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An EMG-Controlled SMA Device for the Rehabilitation of the Ankle Joint in Post-Acute Stroke

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The capacity of flexing one's ankle is an indispensible segment of gait re-learning, as imbalance, wrong compensatory use of other joints and risk of falling may depend on the so-called drop-foot. The rehabilitation of ankle dorsiflexion may be achieved through active exercising of the relevant musculature (especially *tibialis anterior*, TA). This can be troublesome for patients affected by weakness and flaccid paresis. Thus, as needs evolve during patient's improvements, a therapeutic device should be able to guide and sustain gradual recovery by providing commensurate aid. This includes exploiting even initial attempts at voluntary motion and turns those into effective workout. An active orthosis powered by two rotary actuators containing NiTi wire was designed to obtain ankle dorsiflexion. A computer routine that analyzes the electromyographic (sEMG) signal from TA muscle is used to control the orthosis and trigger its activation. The software also provides instructions and feed-back for the patient. Tests on the orthosis proved that it can produce strokes up to 36° against resisting torques exceeding 180 Ncm. Three healthy subjects were able to control the orthosis by modulating their TA sEMG activity. The movement produced in the preliminary tests is interesting for lower limb rehabilitation, and will be further improved by optimizing body-orthosis interface. It is hoped that this device will enhance early rehabilitation and recovery of ankle mobility in stroke patients.

Keywords electromyography, orthosis, rotary actuator, shape memory

1. Introduction

The role of active exercise is fully recognized in the re-acquisition of voluntary movement after a neurological insult, and it is common clinical practice to start workout sessions as soon as the patients' conditions allow that (Ref 1, 2). It is still a matter of debate, whether segmental exercise or functional therapy is more effective in the rehabilitation of motion. Of course, occupational exercise is of paramount importance in later phases of recovery, but a correct approach dealing with acute and sub-acute impairment is likely to have profound effects on the ultimate outcomes of treatment. The control of single anatomical segments is a prerequisite of precise motion skill. Even approximate and non-optimal motion strategies can however lead to functionally sound results, so that inadequacy in the use of certain muscle groups or joints does not prevent actions to be carried out, from a practical standpoint, quite successfully. This way of looking at rehabilitation relies on the plasticity of the human neural system and its capability to adapt to disability. Not all compensatory mechanisms however come free, and the price can be paid in terms of extra energy consumption, stabilization of wrong postures, modifications in the motion patterns of adjacent segments, etc. So, it is safer to aim, especially in the first stages of the rehabilitation process, to a complete rescue of segmental ability, at least until that is definitely shown impossible for a particular subject. Furthermore, focussing (at least initially) on re-acquiring the control of a single joint or degree of freedom may lead to commence active exercise earlier: rehabilitating complex motor patterns may in fact prove challenging in the acute period after the onset of paresis.

This article supports the idea that rehabilitation of gait can start, in selected patients, as early as the acute phase following a neurological insult that caused lower limb paralysis (Ref 3, 4). Moreover, it proposes that by the use of a suitable exercising robot the process of recovery can be continuously sustained from passive mobilization over to voluntary motion. The implementation of this approach is obtained for an ankle joint application. This choice was made considering the importance of the control of this joint in walking proficiency, and limiting our study to the sole flexion-extension degree of freedom.

It will be shown that by evaluating the amount of tibialis anterior muscle activation with surface electromyography (sEMG), the robot can be controlled to provide aided dorsiflexion on demand, that is to say it can administer: passive exercise of the periarticular tissues and muscles; somatosensory (haptic) stimulation of the brain cortex; aided mobilization, i.e., tentative movement amplification; or visual feedback in response to effective voluntary movement. In this manner, without having to switch machines, patients can workout in a modality suitable for their evolving clinical status, and, in particular, initiate active exercise sessions as soon as they can produce a minimal voluntary muscular contraction, even below the motor threshold.

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2. Materials and Methods

2.1 Lower Limb Exerciser with Intelligent Alloys (Leia): The Actuating Robot

The functioning of *Leia* (Fig. 1) will be described briefly here, while greater detail on its design and construction is left for a future dedicated report.

