal walking as an orthosis for therapy.

The basic idea that has been addressed for the design of a controller for walking was to record muscle activity from the prime joint movers relevant for walking in healthy individuals in order to generate stimulation profiles. The stimulation profiles, when delivered to the paralyzed muscles via an external stimulator, should generate joint movements that will lead to normal walking. This approach has been studied in detail, and various soft computing techniques have been tested successfully to generate the sensory to EMG mappings: 1) Multilayer Perceptron type of Artificial Neural Networks (ANN); 2) Radial Basis Function (RBF) type of ANN; 3) Adaptive Logic Networks; 4) Fuzzy Logic; 5) Inductive Learning (IL); 6) Adaptive-Network-Based Fuzzy Inference system and 7) combination of RBF network and IL. The original research and development of these techniques can be found in [28-36].

Figure 2: The general schema of the rule-based control derived from a minimum entropy algorithm. For details see [37, 38]
We illustrate the method of mapping by using the work of Nikolić and Popović [37, 38]. The input data were sensory signals from sensors denoted here by $P_i$, $i=1,2,\ldots,J$, and the output data were EMG activities from muscles recorded during walking and denoted as $R_j$, $j=1,2,\ldots,M$. The data sampled from the sensors were quantified to discrete levels. The schema of the multi threshold, $J$ inputs, and $M$ outputs model is presented in Fig. 2.

This technique provided the map that can be depicted as shown in Fig. 3. The top panels (IF) show the subset of input data, while the bottom panels (THEN) show the actual inductive learning determined by muscle activity superimposed over the actual recordings. Fig. 3 shows only a single threshold result.

**Figure 3:** The subset of inputs used for learning (top panels) and predicted timings of muscle activities superimposed over the EMG recordings used within inductive learning (bottom panels).

The actual result that was implemented in the microcontroller that controlled the stimulator outputs was a series of propositional logic formulas consisting of a disjunction of conjunctions. An example of the logic formula (rule within the rule base) for one muscle that was controlled by five sensors is:

$$ R = \overline{P_1} (P_2 \cdot P_3 \cdot P_4 \cdot P_5) + P_1 \overline{P_2} (P_3 \cdot P_4 \cdot P_5) + \overline{P_1} \overline{P_2} P_3 \cdot P_4 \cdot P_5 $$

The line over the letter “$P_i$” denotes that the sensory value is below the threshold, and the value “$P_i$” denotes that the sensory value is above the threshold. In the example given, the muscle is vastus lateralis, and the sensors include those for heel force, toe force, knee joint angle, hip joint angle, and the displacement of the trunk from the vertical, respectively. This conjunction of disjunctions is very convenient for real-time control. Notice that each muscle has its own threshold (Fig. 2).

V. MECHANICAL MODEL FOR POSTURAL AND WALKING CONTROL

In a previous section we presented a method for formulation of rules where the inputs were sensors and the outputs were profiles of muscle activities (EMG) recorded from healthy individuals. The cloning of EMG patterns is not necessarily the best method for activation of paralyzed muscles. The main reasons are that electrical stimulation does not activate muscles in a physiological manner, muscle properties and neural connections in individuals with disability are modified compared with healthy counterparts, and there are no artificial sensors that can be mounted on the body to replicate the natural sensory system which is essential for biological control. In addition, the individual with a disability is not aware of the actions of the external control since the feedback to higher centers of the central nervous system is greatly modified or completely eliminated due to injury.

**Figure 4:** The mechanical model used for simulation of walking. The simulation results are muscle activations profiles. The walking trajectory and muscle activation profiles form the data to be used for machine learning, that is, synthesis of rules for control.

