Pulsed assistance: a new paradigm of robot training

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***Abstract* —In this preliminary study we compare continuous with pulsed robot assistance in five chronic stroke survivors with a mild degree of spasticity, with the aim of promoting volitional effort and reducing assistance during a reaching task. The protocol consists of one familiarization session and a single training session during which a manipulandum provides subjects with pulsed or continuous assistance in random order. The basic level of assistive force is calibrated for each subject and is the same for both modalities; however, the average force during continuous assistance is about twice the average force in pulsed assistance. In spite of this, the results show that pulsed assistance allows subjects to reach similar performance levels as compared to continuous assistance after a single training session. Moreover, we introduce a novel kinematic-based measure to assess voluntary participation of subjects during the rehabilitation task, which is only applicable with pulsed assistance.**

Keywords — robot therapy; pulsed assistance; stroke survivors.

I. INTRODUCTION

Recent reviews [1,2] suggest that neural plasticity is now recognized as a fundamental property of the human brain that can be exploited with success in upper-limb neuromotor rehabilitation. This is especially true if considering its involvement in all daily activities, both in childhood and old age, in physiological as well as many pathological conditions, in chronic and acute/subacute phases, etc. For these reasons, it is necessary to revise the conception and design of robot therapy in order to promote neural plasticity and enhance motor learning/re-learning. For instance, it is advisable to engage patients in task-oriented actions rather than in passive mobilization, viewing training (for rehabilitation) as part of the motor learning process. In fact, task oriented training has emerged as a leading concept in clinical practice. Accordingly, in a recent paper [3] we argued that it is not movement per se, obtained for example by means of passive mobilization, which is effective in recruiting plastic adaptation. In contrast, the key issues are movement associated with a task [4] and volitional effort [5]. In other words, in order to promote plasticity, a causal relation between *intended actions, actual movements*, and the corresponding *feedback reafference* must be operational.

The questions then are: How can volitional effort be evaluated? How can it be enhanced? Generally speaking there are two ways to detect motor intentions of a motor impaired subject: 1) via physiological signals or 2) via modulated haptic interaction. In parallel there are two corresponding ways to enhance voluntary control: 1) via electrical assistance (functional electrical stimulation) or 2) via robotic assistance. Recent studies demonstrated that electrical assistance could be quite effective if synchronized with the detection of event-related desynchronization signals from cortical activity [6] or by using a contralateral-homonymous paradigm [7]. On the other hand, it appears that for practical reasons this type of assistance is more easily applicable to single muscles or small muscle groups, while the haptic assistance provided by a robot/human therapist can affect large muscle groups and multi-joint movements. Thus, the two methods of assistance (electrical and haptic) are complementary and we believe they will ultimately be integrated, with increased beneficial effects. For the purpose of this study, however, we will focus on the haptic part.

The intention to move, even in severely motor impaired subjects, can be detected directly or inferred/promoted indirectly. Examples of the former approach are provided by the “contralateral-homonymous paradigm” mentioned above or by “body machine interfaces” that extract motor intentionality even from extremely reduced mobility [8]. The indirect approach to the promotion of intentionality can be rephrased by saying that robot assistance must be controlled/modulated in such a way to avoid “slacking”, defined as a reduction of subject’s voluntary control during repetitive, passive mobilizations [9,10]. In previous papers we demonstrated that patterns of *minimal assistance* based on force fields with null mechanical impedance [11-13], which indeed avoid the insurgence of slacking, enhance proprioceptive awareness, regarded as the ability of a subject to perceive his own limb in space relying on kinesthetic information [14,15] and decrease arm stiffness during functional recovery [16]. Here we move a step ahead in the same direction by proposing a novel robot assistance paradigm, where the assistive force field, still with null mechanical impedance, is pulsed in time, with a repetition frequency of 2 Hz, which is compatible with recent theories on intermittent control [17,18]. In this preliminary study, we compare continuous with pulsed robot assistance and we show that the proposed method is at least as effective as the continuous assistance method in promoting functional recovery, but it employs a significantly lower average value of assistive force, reducing the risk of slacking. Moreover, we

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introduce a novel kinematic-based measure to assess voluntary participation of subjects during the rehabilitation task.