Leia is an orthosis mounting a pair of solid state motors based on shape memory alloy NiTi, allowing electromechanical energy conversion and direct generation of torque and rotary stroke. Each one of these actuators contains a sufficient length of trained NiTi wire to accomplish a 40° motion around the ankle joint axis, lifting the foot against viscoelastic resistance and gravity, with a maximal torque of 100 Ncm. This is obtained by coiling the wire along a spiralling sequence of pulleys within the actuator housing. The mechanical design was based on the required stroke and torque, and on a minimization of strain along the wire, so that no section of it is deformed above 4% during actuation. Even with a relatively fast and thin wire (0.25 mm in diameter) and keeping actuator dimensions reasonable for wearing on the human body, it was possible to generate sufficient torque by allowing for a doubled arrangement of the wire, practically resulting in a parallel pull of two equal and perfectly synchronous actuating fibers. This principle and the construction details of the actuator are explained in a pending patent.

Leia is composed of a hinged aluminum structure spanning above and below the ankle joint. The proximal shell encompasses the posterior aspect of the leg to which it can be securely strapped by Velcro®, while the sole of the foot is placed on the distal part of the orthosis. Those contacts provide appropriate interface for the transfer of torque to the patient's limb. At both sides of the hinges, two rotary actuators are fixed, in charge of moving the foot part relative to the leg one, and promoting ankle dorsiflexion. The aluminum structure was designed to guarantee that the axis of rotation of both rotary actuators be constantly aligned with the axis of dorsi-plantiflexion of the ankle joint, and to make *Leia* as lightweight and comfortable as possible.

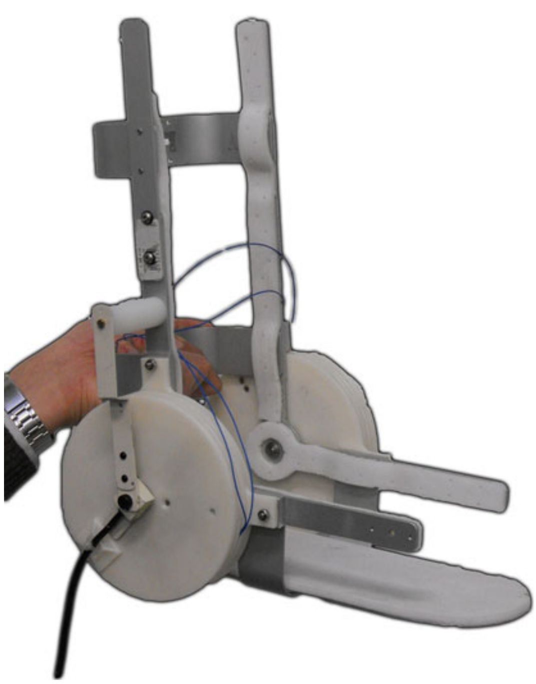

Fig. 1 The active orthosis *Leia*. Rotary actuators are connected in parallel to the ankle joint. By activating the SMA wire inside the actuators, the angle between the leg and the foot part becomes acute, i.e., foot goes into dorsiflexion

Leia can be activated by injection of a direct electric current. In order to provide a full stroke under the weight of a foot, a 30 Vdc 7 s pulse can be applied between the ends of the wires, resulting in around 0.7 A per actuator.

2.2 sEMG

A breadboard prototype of sEMG pre-amplifier with high- and low-band pass filtering stages was assembled using conventional 8-pin PDIP components. Overall amplification gain was 1000. High and low cut-off frequencies were set at 478 and 18 Hz, respectively. Sufficient radiofrequency and common-mode rejection were ensured so that even small signal intensities could be recorded reliably. In addition to this, a feedback loop toward the body (akin to the driven right leg stage used in electrocardiographic architectures) was included. Three Ag/AgCl electrodes (positive, negative, and reference) were used to pick up the signal. Analog waveforms were acquired and digitalised (sampling at 1000 Hz) using an NI9205 (National Instruments, Austin TX, USA) connected to an ordinary laptop computer. A Labview 8.2 routine was used to treat the acquired signal.

2.3 Haptic/Active Control System

The same computer acquiring the sEMG signal was employed to run the control routine. The implementation was also made in the Labview 8.2 programming language. A scheme of the control algorithm is reported in Fig. 2.

A switch is used to select either passive (haptic) or movement amplification (active) mode of use. In the *haptic mode*, therapists can choose the timing of cyclic exercise and the overall duration (number of repetitions) of the exercise session. The control system then switches on and off the power supply to the actuator in an alternate manner, thus creating a repeated dorsi-plantar-flexion motion of the ankle.

