Chapter

Hybrid Neuroprosthesis for Lower Limbs

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Abstract

Assistive technologies have been proposed for the locomotion of people with spinal cord injury (SCI). One of them is the neuroprosthesis that arouses the interest of developers and health professionals bearing in mind the beneficial effects promoted in people with SCI. Thus, the first session of this chapter presents the principles of human motility and the impact that spinal cord injury causes on a person's mobility. The second session presents functional electrical stimulation as a solution for the immobility of paralyzed muscles. It explains the working principles of constituent modules and main stimulatory parameters. The third session introduces the concepts and characteristics of neural prosthesis hybridization. The last two sessions present and discuss examples of hybrid neuroprostheses. Such systems employ hybrid assistive lower limb strategies to evoke functional movements in people with SCI, associating the motor effects of active and/or passive orthoses to a functional electrical stimulation (FES) system. Examples of typical applications of FES in rehabilitation are discussed.

Keywords: spinal cord injury, locomotion, lower limb, functional electrical stimulation, neuroprosthesis, hybrid neuroprosthesis

1. Introduction

1.1 Human motricity and the impact of spinal cord injury

There are three forms of human motricity: voluntary, involuntary, and reflex. Voluntary motricity is represented by the pyramidal system. Cortical motor cells and their extensions form the corticospinal pathway. The motor system is bineuronal and extends from the cerebral cortex to the myoneural junction. The first neuron (central motor neuron) has its cell body in the cerebral cortex from where its axon goes out. Synaptic endings occur in the anterior roots of the spinal cord, where they connect with the second neuron (peripheral motor neuron). This is the so-called pyramidal path. The axons of the pyramidal path pass through the oval center, the inner capsule, and arrive in the brainstem where most of their fibers cross the midline. These axons follow along with the lateral cord of the spinal cord and end up connecting the peripheral motor neuron in the anterior root of the spinal cord.

Involuntary motricity involves the extrapyramidal system. Cell bodies stem from the various nuclei of the base and are associated with areas of the motor cortex, pre-motor, and subthalamic nuclei. There are several paths such as rubrospinal, reticulospinal, vestibulospinal, and cephalospinal paths. The extrapyramidal system and its pathways harmonize the voluntary motor system, guarantee the automatic motricity, and control the postural reflexes of spinal and vestibulocerebellar origins.

Reflex motricity depends on the pyramidal and extrapyramidal systems and represents only a few spinal reflexes.

Peripheral motor neurons are part of the peripheral nervous system (PNS) and are organized into motor units. Nerve fibers protract from the anterior roots of the spinal cord to muscle fibers and organs of muscular proprioception called muscle spindles. The spindles send sensorial signals to the spinal cord, informing the central nervous system (CNS) about the level of muscle contraction.

The CNS triggers nerve impulses that travel along the motor neuron toward the muscle fiber so that contraction can happen. They can excite the neuronal membrane to reach depolarization voltage levels above a triggering threshold that generates a particular wave known as action potential (AP). The AP consistently propagates along the axon toward the synaptic ending. At the synaptic cleft, vesicles deliver the neurotransmitter acetylcholine that connects to cholinergic receptors and depolarizes the myoneural junction. Eventually, the depolarization can generate a new AP that propagates along the sarcolemma leading to muscle contraction. In this process, fibers can stretch, shorten, or remain isometric although producing force. This force, in turn, is transmitted to the tendinous and bony structures [1]. Movement occurs that way. All these forms of motricity work well for a healthy subject. However, everything changes when spinal cord injury (SCI) occurs. **Figure 1** illustrates the central nervous system and its afferent and efferent pathways.

People with SCI have ruptured or impaired communication between CNS and organs that control motor and/or sensory functions. Nerve impulses are electrochemical processes and their transmissions occur in two directions. From CNS to PNS, they trigger muscle contraction processes. From PNS to CNS, they send sensorial processes that capture the stimuli from the surrounding environment as shown in **Figure 1** [1].

SCI harms the neurological responses according to the compromised site, that is, the level of the affected pathways. Thus, a complete SCI, which interrupts all nerve pathways, comprises in the acute-phase (also known as a medullary shock) flaccid paralysis with deep areflexia and muscle hypotonia to, consequently, give rise to spastic paralysis with hypertonia, deep hyperreflexia, and a sign of pyramidal release. Sensitive changes (hypoesthesia or anesthesia) occur for all forms of sensitivity below the level of injury. Partial or incomplete lesions are a consequence of the affected pathways. For instance, if there is a spinal hemisection, then there will be homolateral and contralateral signs and symptoms. On the same side of the injury, paresis or paralysis of the first neuron, abolition of deep sensations and alteration of gait occur. On the contralateral side, thermal and painful anesthesia is observed with no change in strength.

In the last decades, neuroscientists and rehabilitation engineers have been seeking alternatives to recover the mobility of people with SCI. The goal was to provide them with a better quality of life and functional independence. In 2012, the World Health Organization (WHO) asserted that about 0.5% of the population in developing countries needs prosthetic and orthotic devices and that 1.0% of that population needs wheelchairs [3]. Between 250,000 and 500,000 people suffer from SCI in the world, of which the majority are men and women between the ages of 15 and 25 and the elderly over 60 [4]. For the elderly, according to the United Nations Population Report [5], there will be an abrupt increase of people over 60 in 2050, reaching around 1 billion people among healthy individuals, people with SCI, heart