II. Methods

A. Subjects

Five stroke survivors (all females, 45.6±12.5 years old) participated in this study. Subjects were recruited among those followed as outpatients of the ART Education and Rehabilitation Center in Genoa (see Table I for relevant information). The patients were selected according to the following criteria: 1) diagnosis of a single, unilateral stroke verified by brain imaging; 2) sufficient cognitive and language abilities to understand and follow instructions; 3) chronic condition (at least 1 year after stroke); 4) stable clinical conditions for at least one month before being enrolled in this study. This preliminary clinical study did not include a control group and thus is not a randomized, controlled clinical trial. However, the functional assessment was blinded. The research conforms to the ethical standards laid down in the 1964 Declaration of Helsinki, which protects research subjects, and was approved by the ethics committee of the regional health authority. Each subject signed a consent form conforming to these guidelines. The robot training sessions were carried out at the Motor Learning and Rehabilitation Lab of the Istituto Italiano di Tecnologia (Genoa, Italy), under the supervision of experienced clinical personnel and engineers. All subjects underwent clinical evaluations before starting the present study to ascertain their degree of spasticity and residual functional level.

TABLE I. Demographic and clinical data of the subjects

Subject	Age	Paretic hand	FMA (0-66)	ASH (0-4)	F_A [N] ini - fin
S1	37	L	15	2	8.63 – 7.24
S2	39	R	28	1+	4.90 – 6.14
S3	63	L	55	1	3.87 – 3.60
S4	58	R	33	1+	4.94 – 4.98
S5	31	L	21	2	9.79 – 6.34

Age [years]; FMA, arm portion of Fugl-Meyer score (0-66) at the time of the study; ASH, Modified Ashworth scale of muscle spasticity (0-4) at the time of the study; F_A, holding force (defined as the mean force that the robot has to exert to stabilize a subject in different points of the workspace), evaluated at the beginning **(ini)** and at the end (**fin**) of the test session.

B. Experimental setup

The planar manipulandum *Braccio di Ferro* [19] provided assistive forces for helping stroke survivors to accomplish the task. The subjects sat in front of the robot with their shoulders strapped to a chair, holding the end effector of the robot with their impaired hand. The hand and the shoulders were securely fastened using a custom made cast. The vertical position of the robot was adjusted for each subject in order to keep the forearm approximately horizontal. Moreover, to guarantee a sliding movement of shoulder and elbow with low-friction and no influence of gravity along the horizontal plane, a light, soft support was connected to the forearm. The position of the seat was chosen so that almost full extension of the arm was required to reach the farthest points on the workspace. A 19” LCD screen with a 1:1 scale factor was positioned in front of the patient to provide visual feedback for the task.

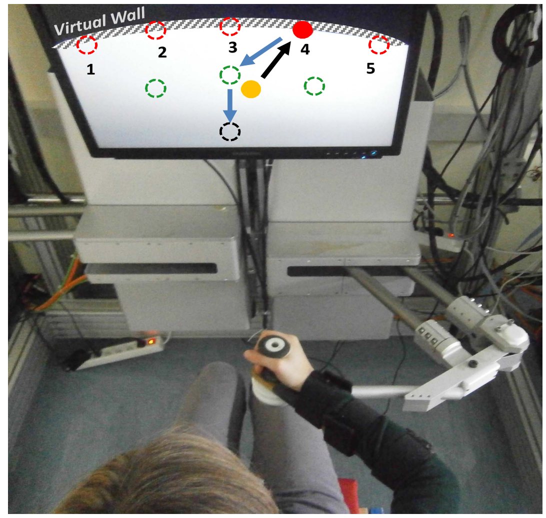

Fig. 1. Experimental setup: the subject sits in front of a screen, holding the manipulandum. The red circles represent the ‘far’ T targets (the filled red circle is the current, visible target). The green circles are the intermediate I targets. The black circle is the starting S target. The yellow filled circle is the current hand position, aimed at the current target (black arrow). The two blue arrows correspond to the two following inward movements. The circles have a 2cm diameter.