Therefore, it is of interest to artificially generate profiles of muscle activation that, when generated by an NP, will lead to the desired trajectories, which are joint angles resembling those that are characteristic of normal walking. One suitable method to determine the necessary levels of...
muscle activations is the optimization which minimizes the tracking error, provides necessary levels of co-contractions securing needed joint stiffness, and minimizes the duration and intensity of stimulation in order to reduce the risk of muscle fatigue. This ultimately leads to the use of a customized model of extremities that are activated by externally controlled muscles. The term customized is used to express the importance of using the model, which is sufficiently complex to incorporate features of the body, yet simple enough to allow identification of all parameters. The literature is rich with models related to biomechanical analysis of standing and walking [reviewed in 39 and 40].

The model of muscle activation that we suggest has two parts: 1) the arms (potentially with walking aids), head, trunk, and contralateral leg; and 2) the ipsilateral leg. An interface force (FH) and torque (MH) acting between the aforementioned two parts directly gives the model presented in Fig. 4. The ipsilateral leg could comprise four rigid body segments connected with pin joints (toe phalange, ankle, and knee) and a ball joint (hip) with a total of five rotations that need to be externally controlled, termed degrees of freedom (DoF). The ground reaction is considered as the external force acting at the sole at the center of pressure, which is moving during the stance phase of the gait cycle. The interface to the body (FH,MH) is treated as the input for the simulation.

The model is defined by the skeleton and the actuators. We developed a practical method for estimating the inertia of body segments for the leg segments that is applicable even for humans with strong spasticity [41, 42]. Muscles, actuators within the FES system, tendons, ligaments, and connective tissues deserve special attention in modeling [43-48]. The design of practical control must consider only that level of complexity where the model parameters can be determined with sufficient precision. Muscles have dynamic characteristics inherent in their ability to develop force, including time delays, nonlinear elasticity, and nonlinear dependencies on the current muscle state (e.g., length and velocity of the muscle fibers) and past muscle states (e.g., history of the shortening/lengthening and activation of the muscle fibers).

Active muscle forces depend on three factors: neural activation, muscle length, and velocity of shortening or lengthening [44-46]. The model that we use is formulated as a function of joint angle and angular velocity, relations between velocity and angular velocity, and the passive properties of the joint (tendons, ligaments, and soft tissues). The actuation on each of the joints [i.e., joint torque (M)] can be assumed to be of the form:

\[ M = M_{\text{agonist}} - M_{\text{antagonist}} - M_{\text{passive}} \]

The three-factor model of the muscle reduced to 2D motion (e.g., sagittal plane) is given by:

\[ M_{\text{agonist}} = A(u)M_{\varphi}(\varphi)M_{\omega}(\omega) \]

where \( M_{\text{agonist}} \) is the active torque generated by the muscle contraction, \( A(u) \) comprises activation dynamics of muscle contraction that follows the control input \( u \) (amount of charge per pulse, which depends on the pulse amplitude and duration), \( M_{\varphi} \) is the dependence on the joint angle \( \varphi \) (change of the muscle length), and \( M_{\omega} \) is the dependence on the angular velocity \( \omega = \frac{dp}{dt} \) (velocity of muscle shortening). The control, \( u \), lies within the interval 0 (minimum) \( \leq u \leq 1 \) (maximum). According to the literature [44], the muscle model described can predict the muscle torque with 85-90% accuracy during simultaneous, independent, and pseudo-random variations of recruitment, angle, and angular velocity. The torque \( M_{\text{antagonist}} \) has the same form. If the muscle is not activated, the activation dynamics is \( A(u) = 0 \).

Figure 5 presents a loaded joint when both agonist and antagonist muscles are active. The torque generated by the active muscles and the torque contributed by tissues around the joint sum to produce the total torque that acts on the load and produces movement. The muscle response to electrical stimulation can be approximated by a second order, critically damped, low-pass filter with a delay [3, 46, 48-50]. Thus, the activation dynamics in the frequency domain is:

\[ A(j\omega) = \frac{\omega_p^2}{\omega^2 + 2j\xi\omega_p + \omega_p^2} e^{-j\tau_d} \]

where \( A(j\omega) \) is the Fourier transform of the muscle's contractile activity, \( U(j\omega) \) is the Fourier transform of the muscle's electrical activity, \( \omega_p \) is muscle's natural (pole) frequency (≈ 1-3 Hz), and \( \tau_d \) is the excitation-contraction (and other) delay of the muscle (≈ 20-50 ms).