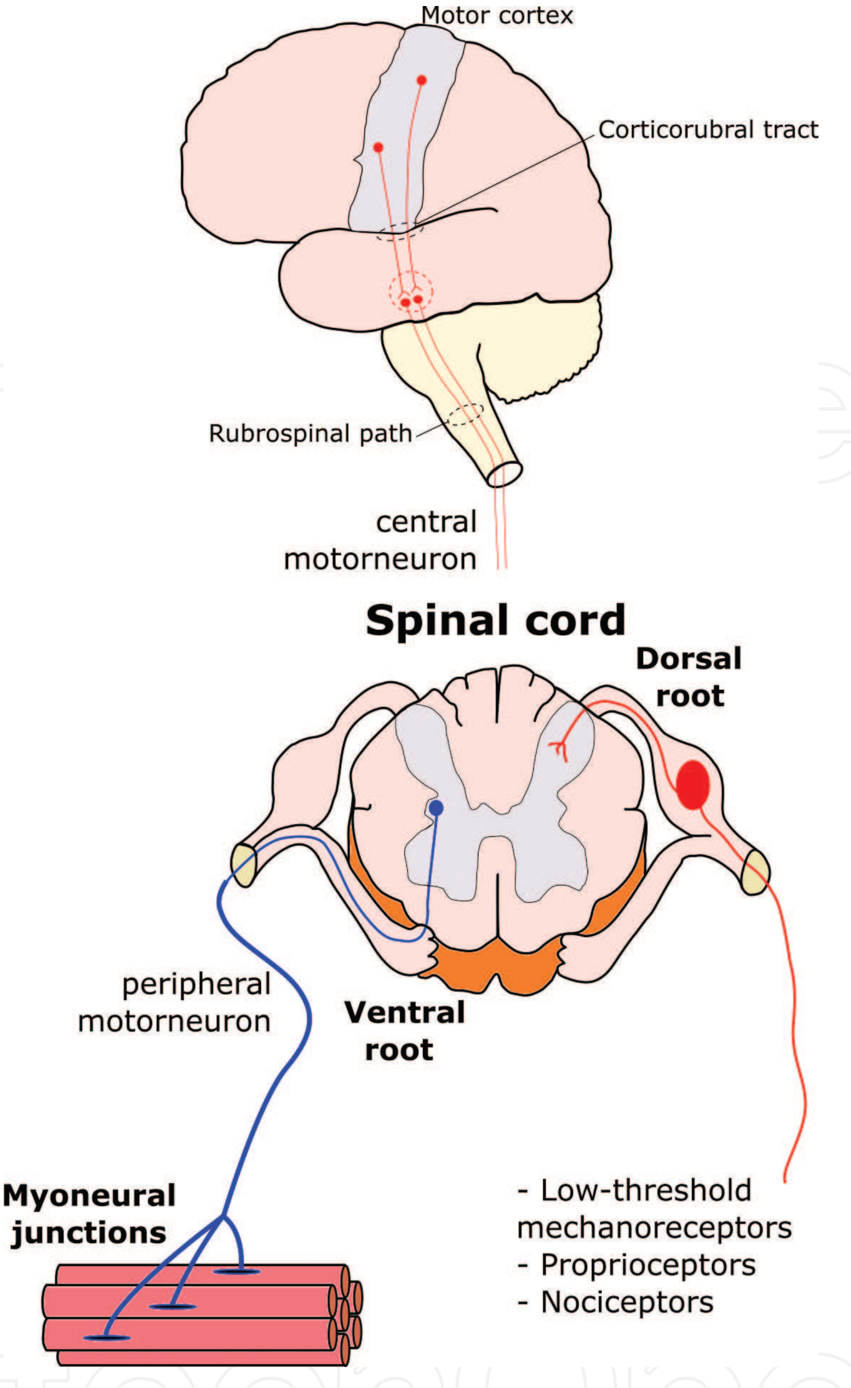

Figure 1.
Central nervous system and its afferent and efferent pathways [2].

disease, and other factors that may compromise locomotion. More than 1.2 million people in the United States have SCI that disables mobility, generating an estimated annual cost of $ 40.5 billion [6]. A Canadian study with 1716 individuals with SCI indicated a median lifetime expenditure of $ 336,000 per person, up to $ 479,000 if bedsores occur early in the hospital [7]. As costs are high, new techniques and devices are researched and evaluated to reduce these costs and/or minimize the impacts caused by immobility.

Devices such as wheelchairs, crutches, and walkers have been in use to aid the locomotion and rehabilitation of the elderly or people with SCI. However, these solutions are not fully effective and users expend great energy. It also requires the assistance of physiotherapists, caregivers, or family members [8, 9]. Therefore, alternatives that reduce the physical demands of users, therapists, and caregivers have been sought. Orthoses, neuroprostheses, and exoskeletons emerged as technologies that assist the individual's general health and therapeutic rehabilitation. These devices are capable of producing more intense training, quantitative feedback, and better functional results [9, 10].

In the case of people with SCI, the condition may be irreversible, resulting in some type of paresis: partial hemiplegia, paraplegia, or quadriplegia. Orthoses and/or functional electrical stimulation (FES) allow those people to perform ambulation (active orthoses) and/or provide them trunk stability (passive orthoses). Such technologies facilitate their social reintegration, increase self-esteem, and improve the general quality of life. This can be achieved since these solutions induce a decrease in other affections caused by limb paralysis, such as muscular atrophy, which reduces muscle strength and can prevent functional movements from happening [11–13]. Among the other sequelae that may arise as a consequence of SCI, one can mention: respiratory difficulties, intestinal and/or urinary incontinence, loss of sexual functions, deficiency of lymphatic and vascular system, muscle atrophy (which can result in spasticity), pressure ulcers, thrombosis, and bone demineralization [14].

2. Neuroprosthesis

Since the 1970s, FES has proved effective in restoring functional movements for both increasing strength and aid in impaired locomotion. It is being applied as a means of excitation of motor neurons. In SCI subjects, it bypasses the injury and bridges the CNS and muscles. Therefore, FES can cause real movement with an artificially evoked contraction as the injured spinal cord has this compromised communication [1, 15].

