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Design and Evaluation of a Functional Electrical Stimulation System for Hand Sensorimotor Augmentation

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ABSTRACT

The aim of this study was to perform a preliminary evaluation of a new method for therapeutic exercise of grasping in patients with upper limb disability. The new method combines active voluntary exercise augmented with electrical stimulation and controlled by using force feedback. The feedback has two functions: automatic control of the intensity of electrical stimulation by minimizing the tracking error, and biofeedback to the patient on the computer screen. The force feedback is realized by the use of a newly designed adjustable hand force measuring device, which comprises two force sensors. The therapy requires from patients to volitionally try to open and close the hand while tracking the target on the screen. The system was evaluated in a pilot study in five healthy and two chronic incomplete tetraplegic subjects. Results in healthy subjects were used for reference and for stimulation controller evaluation. The therapy in incomplete tetraplegic subjects of 45-min daily session delivered during four weeks. The results of pilot study show that augmentation of voluntary grip force control with presented system is possible.

KEY WORDS: *Closed-loop functional electrical stimulation, force tracking task, hand training, incomplete spinal cord injury, isometric conditions.*

Introduction

Grasping and manipulating objects requires versatile control of grip forces. Central nervous system injury or disease can result in the loss of sensory and motor functions in the upper extremities, which then reduces hand functionality. Due to such impairment, patients have trouble or are incapable of grasping and manipulating objects (1,2). Patients with spastic finger flexors after stroke or incomplete spinal cord injury usually preserve some control over finger flexion; however, due to spasticity of wrist as well as finger flexors and weakness of finger extensor muscles, they commonly have difficulties with voluntary hand opening (3–5). As a result, while they are able to hold an object, they are incapable of opening the hand to grasp or release an object.

It is important for therapists to assess hand function to evaluate patients' motor functions and to monitor the progress of therapy. Different methods of evaluating hand function are used in clinical practice (6–8). In general, they consist of pick and place and/or volitional range of motion tests. The performance is either evaluated by the therapist according to a numerical scale or by measuring the time needed to complete the tasks. Time is the simplest parameter to measure, but it is not an accurate descriptor

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of hand function (8). Thus, the evaluation of hand function depends primarily on the therapist's judgment. Furthermore, these tests lack information about grip strength. In clinical practice, dynamometers are used to measure maximal voluntary grip force; however, simple dynamometers cannot provide information about submaximal force control (9).

Force tracking tasks have been shown suitable for measuring sensorimotor control capability in submaximal forces (10,11). During tracking, subjects have to track the target as closely as possible by voluntarily controlling force applied to a force sensor. During the task, visual feedback about their performance is provided to the subjects. In addition to evaluation, tracking systems can also be used for training. It has been shown that training using tracking systems improves the accuracy of grip force control and increases voluntary grip strength (11,12). However, no force tracking methods are available for training and/or evaluation of finger extensor muscles.

In addition to conventional therapy, functional electrical stimulation (FES) can be delivered to paralyzed muscles during the rehabilitation process. There is some evidence that FES may improve upper extremity sensorimotor control (13–15). Therapeutic effects of FES, such as reduction of spasticity (14) and strengthening of stimulated muscles (13), also were reported. FES in these studies was open-loop controlled by the patient or the therapist. For a rehabilitative system to be able to autonomously control FES according to reference signal, closed-loop controlled FES should be used (16–18). An interesting approach was presented by Bowman et al. (19), where voluntary control of wrist extension was combined with FES. When the patient voluntarily reached a preselected wrist angle, the FES was triggered, what resulted in full wrist motion. According to Bowman et al.'s observations, the patients gained a psychological boost when they experienced a full range of wrist motion that was caused by their own muscle power, which motivated them to try to voluntarily achieve higher wrist angle.

The aim of our research was to develop and evaluate a novel system for hand sensorimotor augmentation. The system is designed to allow force tracking training of finger flexors and extensors and to provide objective data on training performance. Incorporated FES adds to reduced finger force generation due to injury, thus motivating the user for better achievements. The system consists of a visual feedback display, the hand force measuring device, and the closed-loop controlled electrical stimulator. The system was preliminary evaluated in a pilot study in five healthy and two incomplete tetraplegic subjects. In the next section, the hardware and software used in the system for hand sensorimotor capability augmentation are presented along with the experimental protocol. In the results section, the pilot study outcomes are outlined. Finally, the results are discussed and the conclusion presented.