The nonlinear function of the torque vs. joint angle can be approximated by a quadratic polynomial. The torque can not be negative; hence:

\[ M_{\varphi}(\varphi) = \begin{cases} a_0 + a_1\varphi + a_2\varphi^2, & \text{if } (a_0 + a_1\varphi + a_2\varphi^2) \geq 0 \\ 0, & \text{if } (a_0 + a_1\varphi + a_2\varphi^2) < 0 \end{cases} \]

Polynomial coefficients \( a_0, a_1, \) and \( a_2 \) characterize the specific properties of the muscles acting upon the joint. The nonlinear curve relating normalized torque vs. angular velocity of the joint is approximated by:

\[ M_{\omega}(\omega) = \begin{cases} K(c_{2\omega} + c_{0\omega}), & \omega = 0 \\ c_{2\omega} = \text{Const. } \land c_{2\omega} \leq M_{0} \\ M_{0} \leq c_{2\omega} \leq M_{0} \land c_{0\omega} \leq M_{0} \leq 0 \end{cases} \]

K, \( c_{2\omega}, c_{0\omega} \), and \( c_{0\omega} \) are the coefficients determining the properties of the muscle.

Finally, the passive (resistive) torque steeply limits the rotation at the end of the functional range of movement (exponential members). The passive torque is also
comprised of Coulomb type friction and elasticity to represent other soft tissues:

\[ M_f = c_{1r} e^{c_{2r}(\varphi - \varphi_0)} - c_{3r} e^{c_{4r}(\varphi - \varphi_0)} + c_{5r} + c_{6r} \omega + K_r \varphi \]

The coefficients \( c_{1r}, c_{2r}, c_{3r}, \) and \( c_{4r} \) determine the amount of ligament action which prevents the motion from extending beyond the physiological range, and \( c_{5r}, c_{6r}, \) and \( K_r \) represent the contribution of passive elasticity and friction at the level of the joint.

The model presented does not include the stimulation frequency that determines the maximum level of force. Manipulation of frequency, especially the use of doublets or triplets (two or three pulses separated by intervals of only two to five milliseconds) results in stronger forces. However, keeping the frequency as low as possible is important, since lower frequency delays the development of fatigue. The lowest stimulation frequencies are those that ensure fused contraction, and for leg muscles this is about 20 Hz.

Parameters \( c_i, a_i, \omega_p, \tau_d, \xi \) and \( K_i \) define the joint properties. These parameters are characteristic of each joint and, ultimately, of each individual. Knowing the parameters is essential for the determination of stimulation profiles \( (u(t)) \). There is no precise method to determine the parameters, but a practical method of identification with sufficient precision for the synthesis of FES controllers was developed by Stein et al. [41] and Chizeck et al. [42]. Chizeck et al. [42] presented the identification of electrically stimulated muscle model parameters in real-time, when both the pulse duration and stimulation frequency are modulated. This work considered different loading conditions, including isometric and non-isometric constant loading, as well as the presence of load transitions. We illustrate (Fig. 6) the importance of the model parameters in parallel with demonstration of the differences between the simulation results and EMG profiles (rectified, low-passed EMG recorded from muscles responsible for walking).

The desired trajectories for the simulation (joint angles) are the walking trajectories measured in a healthy individual. The data were captured in the Motion Laboratory using a camera-based motion capture system as described elsewhere [52]. The original marker positions recorded by the cameras were used to calculate joint angles and joint angular velocities. The original EMG signals were processed, and linear envelopes showing the overall profiles of muscle activities during the gait cycle were generated. The gait analysis and EMG recordings were synchronized.