The bypass is performed by electronic devices known as functional electrical stimulators. The stimulators apply an electric current to the neuromuscular tissue through either implanted or non-implanted electrodes. In general, they consist of a pulse generator circuit or modulator (low voltage—up to 12 V), an amplifier (which increases voltage—up to 250 V) that provides the stimulatory pulses to be applied, and an isolated power supply. **Figure 2** illustrates a generic stimulation system.

2.1 Modulation (pulse generation)

Pulse generators determine the essential features of the stimulatory pulse, such as frequency, duration, and waveform. Many technologies have been used over time as innovations emerged in electronic devices. The pulse generator of a

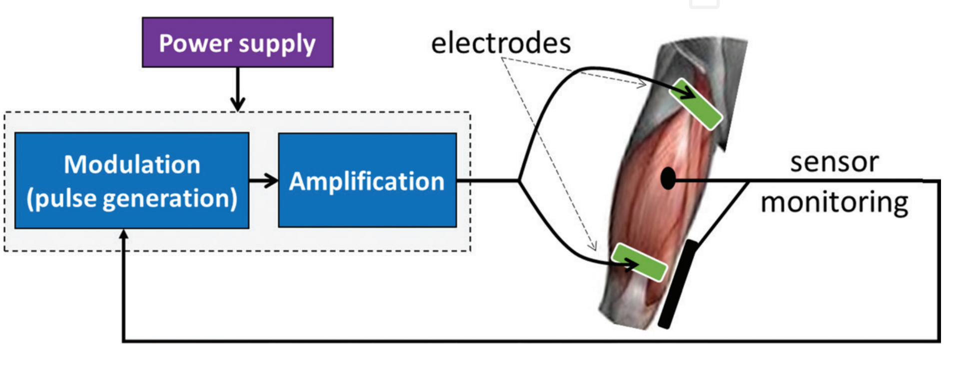

Figure 2.
Block diagram of a single-channel functional electrical stimulator. The essential steps for FES-elicited contraction involve the generation, amplification, and application of these pulses to the neuromuscular tissue. Sensors can help to correct pulse parameters as well as to determine the triggering time. All modules require power supply, whereas sensors may be active or passive.

functional electrical stimulator used two multivibrators, one configured as monostable and the other as astable, based on the LM555 integrated circuit (IC) [16]. Subsequent attempts, in turn, used development boards and more advanced digital components. One equipment used a Texas Instruments TMS320C32 digital signal processor (DSP) to control and generate arbitrary waveforms and to manage the stimulator [17]. The Arduino platform was employed and the ATMEL® microcontroller adjusted the stimulatory pulse frequency by means of a digital potentiometer [18]. Alternatively, others implemented a pulse generation module in a Motorola 68HC11F1™ microcontroller [19]. Generally, the stimulation parameters of a neuroprosthesis are generated and adapted by control algorithms that receive data from walking sensors and allow adaptation to a user's walking speed.

Use of ICs specifically designed for functional rehabilitation with the very large scale of integration (VLSI) has been growing rapidly. A common approach for pulse generation is to use a simple and inexpensive microcontroller for each stimulus output channel. A PIC16F84 microcontroller was used exclusively as a pulse generator [20]. The pulse formatting was performed via a virtual graphical interface. More recently, a solution used a virtual module for pulse generation. It was implemented in a virtual control instrument connected to a data acquisition and control board [21].

In order to achieve the most efficient biomechanical task and meet the appropriate physiological conditions, stimulatory pulses are modulated. This process changes the original waveform depending on the shape or value of a second signal. The resulting signal is indeed the mathematical composition of the previous ones. Pulse amplitude, duration, and frequency modulation facilitate access to more or less deep neuromotor units. Pulse amplitude modulation (PAM) and pulse width modulation (PWM) are the most frequent techniques [22]. PWM allows the energy delivered to the biological load to be controlled.

Stimulatory waveforms can be rectangular or exponential in shape, have a pulse width ranging from 0.01 to 1 ms, and burst frequency from 20 up to 100 Hz. Possible pain sensation is related to the transfer of electric charge from the stimulator to the user. Pulse durations between 100 and 300 μs cause little heat dissipation in the application area, leading to lower levels of pain [23]. Values shorter than 10 μs reduce the risk of intramuscular damage and those between 64 and 1230 μs can cause small skin irritation. The minimum burst contraction frequency is 10 Hz; however, apparent muscle tremor can be observed. Tremor is imperceptible with 30-Hz burst frequency. Muscle relaxation occurs from 300 to 700 Hz. There is less skin irritation with 2-kHz pulse frequency modulated at 50 Hz and lasting 10 ms [23].

For a muscle contraction to occur, a single pulse frequency can be selected between 20 and 400 Hz. However, the frequency considered optimal for force production ranges from 2.5 to 5 kHz, with burst frequency from 150 to 500 Hz, and pulse width from 10 to 30% of the duty cycle. Despite this, studies indicate that pain levels can get considerably high. To minimize the painful perception, frequencies between 9 and 10 kHz are more adequate, although they do not provide the same muscle strength [23, 24].

One of the widely used wave patterns is the so-called Russian current. Pulse frequency is 2.5 kHz, with 50-Hz bursts, 10-ms per train duration, and 10-ms intervals [24].

2.2 Amplification stage

The second most important stage when building an electrical stimulation device is amplification. This module consists of an arrangement of electronic components for commuting voltage or current signals. These circuit components shape signals accordingly to stimulate the neuromuscular tissue. A matter of the highest concern is

Type of motor unit	Contraction velocity	Fatigue resistance	Force
Slow	Slow	High	Low
Fatigue resistant	Fast	Intermediary	Intermediary
Fast fatigable	Fast	Low	High

Table 1.
Motor units and their features.

user safety. Amplification usually involves safety precautions since it is the last stage in the signal chain to be in contact with the user's skin and tissues through leads and electrodes. There are two modes to apply current to neuromuscular tissue: constant voltage or constant current. The difference is that current intensity depends on biological and interface impedances in the former but not in the latter. Determining this feature implies choosing a particular topology for the output circuit [25].