Methods

Training System

The conceptual scheme of the training system for upper extremity sensorimotor augmentation is presented in Fig. 1. The system is designed to allow voluntary control of hand forces under isometric conditions; FES was added to augment the subjects' compromised force generation due

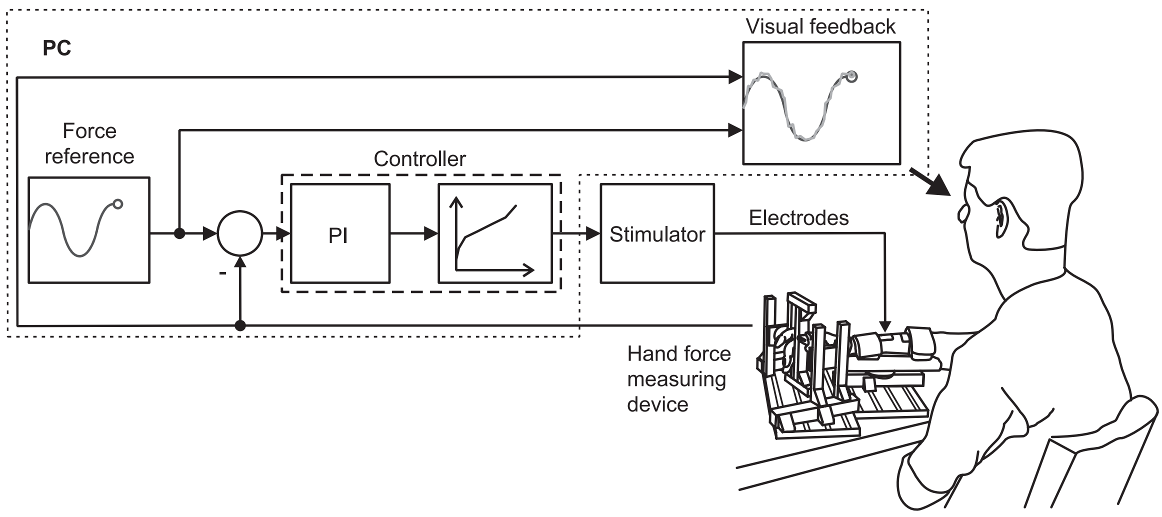

FIGURE 1. Conceptual scheme of the training system showing its main components: hand force measuring device, visual feedback, and closed-loop controlled stimulator.

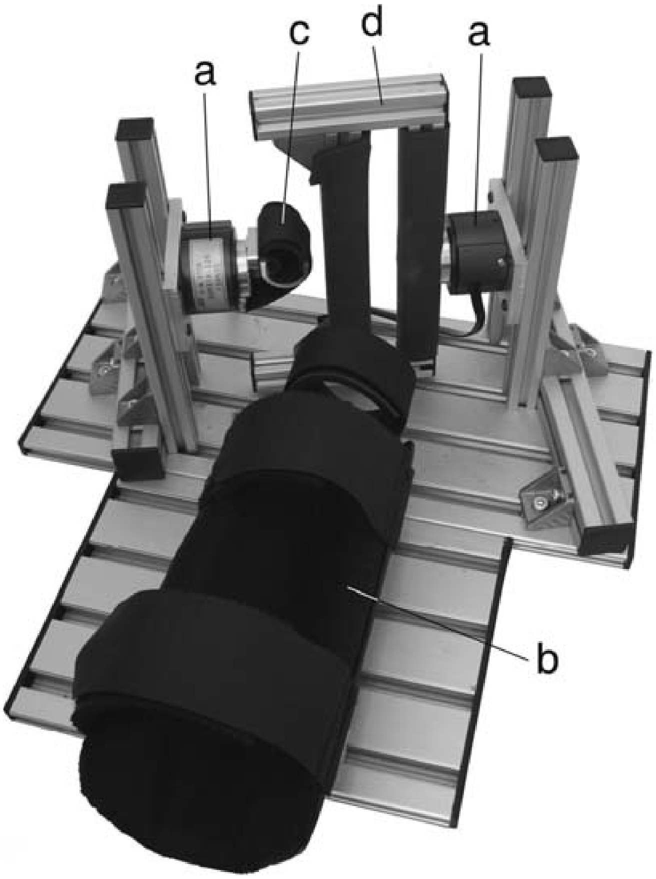

FIGURE 2. Hand force measuring device. (a) Two force/torque sensors for hand force acquisition; (b) forearm support for fixation of the arm; (c) thumb; and (d) index, middle, ring, and little finger support for fixation of fingers.