In the *active mode*, the therapist has to select two patient-specific sEMG threshold values. The lower one is set to the minimal required level of exercise (which can lie even lower than the muscular motor threshold, in some cases); the upper one is set to an appropriate activation representing the ultimate (or a higher) therapeutic goal, i.e., an effective motion.

The measured sEMG value to be compared with the set thresholds is rectified and continuously filtered to extract only the low-frequency components (e.g., below 3 Hz). The obtained waveform is sufficiently smooth that it can be employed as a measure of instantaneous muscular activation.

At the start of the active exercise session, a visual cue is presented to the patient to dorsiflex the ankle. Then the system is set on hold waiting for the sEMG from tibialis anterior to cross one of the threshold values. If the lower threshold is reached, then the system waits a few milliseconds for the upper threshold also to be reached. If this latter event does not occur before the time-out, then the orthosis power is turned on and *Leia* completes the motor task for the patient. If, on the contrary, the upper threshold is reached, then a visual feedback is provided to the patient that the higher goal was hit, while the orthosis does not intervene to support the movement.

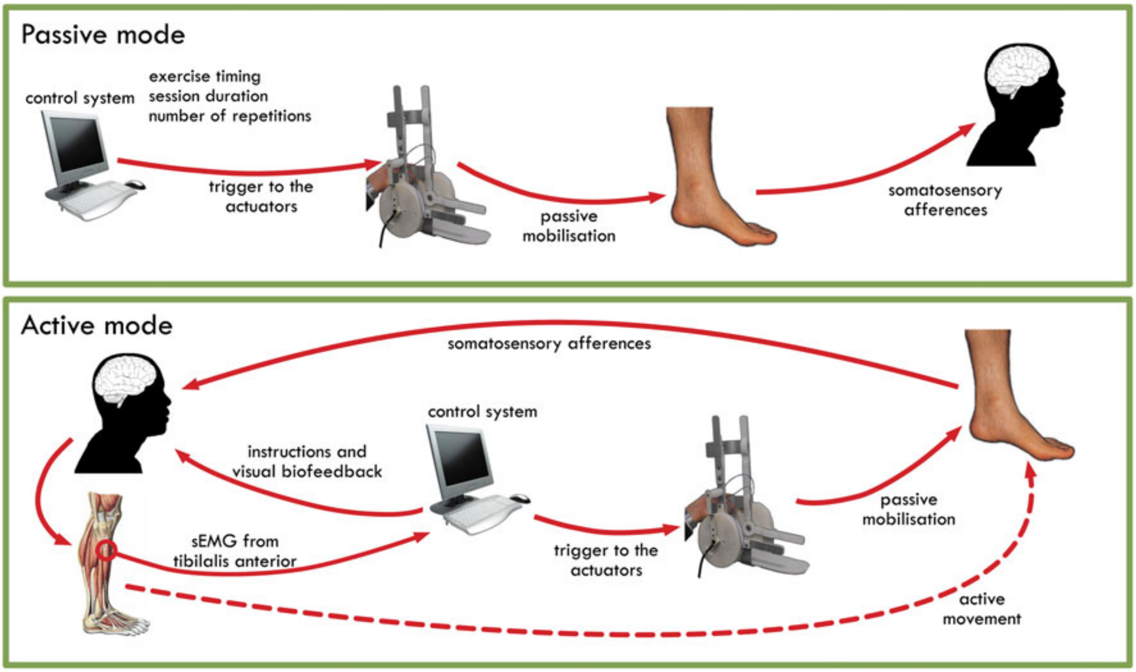

Fig. 2 Schematic representation of the functioning of *Leia* in the haptic (passive) and active modes. In particular, the dashed line in the active mode box stresses the fact that during the course of recovery voluntary movement may not be fully effective

3. Experimental Procedures

3.1 Preliminary Technical Characterization of the Actuators

The presented rotary actuator had already been tested before the beginning of this study. Angular stroke and torque output were measured by attaching different weights (0.11, 0.22, 0.39, 0.58, 0.63, 0.71, 0.76, 0.82 kg) at a set distance of 12 cm from the axis of rotation and powering the wire at 30 Vdc for 7 s. Figure 3 shows the results of these tests. It can be remarked, in connection to the present application, that angular stroke tends to stabilize once sufficient loading is provided to re-deform the wire upon cooling.