Current can flow in and out of the tissue depending on the signal reference. The allowed reference signals demand appropriate output stages. Depending on the reference, the current waveform can be monophasic or biphasic. Monophasic outputs allow only unidirectional current flow. This current waveform creates charge imbalances in the tissue. Conversely, biphasic outputs allow applying to and taking electrical charges out of the tissue. This bidirectional current flow prevents charge accumulation. This phenomenon is harmful to the subject for it can cause chemical burns mainly around the application site [26].

FES equipment can have one or more stimulatory output stages, usually called channels, each one responsible for the stimulation of a different muscle. For gait applications, usually, multichannel FES equipment is necessary to evoke more natural movements.

The motor neuron must be intact for FES application, otherwise there will be no muscle contraction. Thus, SCI type and level determine the use or avoidance of this technology.

Another issue to consider is the muscle to be stimulated. Each muscle is formed by a specific fiber type. When excited, the motor units contract at distinct speeds and resistance to muscle fatigue, delimiting the generation of force [1]. These features can be observed in **Table 1**.

3. Hybrid neuroprosthesis

Focus on technologies that bring users functional independence is increasing and, therefore, FES devices and active orthoses are thriving. In the previous session, the neuroprostheses were described in terms of their basic principle of operation and main characteristics. Like conventional prostheses and orthoses, neuroprostheses have advantages and disadvantages compared to other techniques when it comes to generating movement. Choosing a particular neuroprosthesis may depend on the severity of the user's condition, the type of task and the performance to be obtained, the cost of acquisition and maintenance as well as durability and energy efficiency.

Employed individually, a conventional orthosis has a large physical demand on the upper limbs and its energy cost is quite high [27]. As the upper limbs have a limited range for ambulation, gait patterns obtained with this technique are inappropriate. Over time, users stop using such orthoses to perform functional tasks and end up using them only for therapeutic purposes [28].

Initially, neuromuscular electrical stimulation served to compensate for muscle atrophy or counterbalance the effects of spasticity. The first efforts in the application of electrical stimulation with functional purposes began with the attempt to maintain a person with SCI in the orthostatic position, through the production of isometric tetanic muscle contractions, as originally proposed by Bajd et al. [29]. The attempt to ambulate using neuroprostheses followed and now it is possible to find many applications using FES to perform joint movements [30]. The main advantage of FES over conventional orthoses is that the muscle itself works as the supporting structure and the motor that propels the intended movements. The disadvantage, however, is that during FES sessions, rapid installation of muscle fatigue occurs. The depletion of metabolic resources becomes an obstacle to ambulation for long distances and it hampers the fine control of joint angle trajectories [31]. The control strategies of electrical stimulation combined with invasive electrodes have been proposed to minimize the well-known disadvantages of using FES.

Robotic systems have been receiving increasing attention from both health science and engineering researchers. This is due to the various possibilities of use and locomotor training. Some examples are the LOKOMAT® system that assists people with SCI to perform walking, through mimicking human gait on a treadmill, and the active orthoses that not only mimic gait but also give such people the freedom to move around the environment and sit/stand, walk, and climb/descend stairs [32]. These orthoses are wearable devices. They have (motor, hydraulic, or pneumatic) actuators parallel to the hip, knee, and ankle joints. They can only mimic the gait with the aid of mechanical devices that limit or expand the joint degrees of freedom. This makes the gait robotic in appearance and the actuators consume a lot of power, thus considerably reducing the system's autonomy, limiting the user's independence and its application for the recovery of compromised movements [33, 34].

Although a technological solution that demanded a high degree of creativity in its development, the active orthosis produces a gait process in which the patient is only a passive element. There are no important direct physiological benefits to the paralyzed muscles during their use, but passive mobility. In contrast, FES has many physiological benefits such as the improvement of muscle tone and blood circulation in paralyzed limbs, the prevention of pressure ulcers, respiratory and/or urinary problems, and the reduction of spasticity among others [9, 12, 14]. However, the intense, long, and non-selective contraction causes the muscles to be unable to maintain the force for long periods. That is because muscle fatigue is installing and hampering the maintenance of the movement stability, denoting, therefore, limitation in the control over the movement as well as the time of use [12].

In this context, hybrid neuroprostheses (HNPs) are interesting because they combine FES with other techniques to perform functional movements. Particularly, systems that combine FES and orthoses have been introduced in the literature. This hybrid approach opens the way for the elaboration of strategies that focus on the advantages of each individual technique. Quite often in this combination set, the power applied for the occurrence of movement is provided by the electrical stimulator and the structure of the orthosis serves to stabilize the movement.

4. Hybridization of neuroprosthesis for the lower limbs

This session presents some hybrid applications of neuroprosthesis. It indicates joint control technologies, main control elements, and control strategy.

The first studies identified consisted of simulations of hybrid control systems. The device involved an active orthosis containing actuators in the hip, knee and ankle joints, angle and angular velocity sensors, and a conventional FES system. The whole system was controlled in a closed loop, using both biomechanical and dynamic equations to perform gait movements [35]. According to the results, simulations allow the reproduction of gait although studies with volunteers with real system and setting would still be required.

Researchers developed and evaluated an HNP based on a variable-impedance knee mechanism [36]. It regulates knee flexion to overcome the challenges of controlling eccentric contractions. This mechanism consists of a four-bar linkage with a magnetorheological damper, hip and knee hydraulic actuators. The solution reduced the amount of stimulation required for walking and could restore biologically correct knee motion. It locks the knee joint motion and, during the stance phase of gait, it supports the body against collapse. The association with an FES unit by means of a finite state machine (FSM) controller allowed the body forward. Heel contact detectors and joint angle sensors controlled the knee motion during stance and swing phases. Tests compared the HNP and only FES application. HNP decreased the load response by up to 40% in knee extensors.