to the injury. The system is aimed at training finger flexors and extensors by performing the force tracking task. It consists of hand force measuring device, personal computer and stimulator. Personal computer is used for reference force generation, actual grip force acquisition, visual presentation of the reference and hand force, and stimulation control. The software application for controlling the system was developed in the C++ programming language.

The device for measuring the hand force is shown in Fig. 2. The device is made of aluminum strut elements. Two JR3 force/torque sensors (50M31A-I25; JR3 Inc., Woodland, CA, USA) and a forearm support are mounted on a mechanical support. As shown in Figure 2, the left sensor is intended for measuring the thumb force. The thumb is fixed to the force sensor by means of a finger support and a Velcro strap. The right sensor is used for measuring the compound force of the other four fingers. Finger fixation is achieved by two parallel aluminum supports that fully constrain finger motion in the direction of flexion and extension. All finger supports are padded with neoprene material to prevent unpleasant sensations. The finger fixation enables the acquisition of the isometric forces of hand opening and closing. To ensure the proper position and to prevent the arm and wrist from moving during training, the forearm is fixed to the arm support by Velcro straps. The use of the strut supports enables the sensors and forearm support to be adjusted for each individual, as well as to assess either the right or the left hand. A PCI board was used for data acquisition from the force sensors. The data are sampled with a frequency of 100 Hz and then filtered in real time using an on-board integrated filter with a cut-off frequency of 31.25 Hz and a delay of approximately 32 msec.

The force reference signal is composed of four periods of the sinusoidal signal with a superimposed DC component. Between the periods, 15-sec rests are inserted. The amplitude of the reference signal ranges from zero to the maximum positive value for training flexor and from zero to the maximum negative value for training extensor muscles. During the tracking task, the reference and actual hand force are displayed on the monitor screen to provide visual feedback to the subject. The current values of both signals are shown in the middle of the screen. The reference force is marked by a blue ring, and the actual finger force by a red circle. As the flexor force increases or extensor force decreases, the red circle moves upward; when the flexor force decreases or extensor force increases, the red circle moves downward. The past values of both signals are displayed as two time-varying signals, going from the center of the screen to the left side of the screen. During training, the subjects are required to track the reference force as closely as possible by adjusting their hand force. The difference between the momentary value of reference force and finger flexion/extension force serves as the input to the proportional-integral (PI) controller that transforms the difference between both signals into the width of the current stimulation pulse. The integral part of the controller improves the tracking capability; however, it also slows down the controller dynamics. To cancel the nonlinearity of the muscle response, the inverse approximation of the muscle recruitment curve was added to the end stage of the controller. The stimulation parameters are sent to the stimulator via an RS-232 serial connection with a frequency of 33 Hz, which determines the width of each pulse. Two channels of stimulation were used to stimulate the finger flexors and finger extensors independently.

Participants

Two chronic incomplete tetraplegic subjects participated in pilot evaluation of the system. Subject AD was a 28-year-old man who had a spinal cord injury at C5/C6 almost four years earlier. He had strong but spastic finger flexors, which caused him difficulties with finger extension. He

TABLE 1. Demographic Data in Both Incomplete Tetraplegic and Five Healthy Subjects

	Gender	Age (years)	Height (cm)	Weight (kg)	Lesion level	Postinjury time (years)	American Spinal Injury Association impairment scale
Incomplete tetraplegic subjects							
AD	Male	28	182	75	C5/C6	3.75	C
AS	Male	15	192	90	C3/C4	0.75	D
Healthy subjects							
AH	Male	26	174	75	/	/	/
JP	Male	27	175	68	/	/	/
JR	Male	30	172	78	/	/	/
PC	Male	29	185	105	/	/	/
RK	Male	40	174	78	/	/	/

was highly motivated to improve hand functionality, especially hand opening, which he could partially achieve passively using a tenodesis grasp. Subject AS was 15 years old; he had incomplete tetraplegia at C3/C4 due to an injury eight months earlier. He had considerable voluntary control over finger flexor and extensor muscles. Both subjects used the experimental system in addition to their regular treatment at the Institute for Rehabilitation, Republic of Slovenia. In addition, five healthy subjects with no previous history of upper extremity problems participated in the preliminary evaluation of the system. This preliminary study was in order to assess average tracking performance and maximal hand force of healthy persons and to evaluate the controller performance. Data of the participants are presented in Table 1. All subjects gave informed consent to participate in the study.