The plots in Fig. 6 show the recorded EMG superimposed on the profiles of muscle activations calculated by the simulations. The simulation in Fig. 6(a) used the model parameters determined from the same subject in whom the EMG was recorded. Note that the simulation-generated muscle activations and EMG are similar. Both profiles show a certain level of coactivation of agonist and antagonistic muscles. The level of coactivation is less pronounced in the profiles determined by the simulation since the penalty function causes muscle activities to be minimal. The minimum muscle activation constraint was introduced for the purpose of eliminating long periods of stimulation and reducing the amount of muscle activation, with the expectation that this will sufficiently delay the onset of fatigue in muscles that are non-physiologically activated (a critical factor for successful application of FES).

Figure 6: Muscle activation profiles calculated in the simulations (SIM) compared to the recorded EMG. The plots show one gait stride starting with a heel strike. The simulations used (a) the model of a healthy subject and (b) the model with the reduced muscle strength (hemiplegic individual). Calculated muscle activations are signals constrained to be between 0 (muscle relaxed) and 1 (muscle fully activated). The EMG is normalized to the value recorded during maximal voluntary contractions. For the healthy muscles, the EMG and simulation-generated activations follow the same trend. However, if the muscles are weak, the profiles are very different. Simulation is described in details elsewhere [52-54].
Fig. 6(b) shows the simulation results that were obtained for the same desired trajectory with the parameters determined in a hemiplegic individual who was able to stand on his paretic leg with support over the tripod. The comparison of bodily parameters of healthy and hemiplegic individuals led to the conclusion that the muscles of a hemiplegic can produce only about 60% of the output of a healthy individual. As a result of the weaker muscles, the calculated activation levels are higher [compared to Fig. 6(a)], saturating in some periods at the maximum level of 1.

The term saturation is used to express the situation in which the muscle, even when stimulated at maximum intensity, cannot generate the required joint torque. Importantly, the simulation-generated profiles for this case are very much different, both in the levels and timing of muscle activity, from the EMG characterizing normal walking. These observations illustrate the fact that it is indeed very important to take into account the current status of the musculoskeletal system of each user when designing the control of an NP, and to use the profiles of simulation that correspond to the user’s abilities.

The simulation has another advantage in that it can show that some walking patterns are not achievable. In this case, a different (slower) walking pattern needs to be considered for walking.

VI. CONCLUSIONS

The standing and walking of incomplete paraplegics, tetraplegics, and hemiplegics is becoming feasible with currently available technology. The development of multi-pad electrodes [55, 56] is starting to allow relatively simple positioning of electrodes that allow selective activation of muscle groups of interest for a specific movement. The use of multi-pad electrodes opens the possibility of lowering the stimulation frequency to levels closer to physiological activation of sensory-motor systems in paralyzed individuals. MEMS technologies available today provide miniature, low-cost, easy to mount, robust sensors that can be integrated into a network; therefore, a replica of exteroceptors and proprioceptors relevant for detecting the state of a system is now available. The size of the sensors and the possible wireless communication with the host microcomputer allow for the mounting of the sensor system to the body. The specific sensors that we envision include accelerometers, gyrosopes, distance sensors, miniature cameras, and force and pressure sensors. The currently available microcomputing power and memory sizes are overwhelming. The electronic stimulators, as described by Broderick et al. [57], are available and support multi-sensor input and output. Finally, rule-base control that we suggest is directly applicable to the control of a multi-channel NP. The rule-base can be generated not only with the method that we described, but also with many other soft-computing techniques.

The NP that we suggest does not resolve the problem of balance, with the result that additional arm and hand support is likely required. Surface electrical stimulation is more favorable for therapy than implantable systems; however, implants have definite advantages of long term and home usage.

VII. REFERENCES


VIII. ACKNOWLEDGMENT

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