Pressurized hydraulic fluid from an accumulator was also tried to supply energy to an HNP instead of electric motors [37]. The hydraulic drive provided adjustable assistive torque to an exoskeletal hip joint beyond the possibilities of FES systems. The volume of valves was controlled in both directions, which allowed the estimation of hip flexion angle. Springs and a clutch system provided the knee-locking capability.

The same research group built a muscle-driven controllable exoskeleton to restore walking, sitting, and standing to people with SCI [38]. They combined the mechanics used in [36] and adaptations to the hip control described in [37]. An external controller that reads embedded sensor signals and determines the appropriate adjustments to a finite state machine drives an implanted FES unit. The state machine commands and synchronizes both the exoskeleton and the stimulator. Users choose the desired functions by a wireless switch controller. Although implanted devices demand surgical intervention, they were able to restore the stepping function to three subjects. This HNP supports stair descent patterns, overcoming a task that is hardly achieved with only FES.

An active orthosis had knee and ankle joint actuators and an FES system [9]. A central controller adjusted FES parameters using two closed loops and six stimulation channels. Gait consists of two phases: balance and support. It uses FES during the balance phase and the orthosis is active during the support phase, which is the moment of gait that demands more energy expenditure from the patient. The solution presents the clear strategy of shortening the time interval between FES activations. This way, there will be more time for muscle recovery and, consequently, fewer episodes of fatigue over time. With the activation of the active orthosis, the system does not stay connected all the time and, thus, the energy demand of the system decreases.

Figure 3 shows a diagram that represents a hybrid neuroprosthesis. It consists of an electric stimulator and an active orthosis with actuators in its joints, feedback sensors for gait, and a real-time control system. This control system identifies the gait phases, evaluates the current position of the lower limbs, makes decisions to activate the actuators, or triggers the FES unit.

The controller is the main component. Gait takes place safely and naturally depending on how robust the interaction with mechanical parts is and how fast it reacts after receiving feedback information. Thus, increasing the feedback information volume may be an impacting differential between existing orthoses. More degrees of freedom per joint allow a more natural gait and this is an important factor. However, in the case of people with SCI, using only one degree of freedom per joint brings greater stability and safety during movement, thus avoiding joint injuries [33, 34].

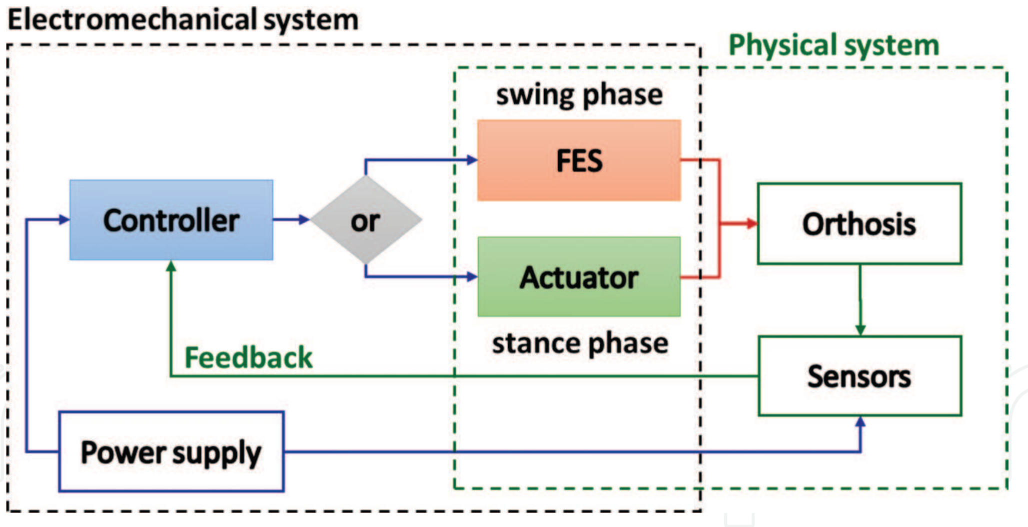

Figure 3.
Basic diagram of a hybrid neuroprosthesis system. It demonstrates the integration of the control system with the FES unit, actuators, and feedback sensors.

A semi-active hybrid orthosis [39] employed electric motors to actuate on hip movement whereas FES actuated on knee and ankle movements. Wrap spring clutches combine high torque capacity and millisecond response time. Their use prevented the knee flexion when they were locked while still holding the torque in one direction (knee extension). This approach eliminated the need for FES application during standing and to the stance leg during a step, saving muscle metabolic energy. Researchers intend to incorporate an FES channel in future HNP versions to stimulate the peroneal nerve and provide redundant actuation at the hip joint.

Another approach combined a commercial FES unit (RehaStim 2, Hasomed, Germany) and a knee exoskeleton for controlling knee joint swinging movements [40]. The main contribution is that the interactive forces of this solution are measurable and it helps in better cooperative control. The continuous decrease in muscle force performance is an indication of muscle fatigue installation. Therefore, the torque required to perform a task is distributed between both actuators (muscles under FES and electrical motor). Two interactive force sensors that measure the mutual force between exoskeleton and shank accomplish force readings. Each interactive force sensor provides equivalent summed value from six previously calibrated force-sensing resistors.

A cycling-induced HNP implemented a repetitive learning controller that allows uncertain, non-linear cycle-rider systems to track the desired cadence [41]. The controller feedback depended on the crank angle. FES applied to lower limb muscle groups drives the movements, whereas the electric motors come into play when stimulated muscles yield low torque values. The system tried to track the cadence of five able-bodied subjects and three subjects with neurological conditions. The results presented low root mean square errors.

Some devices are still in enhancement. An HNP consisted of orthotic components (reciprocating gait orthosis and rigid ankle-foot orthosis), powered backdrivable knee joints (obtained with custom harmonic drive transmissions and brushless DC motors), and was intended for use with FES [42]. One of the main contributions of this work was to be able to execute movements with a lightweight device, up to 10 kg.