Experimental Protocol

The experimental training in two incomplete tetraplegic subjects lasted for four weeks. Five training sessions were completed each week, one session per working day. In both subjects, the dominant right hand was trained. The training was supervised by an experienced physiotherapist. During training, the subject was seated behind the desk in front of a computer screen (Fig. 3).

Each training session began by adjusting the system to the individual's needs. The physiotherapist positioned one of the surface electrodes on the distal forearm near the wrist (1–3 cm from radial and ulnar styloid process), while she moved the other electrode across the subject's forearm to obtain appropriate flexion/extension of the index, middle, ring, and little fingers without flexion/extension and ulnar/radial deviation of the wrist. During electrode placing, the subject held his arm in the same position as in the hand force measuring device (elbow in about 75° flexion, wrist in 0° flexion/extension, and 0° ulnar/radial deviation). After placement of the electrodes, the fingers and forearm were fixed onto the force sensors and the forearm support, respectively. By not changing the arm position, the same response to the stimulation during training was assured. The maximal amplitude of the stimulation pulse was determined by manually increasing the pulse amplitude to the level at which the subject felt that the stimulation was uncomfortable. During maximal amplitude adjustment, the stimulation pulse width was set to a maximum value of 500 μsec. The maximal pulse amplitude determined during this initial testing was then used throughout the training session. A model of electrically stimulated muscle under isometric conditions was used to tune the PI controller, that is, to set the gains KP and KI. The muscle was represented by the Hammerstein

FIGURE 3. Incomplete tetraplegic subject accomplishing the force training task with the system for upper extremity sensorimotor augmentation.

model (20), which consists of nonlinear static recruitment and a linear discrete-time transfer function. To derive the recruitment curve of the isometric muscle response, linearly increasing stimulation pulse width was applied to the muscles from 0 to 500 µsec in increments of 10 µsec. Five identification trials were performed under the same conditions. The results were averaged to obtain an approximation of the muscle recruitment curve, which represented the nonlinear relationship between the pulse width and the finger force. In the next step of muscle response identification, the response to the pseudo-random binary signal was measured and used to determine the linear discrete-time transfer function.

To tune the PI controller, the discrete model of the closed-loop FES system and the muscle model was developed using the Matlab-Simulink simulation environment. Parameters KP and KI were defined by the optimization procedure, which minimized the tracking error. The Simulink Response Optimization toolbox was used for this purpose.

After successfully adapting the system to the individual, the maximal force that the subject was able to achieve voluntarily was determined. The forces measured from both force sensors were used for this purpose. The training consisted of three repetitions of two tracking tasks: *Task A* followed by *Task B*. In *Task A*, the subject had to accomplish the tracking task only by voluntary activity of finger flexors/extensors, while in *Task B*, the FES was added to augment the subject's voluntary effort. During tracking, the reference and actual force signal and the stimulation output were sampled at a frequency of 100 Hz and saved in a file. The training procedure was repeated separately for training the finger flexors and extensors.

In the preliminary evaluation, five healthy subjects tested the system for four successive days. Each day, the maximal hand closing and opening force was measured prior to the tracking tasks. The maximum value of reference force was then set at about 50% of the measured value. Each subject repeated *Task A* three times for finger extensors and finger flexors, respectively. The controller was evaluated during the last, fourth day, separately for flexor and extensor muscles. The adjustment of the controller was accomplished in the same way as in both incomplete tetraplegic subjects. In the evaluation of the tracking capability of FES controller, the healthy subjects were asked not to voluntarily interfere with the FES activity of their muscles during the tracking task. For this purpose, the visual feedback was not provided to the subjects. The amplitude of the reference signal was for easier toleration of pain caused by stimulation set to about 40% of maximal hand force.