3.2 Technical characterization of Leia

Technical tests were carried out on the assembled orthosis before attempting any pre-clinical evaluation. Increasing loads were attached to the foot part of the orthosis at 13 cm from the axis of rotation, while the orthosis was held aloft by a static support. The weights used in this test were 0.22, 0.41, 0.63, 0.83, 1, 1.23, 1.41 kg, each producing a resisting torque in the range 28-180 Ncm. As for the tests on the single actuator, a direct current injection at 30 V for 7 s was applied to the actuators, connected in parallel; then 30 s were allowed for position reset through natural cooling and the action of the weight. The resulting angular upward and downward strokes were measured by means of electrogoniometer SIM-HES-EG 042 (Signo Motus, Messina, Italy).

3.3 Tests on Healthy Volunteers in the Haptic and Active Modes

Three healthy volunteers (28.17 ± 6.08 years old) were enrolled. Ag/AgCl electrodes were placed on the belly of the *tibialis anterior* muscle, the corresponding distal muscle-tendon

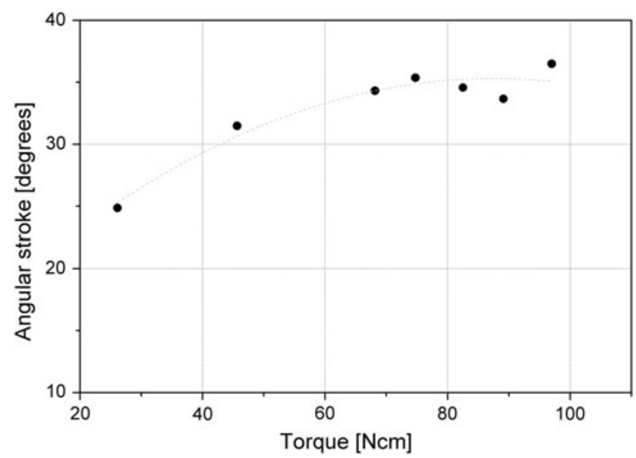

Fig. 3 Angle vs. torque graph of measured performance of the single rotary actuator during preliminary technical characterization. The dashed line is only a guide for the eye

junction, and the internal malleolus (driven electrode). The dominant limb was tested for each subject. Subjects were asked to:

- (TASK 1) Remain still and relaxed without either accompanying the movement or hindering it, while *Leia*, strapped on their legs, was activated in the haptic mode. Actuators were powered in parallel by 10 s 30 Vdc pulses followed by intercalating periods of 30 s, while the resulting angular stroke was measured by means of the same electrogoniometer used in the previous tests. Simultaneously, sEMG was recorded from the *tibialis anterior* muscle.
- (TASK 2) Perform a maximal isometric contraction at the ankle neutral position, then to sustain the minimum

voluntary activation of *tibialis anterior* they could manage. Subsequently, values were set for the lower (110% of minimum individual contraction) and upper (60% of individual maximum isometric contraction) thresholds. Subjects, oblivious as to the functioning of the system, were asked to follow on-screen instructions (graphic and written) trying to respond with just a supra-minimal contraction when cued to dorsiflex the ankle.

4. Results

4.1 Technical Characterization of Leia

Figures 4 and 5 show the results of the orthosis characterization. At lower values of resisting torque (i.e., in the range

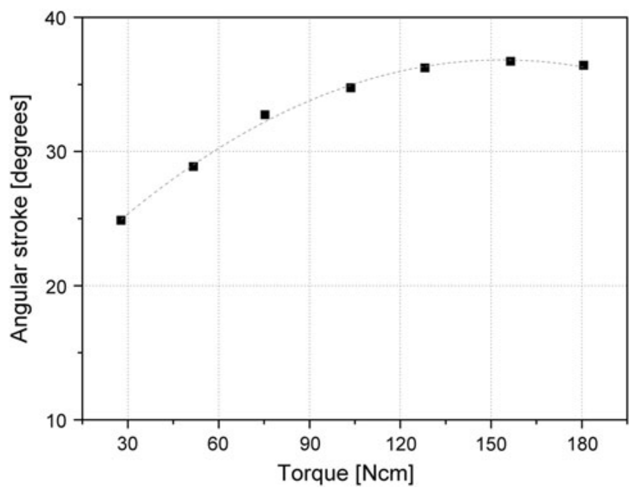

Fig. 4 Angle vs. torque graph of measured performance of *Leia*. The dashed line is only a guide for the eye. Dividing the torque value by two (number of actuators per orthosis), the curve closely resembles the characteristic of the single actuator