Another attempt was the development of a 17.05-kg HNP prototype for short-range walking [43]. This device consists of a quadriceps single-channel FES unit and a passive energy-storing exoskeleton. Gas spring stores energy during knee extension and it delivers to knee and hip joint control when completing the gait cycle. Its walking speed could achieve 0.27 m/s.

The Research Group of the Rehabilitation Engineering Laboratory of the Pontifical Catholic University of Paraná (PUCPR) has also developed an active lower limb orthosis with electric actuators with angular feedback sensors in the hip and knee joints to perform gait [34, 44, 45]. Normally, an orthotic system has actuators in their joints, which are transverse to the joints of the individual who is "wearing" it, which facilitates the design of the system. Thus, the mechanical hip joint has a 26° flexion and a 13° extension limited by mechanical stops, as shown in **Figure 4** (1) [34]. Knee flexion and extension are in the range from 0° to 90°. The controller output will obey these limits as well as the mechanical stops, as shown in **Figure 4** (2).

Engines and transmissions will be coupled directly to the hip and knee joints, as shown in **Figure 4** (3 and 4). The gear ratio in the planetary-type transmission system is 4.75:1; so, the torque at the joint output will be 4.7 times larger than the engine torque [34].

For the control, it is necessary that there is an angular sensor in each joint, each one coupled to the axis of the motor by means of pulleys and synchronizing belt. In this way, it is possible to make the angle reading directly and in real time. Its structure with a partial weight support system can be seen in **Figure 5** [34].

The controller of the active orthosis synchronizes both knee and hip joints. A state machine simulates the gait phases and determines when each joint will engage, as shown in the diagram of **Figure 6**.

The control of each joint has predefined set points that correspond to the maximum and minimum angles of a healthy gait. It receives voltage readings of the angle sensor, calculates the actuating error, adjusts the PWM mathematically, and sends it to the motor drive. The error is determined as the difference between input angular value and angle sensor reading on the motor axis. The readings of angular sensors vary according to motor rotation. This may increase or decrease the error.

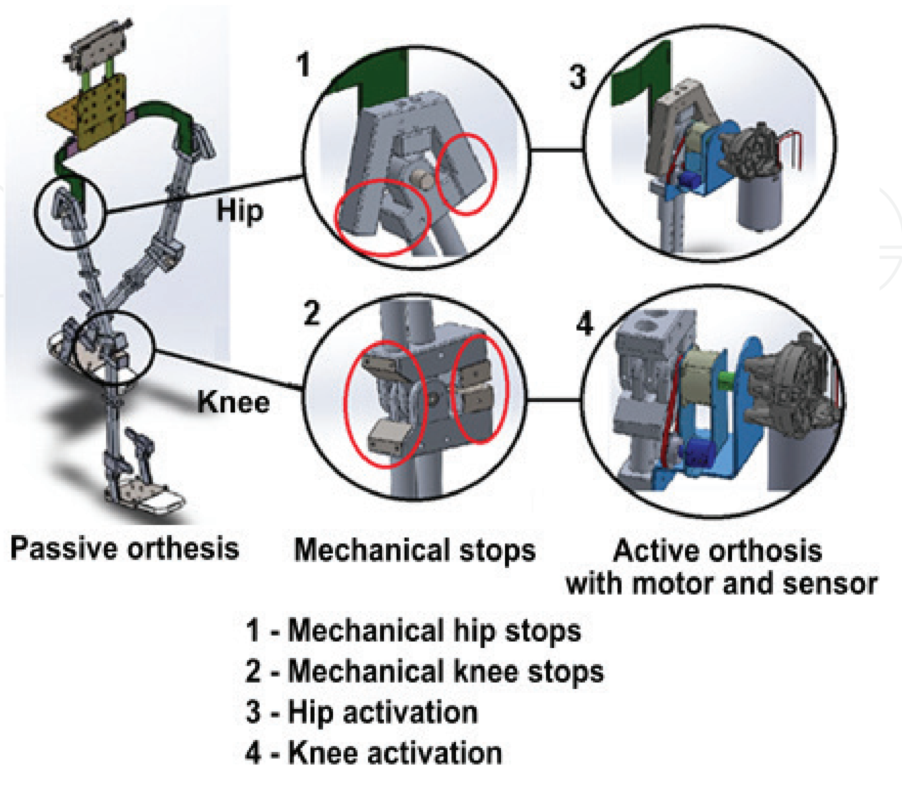

Figure 4.
Mechanical structure of the passive orthosis and activation sequence of actuators in the hip and knee joints.

Figure 5.
Mechanical structure of the active orthosis with a volunteer on the partial weight support system.