To assess tracking performance, the relative root mean square error (rrmse) of each tracking task was calculated according to (Equation 1):

TABLE 2. Results of the Training System Evaluation in Healthy Subjects. Maximal Hand Forces, Tracking Errors (rrmse) for the First and Fourth Days, and Tracking Errors From FES Guided Tracking

Healthy subjects		Voluntary activity F_{max} (N)	Voluntary activity rrmse first day	Voluntary activity rrmse fourth day	Only FES rrmse
AH	FLX	128.80 ± 15.00	0.30 ± 0.05	0.17 ± 0.05	0.52
JP	FLX	133.70 ± 14.81	0.20 ± 0.02	0.20 ± 0.02	0.29
JR	FLX	128.08 ± 22.25	0.23 ± 0.02	0.23 ± 0.01	0.68
PC	FLX	146.33 ± 37.90	0.35 ± 0.03	0.32 ± 0.02	0.50
RK	FLX	209.05 ± 19.00	0.25 ± 0.04	0.18 ± 0.01	0.37
AH	EXT	42.73 ± 4.28	0.62 ± 0.22	0.25 ± 0.22	0.25
JP	EXT	63.00 ± 6.78	0.40 ± 0.22	0.29 ± 0.02	0.23
JR	EXT	54.75 ± 26.21	0.35 ± 0.06	0.37 ± 0.05	0.35
PC	EXT	66.58 ± 8.94	0.70 ± 0.26	0.67 ± 0.04	1.02
RK	EXT	76.35 ± 9.09	0.45 ± 0.26	0.47 ± 0.22	0.42
Average	FLX	149.19 ± 37.68	0.26 ± 0.06	0.22 ± 0.06	0.47 ± 0.15
	EXT	60.68 ± 16.72	0.51 ± 0.22	0.41 ± 0.17	0.46 ± 0.33

EXT, finger extension; FES, functional electrical stimulation; FLX, finger flexion; rrmse, relative root mean square error.

$$rrmse = \sqrt{\frac{1}{T}\sum_{t=0}^{t=T}\frac{(F_{ref}(t) - F_{act}(t))^2}{\max(F_{ref})^2}} \quad (1)$$

where F_{ref} stands for the reference force, F_{act} denotes the actual grip force, and T stands for the signal's duration. The tracking error was normalized by the maximal value of the reference signal to allow the results to be compared.

Results

In Table 2, the results in five healthy subjects are presented. The table shows the maximal hand force (F_{max}) (mean ± SD) and tracking error (rrmse) (mean ± SD) for the first and fourth days. The last column shows tracking error of FES controller tracking without voluntary control of the subjects. At the bottom of the table, the average results (mean ± SD) in all five subjects are presented. The average results from the fourth day served as reference values for comparison with both incomplete tetraplegic subjects.

In Fig. 4, an example of the tracking performance of *Task A* and *Task B* during training of finger flexors in subject AD is presented. The upper graph shows the tracking performance over four reference signal periods during voluntary activity (*Task A*). Large deviations can be observed, especially when the subject sustains the grip force for a longer time; better tracking is noted for the shortest period of the reference force. The tracking performance in *Task B* is presented in Fig. 4B and shows that FES considerably improved force tracking performance. In order to illustrate the FES contribution, the stimulation intensity represented by the pulse width is shown in Fig. 4C.

TABLE 3. The Average Maximal Hand Force (Mean ± SD) and the Average Tracking Error (rrmse) (Mean ± SD) in *Task A* in Both Incomplete Tetraplegic Subjects as Obtained in First and Last Five Sessions

Incomplete tetraplegic subjects		First five sessions		Last five sessions	
		F_{max} (*N*)	rrmse	F_{max} (*N*)	rrmse
AD	FLX	65.60 ± 8.28	0.42 ± 0.30	93.52 ± 11.16	0.67 ± 0.83
AS	FLX	54.42 ± 8.55	1.50 ± 0.81	58.10 ± 4.39	1.20 ± 0.78
AD	EXT	9.40 ± 8.08	0.61 ± 0.87	31.98 ± 4.60	0.29 ± 0.05
AS	EXT	30.96 ± 6.07	3.98 ± 2.72	65.52 ± 8.86	1.02 ± 0.60

EXT, finger extension; FLX, finger flexion; rrmse, relative root mean square error.