C. Task and protocol

The protocol was designed in order to promote the active execution of large outward movements that require arm extension and lateral rotation of the shoulder, which are more severely impaired in the subjects involved in our study, while inward movements could be executed almost without assistance. Five ‘far’ targets (T) were arranged at a distance of 26 cm from a starting position (S) for center-out movements. Three intermediate targets (I) were added for the return movements, at distance of 13 cm from S, as shown in Fig.1. The task consisted of reaching one target, chosen randomly in the set of points, with a threshold of 2 cm. Reaching sequences followed the scheme: $S \rightarrow T \rightarrow I \rightarrow S$. An acoustic feedback signaled that a reaching movement was completed and a 1s pause was introduced before presenting a new target in the sequence.

Two conditions were tested, namely *vision condition* and *no-vision condition*: in both cases haptic feedback was provided to the subjects by the robot generated force field. In the former condition a visual feedback was provided on the computer screen (Fig. 1). On the contrary, in the latter condition subjects were asked to wear a suitable mask to eliminate visual feedback relying on the proprioceptive channel in order to understand the direction of the assistive force field and thus complete the task.

Haptic feedback was generated by the robot according to the following control law:

$$F(t) = A(t) - B\dot{x}_H - K_W(x_W - x_H) \tag{1}$$

where x_H is the hand position vector, $A(t)$ is the time-varying assistive force field (see below), B is a viscosity coefficient (12Ns/m) acting as damping factor on the hand, K_W is the stiffness (1000 N/m) of the "virtual wall", shown in Fig. 1, that prevents the hand to go beyond the target distance, and x_W is the projection of x_H on the wall. The last component of the force field is activated only when the hand reaches the "wall".

The protocol included two sessions on two separate days: a familiarization session and a test session made by two *evaluation blocks* and a single *training block*.

In the *training block* the subjects were asked to complete a total of 6 target-sets (3 with vision, 3 without vision); each target-set consisted of 30 outward movements ($S \rightarrow T$) and 60 inward movements ($T \rightarrow I$ and $I \rightarrow S$), for a total of 90 movements.

The *evaluation blocks* aimed at estimating the average *holding force* F_A applied by the robot to the subjects in order to keep them stable in different positions of the workspace[1]. The subjects were instructed to relax and let the robot passively move their hand on each target in the training trials. The robot held the different positions for 1s, averaging the corresponding holding force. The overall mean force was taken as the estimate of F_A. The two evaluation blocks were at the beginning and at the end after the training session. The first block aimed at automatically selecting the level of assistive force used in the training trials; the second block at the end of a session was to test the effect of training on muscle tone by detecting changes in the *average holding force*. Table I reports the values obtained for the estimate of F_A in the initial and final evaluation blocks for each subject.

D. Generation of the assistive force field

Two mechanisms of generation of the assistive force field were used in the experiments: 1) *constant field*, 2) *mixed field* made by a constant component plus a periodic sequence of force pulses. In both cases, the force field was convergent to the current target independently on the subject interaction force; moreover, the field was turned on smoothly (by gating the field generator with a ramp-and-hold function) and was turned off suddenly as soon as the target was reached. Both mechanisms are represented by the following generation function:

$$A(t) = [kF_A + (1-k)F_A \cdot I_{\Delta t}(t)] \frac{(x_T - x_H)}{\|x_T - x_H\|} \cdot R(t) \qquad (2)$$

where F_A is the previously defined holding force; $R(t)$ is a ramp-and-hold function (rising time 0.1s); x_T is the target position and x_H the hand position; $I_{\Delta t}$ (t) is a smooth impulse of duration Δt, unitary peak value, repeated with a frequency of 2 pulses/s (top panel of Fig. 2); k is a weighting factor ranging from 0 to 1, which selects between the *constant field* and the *mixed field* mechanisms. In the experiments we used k=1 for the pure *constant* field and k=0.5 for the balanced *mixed field*. In particular, for the force impulse we used the following minimum jerk profile:

$$\xi = t / \Delta t$$

$$I_{\Delta t}(t) = \begin{cases} \frac{1}{1.875}\left[30\xi^4 - 60\xi^3 + 30\xi^2\right] & \text{for } 0 \le \xi < 1 \\ 0 & \text{for } 1 \le \xi \le T/\Delta t \end{cases} \qquad (3)$$

T=0.5s is the repetition period; Δt=0.2s is the duration of the impulse, followed by a 'refractory time' of 0.3s.