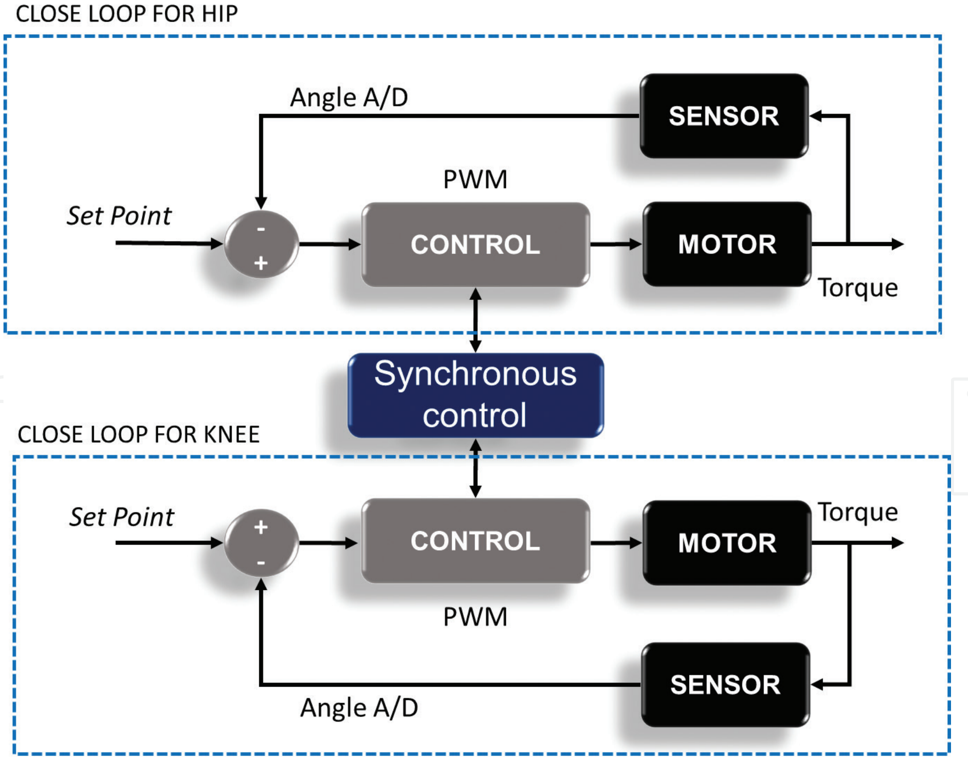

Figure 6.
Control block diagram of the active orthosis. Two loops operate synchronously to generate appropriate torque to hip and knee joints.

The closer the angle sensor reading is to the determined set point, the smaller the error. Consequently, because the motor is proportional to the driving error, the lower is the motor speed [34].

The results of active orthosis control tests, without volunteers, were normalized using MATLAB™ software (**Figure 7**) in comparison to the gait of a healthy individual (**Figures 8** and **9**) [34].

Figure 8 allows the observation of important factors. The maximum knee flexion angle achieves 50°. There is a difference presented as an initial flexion between 0 and 40% of the gait cycle, which appears in the dashed graph of the healthy individual with the amplitude of approximately 20°. The divergence is due to the reference point difference at which the angle was obtained, a bending that occurs in relation to the global system. Such lag exists because of the empirical way in which the control data are normalized. It is also because the hip could only flex and is limited to 26°, whereas knees could flex and extend. Therefore, there is a difference between the gait cadence of the developed system and the healthy gait. The former is faster than the latter. Nevertheless, the control will apply a 4.7:1 reduction in this speed, slowing down 4 times, making the orthosis cadence slower than a healthy one.

Another observation is that the healthy knee's initial angle is zero and this is due to the calibration of the system. If signals were stapled to have the same reference, the maximum bending angle would be less than 50°. This is because the control set points of the orthosis for people with SCI were adjusted to be lower than that of a healthy individual. Therefore, the control program should not allow angles to exceed the biomechanical limits equal to those of a healthy one.

Considering hip angle analysis, there were also angular differences due to the mechanical limits of the bracing. Hip flexion is limited to 26° and extension limited to 13°. In **Figure 9**, the negative half-cycle of the amplitude of the signal represents the hip extension.

Once again, the programmed angles were not limited to the maximum normal hip values, which worked as activation thresholds to the electronic safety circuit. This alternative prevented movements that exceed the biomechanical limits of the user using the orthosis. The control operation depends on the percent of gait. Consequently, the controlled movement duration is shorter than the gait of a healthy individual since the angle of movement is smaller. The solid line also denotes higher speed and sharp transition with respect to the healthy individual. However, once again, there is a 4.7:1 reduction in the speed, smoothing the gait cadence.

Studies with this system are still ongoing. One of the efforts is the hybridization of the orthosis with FES, using a differentiated control, so that there is a possible reduction in the energetic expenditure and muscular fatigue. The embedded strategy difference is to keep the hip joint under control throughout the gait and apply FES to knee extensor muscles only during gait swing phase. **Figure 10** presents the proposed system.

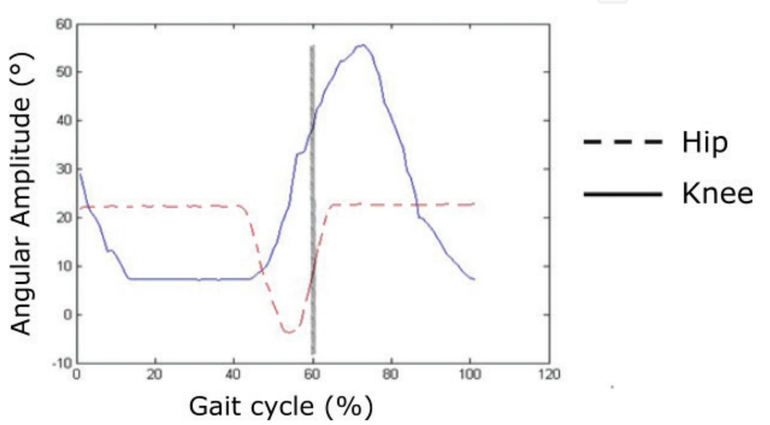

Figure 7.
Angular variation of both hip and knee joints (interval of one gait cycle).

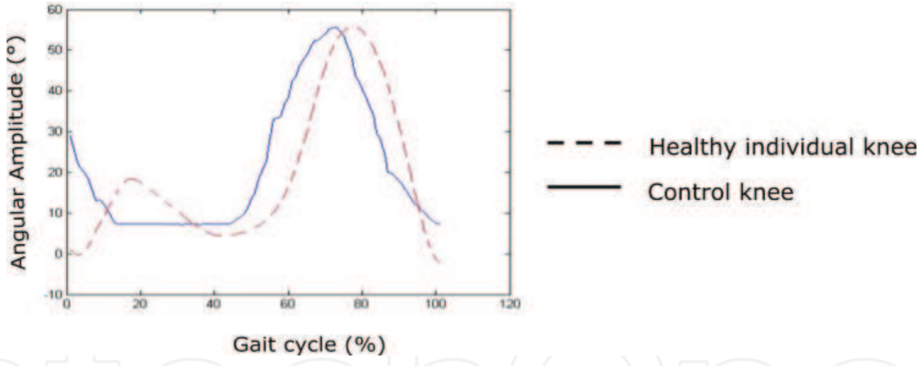

Figure 8.
Comparison of knee angular variation between orthosis control and healthy individual (interval of one gait cycle).