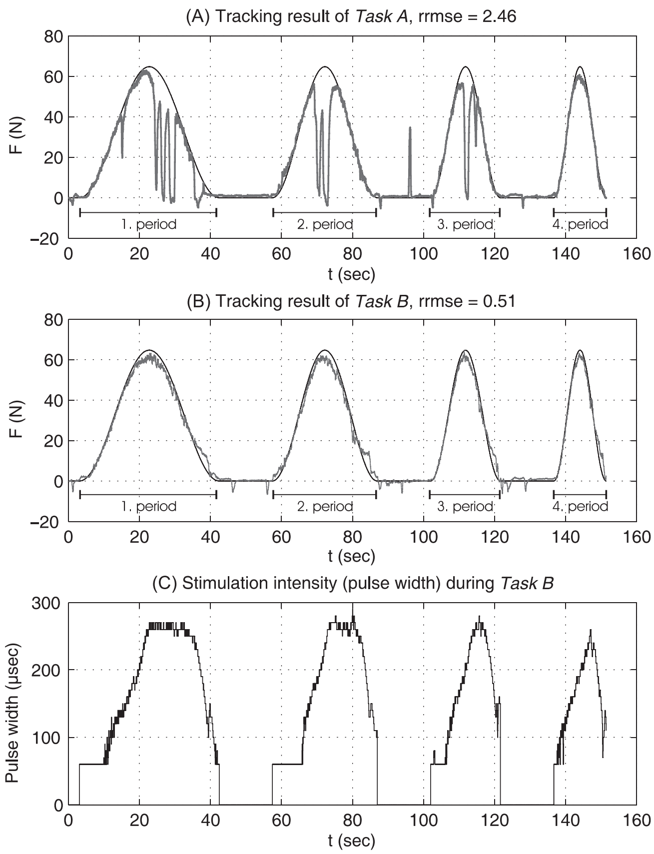

FIGURE 4. Force tracking of subject AD during training of finger flexors. (A) *Task A*; (B) *Task B*; and (C) stimulation intensity (pulse width) during *Task B*.

The maximal voluntary hand forces as assessed in both incomplete tetraplegic subjects are presented in Fig. 5 and Table 3. Subject AD demonstrated a steady improvement of maximal voluntary force in hand closing and opening during the whole training period. His hand opening force was increased by 240% (from 15% to 52% of healthy subjects' reference value) and his hand closing force was improved by 40% (from 44% to 63% of healthy subjects' reference value). Subject AS also achieved steady improvement of maximal voluntary force in hand opening. He improved his hand opening force by more than 100% (from 51% to 108% of average hand opening force achieved by healthy subjects). His maximal voluntary force of hand closing quickly improved during the first day of training and remained relatively constant afterward. The average improvement of hand closing force was 6% (from 15% to 52% of healthy subjects' reference value).

Tracking error results, presented in Table 3 for the first and last five sessions in *Task A*, show that subject AD doubled his tracking ability in extensors tracking task and achieved low variability in performance (SD = 0.05). In flexors tracking task, his average tracking in last five sessions deteriorated as compared to first five sessions. Subject AS improved tracking with finger flexors for 20%. His performance in finger extensor tracking task was inconsistent in the first sessions, resulting in large rrmse and SD, but he improved his tracking during training, resulting in smaller rrmse and SD.

Discussion

A novel system for training finger flexor and extensor muscles under isometric conditions was developed. The system is based on closed-loop control of antagonistic muscle groups, while simultaneously promoting subject's voluntary efforts through biofeedback loop based on tracking approach. Five healthy subjects and two incomplete tetraplegic subjects participated in the experimental evaluation of the system.

In evaluation of the system with five healthy subjects, tracking performance and maximal hand force were assessed. These results served as reference values for comparison with both tetraplegics. Five healthy subjects also participated in the evaluation of FES controller performance. The tracking errors in finger extension tracking task showed that errors in FES controller tracking are similar to those obtained in voluntary tracking. Somewhat worse results of FES controller performance were observed in finger flexion tracking task.