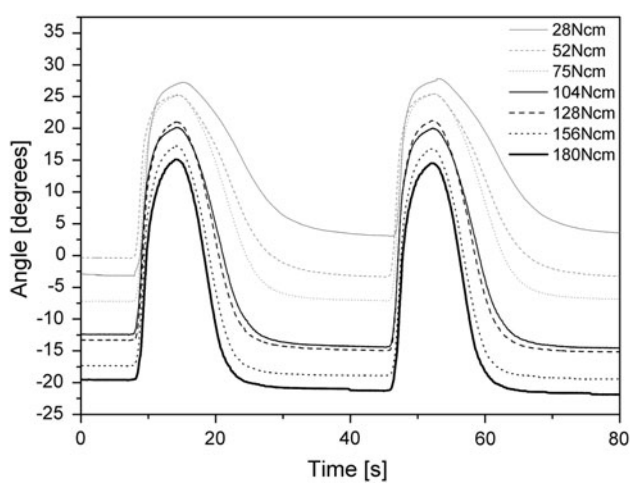

Fig. 5 Activation curves for *Leia* for increasing resisting torques. Larger weights produce a shift of the curves downwards but the behavior is stable in consecutive cycles. It can be noticed that more effective detwinning of the martensite during the cooling phase results in sharper peaks for increasing loading conditions

28-120 Ncm), angular stroke is not stabilized and steadily increases from 24° to 36°. For torques above 120 Ncm, angular stroke is quite stable at 36°. Curves steadily shift toward negative angles (zero being the horizontal position, and negative in the direction of plantarflexion) with increasing torques.

4.2 Tests on Healthy Volunteers in the Haptic and Active Modes

With all subjects, *Leia* was comfortably and firmly secured to the leg and to the foot. Threshold levels for the sEMG activity were set in the above-mentioned proportions to the individual measurements taken for each subject. During TASK 1, the promoted ankle dorsiflexion stroke was $15.85° \pm 1.39°$ with a maximum dorsiflexion angle of $8.24° \pm 4.36°$ and a maximum plantarflexion of $-7.61° \pm 5.71°$. In TASK 2, the subjects were able to produce supra-minimal activation with no significant movement. When the lower threshold was crossed, the system triggered the powering of the orthosis, which completed the movement of dorsiflexion (assisted active session). The measured angle and sEMG time courses are shown, for a representative subject, in Fig. 6. It can be appreciated how passive mobilization can be triggered by a very subtle muscular contraction, which apart from the first cycle (arrow) does not correspond to any effective voluntary movement. The movement produced by the orthosis as a consequence of a minimal contraction brings along some degree of reflex sEMG activity: this may also be thought of as an interesting result to the effect of rehabilitative exercise. In the case of the first cycle in Fig. 6 (and inset), this reflex activity is fairly pronounced, leading to a temporary overshoot above the upper threshold. As programmed in the control software, as long as sEMG tracing (white line) remains above that threshold, powering of the orthosis is switched off: in fact, a short plateau is noticeable in the angle tracing (upper graph, star) just after the upper threshold is crossed; orthosis powering is resumed when sEMG signal crosses back below threshold and power stays on until the end of the pre-set heating time.

5. Discussion

The presented rotary actuator was designed to produce up to 40° against a resisting torque of 100 Ncm. These design constraints were set with the aim of producing extensive ankle movement in a large part of adult population by utilizing two actuators in parallel. In fact, foot weight generates a resisting torque around the ankle in the range of 100-150 Ncm (i.e., 50-75 Ncm acting on each actuator). *Leia*, mounting two actuators, was proved to generate sufficient torque to cover the desired range of motion (at least from −5° to +10°) against resisting torques up to 180 Ncm. The values of output torque and stroke for *Leia* are in perfect agreement with the measurements conducted on the single actuator, and are dependent on the load attached.