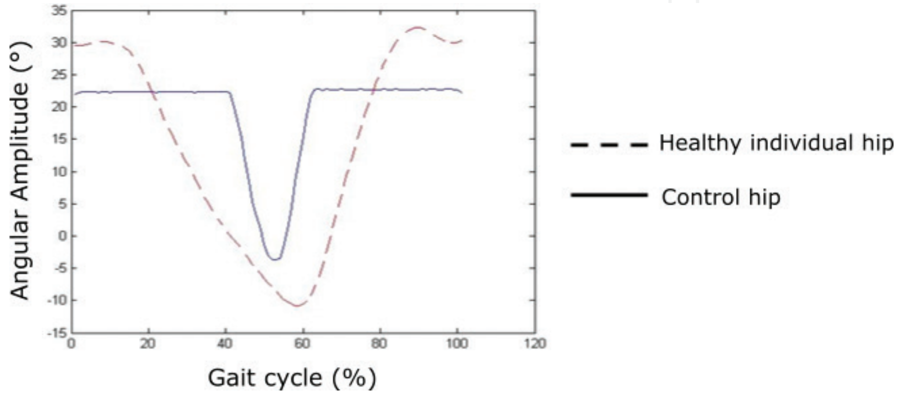

Figure 9.
Comparison of knee angular variation between orthosis control and healthy individual (interval of one gait cycle).

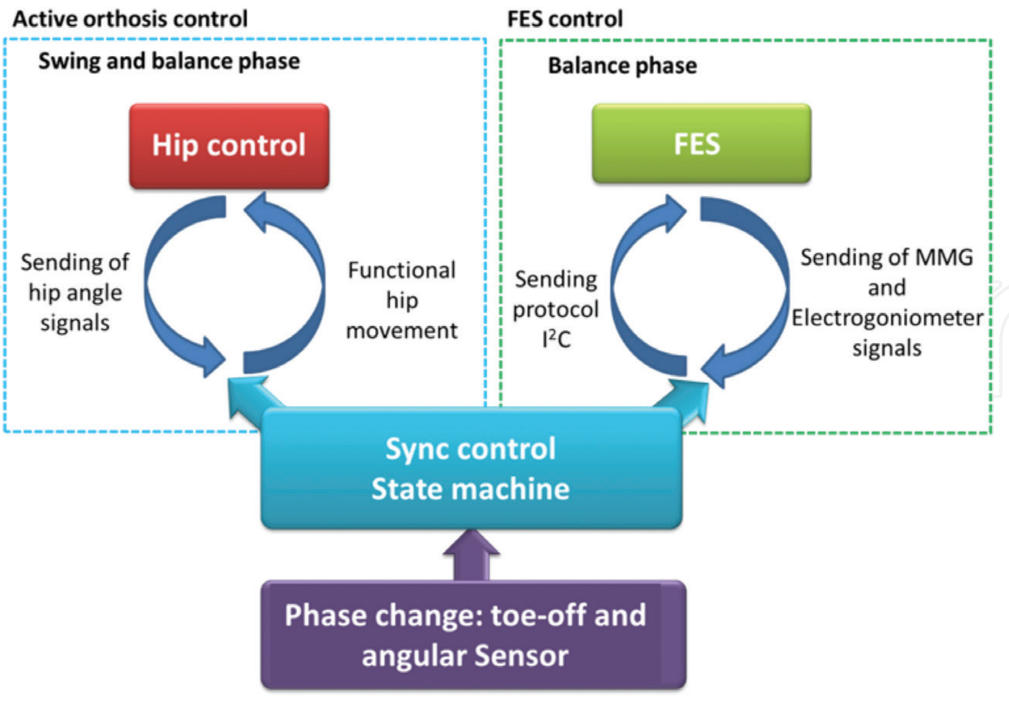

Figure 10.
Controller scheme with mechanomyography (MMG) and joint angle feedback at each stage of the gait.

Using the orthosis during both gait phases lessens the weight of the whole system on the user. In existing systems, when FES activates the muscles, the user has to raise both the mechanical apparatus and the limb. This weight-bearing requirement hastens the installation of muscle fatigue [9, 14] although it is smaller in relation to other systems that use only FES, or FES with passive orthosis [33].

This HNP has a fatigue detection system that processes mechanomyography signals [15]. The goal is to identify signal pattern changes and, in case of fatigue installation conditions, the HNP will activate the joint motors, guaranteeing user's safety.

Hybrid systems for locomotion using FES and active orthoses are relatively new devices. Despise this, there are a few projects addressing this aspect, which are enough to visualize and glimpse the benefits and perspectives of these technologies.

5. Conclusions

This chapter presented the principles of the neural prosthesis and discussed the reasons for the hybridization of these systems. The methods used to stimulate the muscles and develop a neuroprosthesis remain valid and the stimulatory parameters are the same. Researchers, however, use new embedded systems technologies and graphical interfaces to program and configure the internal parameters of FES equipment, also using hybridization of orthosis and neuroprosthesis to combine the advantages of individual techniques to counterbalance their individual disadvantages. The joint use of FES and orthosis attempts to reduce user's energy expenditure, postpone muscle fatigue installation, increase posture and movement stability, and reduce the system energy costs. Thus, hybrid neural prostheses increase system efficiency and prolong the time of use, consequently, achieving health benefits.

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