It is worth observing that the average value of assistance in the pulsed/mixed case is always smaller than in the constant case, where it is equal to F_A, because the following relation holds for all values of k smaller than 1:

$$kF_A + (1-k)F_A \hat{I}_{\Delta t} < F_A \qquad (4)$$

$\hat{I}_{\Delta t}$=0.0618 is the mean value of the pulse of unitary peak height in the period T.

For the purpose of our experiment, we randomized the type of assistance in the trials of the training blocks, balancing the number of trials with continuous assistance (k=1) and trials with mixed assistance (k=0.5). For one subject (S3), who exhibited a value of F_A smaller than 4N, which was close to the subjects' kinesthetic threshold [21], the amplitude of the pulses in the mixed trials was increased from $0.5{\cdot}F_A$ to $0.75{\cdot}F_A$ in order to facilitate force perception during trials without vision.

E. Data analysis and performance measurements

Hand trajectories in Cartesian coordinates were reconstructed from the robot primary encoders (17-bit, positional end effector resolution lower than 0.01 cm). The forces transmitted from the robot to the hand of the subjects were estimated from the input current to the motors. Both vectors were recorded at a sampling rate of 100 Hz. The first three time derivatives of the hand trajectories were estimated by using a 4th order Savitzky-Golay smoothing filter with an equivalent cut-off frequency of 9Hz. The analysis of the results was limited to the outward movements, which are more relevant for the purpose of this study.

Mean trajectories were evaluated for each subject in the 4 different training conditions: continuous assistance with vision (**VC**), pulsed assistance with vision (**VP**), continuous assistance without vision (**C**), pulsed assistance without vision (**P**). Performance in all the different conditions was quantified according to the following three indicators[2] [14], plus a novel indicator (AC) specifically used only for the mixed field:

1. *Mean speed of movement* (V_m; in m/s): it is the mean value of the speed computed by the time of target presentation considering a speed threshold on 0.01 m/s, to the instant in which the target is reached. We define T_{tot} this time interval.
2. *Endpoint error after the first submovement* (E_1; in cm): it is measured as the distance between the target and the hand position at the end of the first submovement, which is identified on the speed profile by two consecutive

[1] In control subjects the holding force is close to 0 [15].

[2] The fourth indicator used in [14], namely the number of sub-movements (that is the number of peaks in the speed profile), has been discarded here, due to the low reliability of this measure in trials with pulsed assistance.

minima, one before and one after the first point of peak velocity. It ranges from 0 to 26 cm.

3. *T-ratio* (*TR*; a-dimensional): it is defined as the ratio between the duration of the first submovement and T_{tot} .
4. *Active Contribution* (*AC*; a-dimensional). It ranges from 0 to 1. In the trials with pulsed assistance it is meant to discriminate the relative contribution of active and passive components to the overall response of the subjects, i.e. it attempts to quantify the phenomenon known of 'slacking' [20]. We computed *AC* in the following manner:

$$AC_n = \left\| \sum_{j=1}^{P_i} \vec{v}_j + \sum_{k=1}^{R_i} \vec{v}_k \right\| \tag{5a}$$

$$AC = \frac{1}{N} \cdot \frac{\sum_{n=1}^{N} AC_n}{\max(AC_n)} \tag{5b}$$

$\vec{v}$ is the hand velocity vector, P is the number of samples in the active impulse phase (20 samples), and R the number of samples in the refractory phase (30 samples); AC_n then is the norm of the vectorial sum of velocities during a single impulse period (0.5s). The rationale of the indicator is illustrated by Fig. 2, which shows a trial with pulsed assistance. The speed profile and the corresponding hand trajectory are divided in segments of different colors: red, for the segments that correspond to force impulses, and blue/black for the segments of the 'refractory time'.