The incomplete detwinning of martensite at lower torques suggests that the NiTi wire utilized in the trials with the healthy volunteers was probably too thick for the feet size of the experimental subjects, i.e., for the weight and visco-elastic resistance imposed by their particular limbs. In fact, the use of NiTi actuators often requires a fine adjustment of SMA

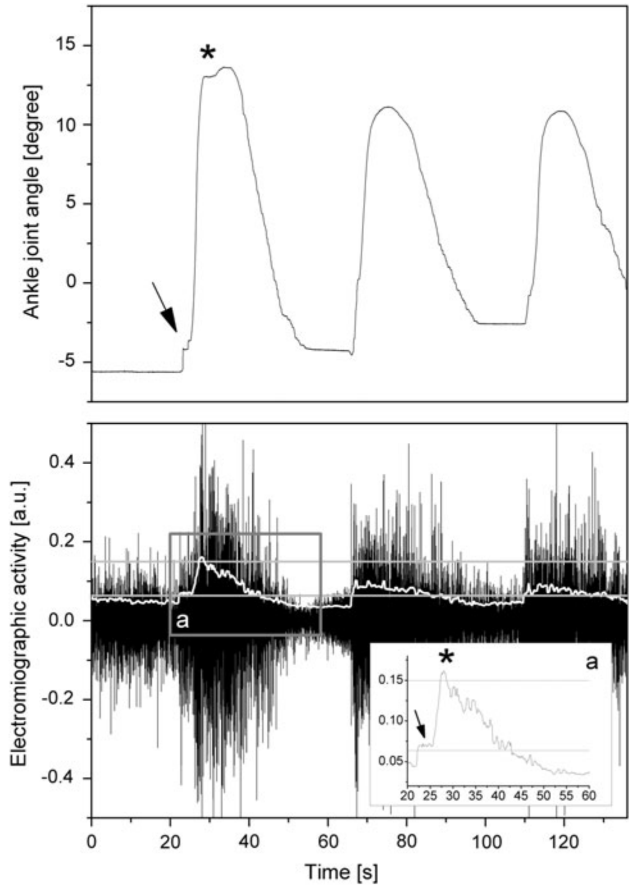

Fig. 6 Ankle angle and sEMG recordings during active mode tests with *Leia* for a representative subject. The lower graph shows sEMG signal (black line) and the same signal rectified and lowpass filtered (white line). In the inset a zoom on the first cycle shows interesting features of the filtered sEMG tracing: an arrow marks the minimal voluntary contraction triggering the switching on of *Leia*; the star marks an interval during which sEMG activity lies above the upper threshold. The same marks can be found also in the angle tracing (top graph): here they indicate the mechanical results of such bioelectric activity (see text for further details)

characteristics, so that sufficient force is produced during the active phase and a limited resisting force withstands position reset. This adjustment is possible, considering the different mechanical characteristics of NiTi austenite and martensite: it may be achieved by selecting a material composition, a thermomechanical history and a wire size, such that a suitable coupling of functional properties is built into the wire. The appropriate wire will only be minimally extended by a given load when in its austenitic state (heating—active phase of the device working cycle), while it will elongate largely through extensive lattice detwinning when its matrix becomes martensitic (cooling—reset phase). As the actuators were designed to be able to mount also thinner gauges of wire, that will be attempted to select appropriate wires for different patients.

It is worth noticing, in any case, that, even at the lower values of measured angular stroke, rehabilitation with *Leia* could still have clinical importance. In fact, during gait, which can be considered the most important functional task for the lower limbs, ankle range of motion is usually limited to a maximum dorsiflexion of 10° (Ref 5): this means that with a repeated angular movement of 15° (the lowest measure presented in this article), ankle ease of movement could be safeguarded and the detrimental consequences of immobility avoided. The preliminary results obtained so far are promising and much improvement will depend on better fixtures to connect the orthosis to patients' limbs: in this manner, it is hoped that most of the available working capability of the construct, as shown by technical tests, will be made exploitable also during exercise sessions.

Even though the tests conducted so far were carried out on healthy subjects, the extremely low value of the myoelectric activity necessary to activate cycling in the active mode and the excellent selectivity of the thresholding algorithm proposed prove in principle that this device can be used to aid dorsiflexion, not only passively but also during the earliest attempts at active work-out coming from a flaccid paresis of the ankle.

6. Conclusions

SMA actuators can be employed safely to build active medical devices for the treatment of recovering paretic patients. The use of this type of technology ensures that suitable articular ranges are exercised and automatically complies with biomechanical constraints. The implementation of dedicated software demonstrated that the activation of SMA wires can be triggered through physiological signals with good repeatability and precision.

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