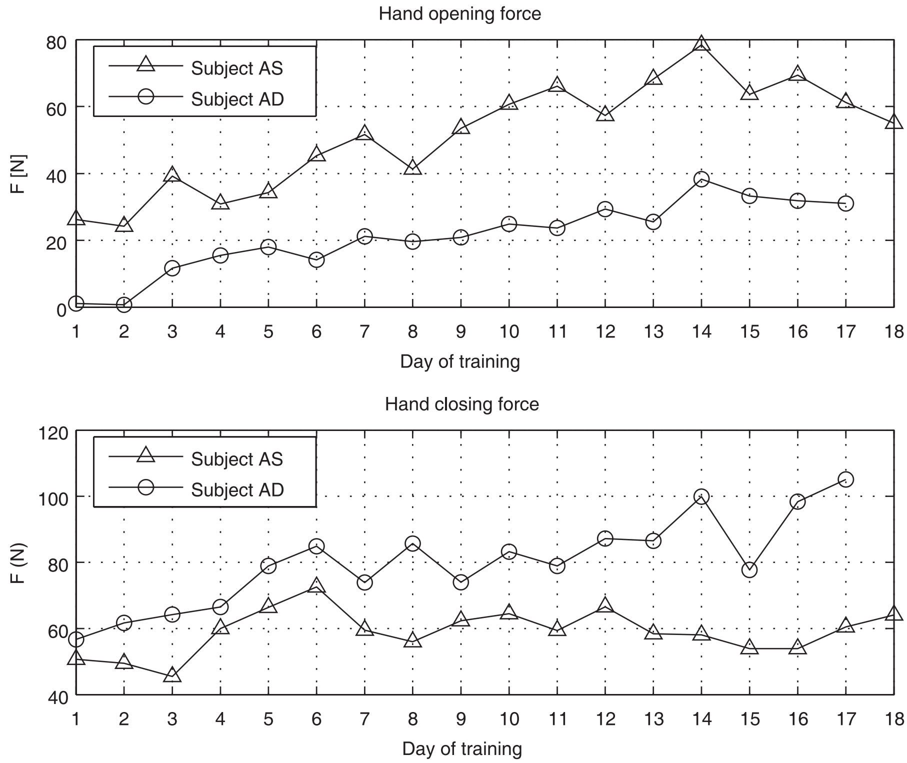

FIGURE 5. Maximal voluntary hand forces in both incomplete tetraplegic subjects during training. Maximal voluntary hand opening and closing forces are shown in the top and the bottom graphs, respectively.

Based on maximal hand force measurements as assessed in both incomplete tetraplegic subjects during training, we observed that subject AD showed a steady improvement of maximal voluntary hand force in hand closing and opening. Subject AS also had steady improvement of maximal voluntary force in hand opening. Surprisingly, subject AD, who was in chronic phase after the injury, considerably improved maximal voluntary hand force in hand opening and hand closing. Better results were expected for subject AS, as only nine months elapsed since injury. However, he improved his hand opening force, which was at the end of training comparable to the ones assessed in healthy subjects.

Subjects tracking results show that subject AD improved his performance in finger extensor tracking task. At the end of training, his average tracking performance was even better than the average tracking of healthy subjects. In finger flexion tracking task, the average tracking results for subject AD indicate that his tracking capability decreased. However, larger average rrmse is a result of worse performance on the last day, likely due to reduced motivation and muscle control caused by fatigue. The tracking results of subject AS show that his tracking performance was inconsistent; however, some improvement was observed. Better results were observed in finger extension tracking. At the beginning of the training, during longer activation of finger extensors, finger flexor spasms affected his grip control, resulting in large rrmse and SD. On the other hand, the spasms did not occur during tracking of shorter periods. Toward the end of the training, the number of spasms was minimized; this can be attributed to the therapeutic effects of FES.

Conclusion

The results of training in two incomplete tetraplegic subjects suggest that augmentation of voluntary grip control and increased hand strength is possible with presented system. It was hypothesized that "carry-over" effect of FES training can only occur when FES is combined with simultaneous voluntary effort (21). Visual feedback provided through tracking task promotes subject's voluntary involvement in the rehabilitation process. Larger studies are needed to fine tune the new therapy and adapt it for different users.

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Conflict of Interest

The authors reported no conflict of interest.

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