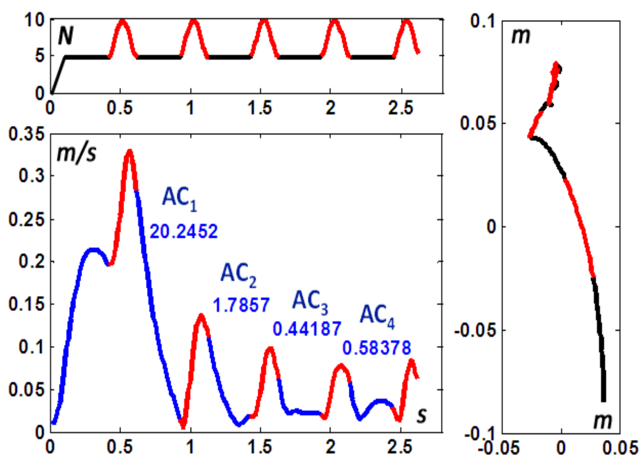

Fig. 2. Example of calculation of the *AC* index. The left, top panel plots the profile of assistive force and the bottom panel the corresponding velocity profile. The red portions of the profiles identify the phases of pulsed assistance. The panel on the right holds the hand trajectory [m].

Therefore, *AC* is the mean value of the trial's partial contributions, normalized by the maximum value amongst them. This normalization is fundamental to compare the values of *AC* in subjects who operate with different assistive forces. This indicator gives a measure of the coherence between the sub-movements executed during the force pulse with respect to those in the force refractory time. The more the direction of displacement is preserved after the termination of an impulse *n*, the more its partial contribution indicator AC_n increases. Consequently, the more the trajectory is rectilinear, the higher the AC_n score will be. This implies that the trajectory sections in which a passive displacement occurs will lead to a partial score that is close to zero. Conversely, the maximum value (*AC*=1) is earned only if the subject is able to reach the target point in a single stroke or by following a rectilinear trajectory. As a final remark, we would like to highlight that the partial scores underline the presence of an active and consistent response to the haptic stimuli based on the matching between the direction followed during a pulse and the following refractory period, but do not consider its coherence in relation to the target position.

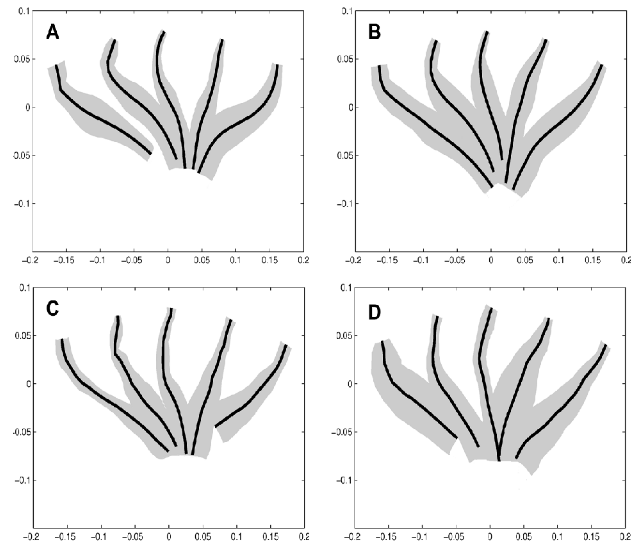

Fig. 3. Mean trajectories (with the corresponding standard deviations) of subject S1 for the 4 different conditions: **VC** (panel A); **C** (panel B); **VP** (panel C); **P** (panel D). The measurement unit on both axes is m.

III. Results

Fig. 3 shows the mean trajectories and the corresponding standard deviation for subject 1, across the different target points in all four experimental conditions. Overall, also the other subjects exhibit a similar behavior. The first qualitative result is that, in comparison with what was achieved with continuous assistance [14], the new pulsed assistance works, in the sense that it is well tolerated by the subjects and allows completing the task even with a much lower level of assistance.

In Fig. 4 the velocity profiles in the two haptic conditions **C** and **P** are depicted (blue lines). It can be noted that in case of pulsed assistance, passive artifacts due to impulse dynamics are superposed on the voluntary component and no immediate clear distinction can be made between intentional and non-volitional contributes.

The statistical analysis of the four performance indicators found that they are not distributed normally (Lilliefors test), suggesting that performance patterns differ with target location. In particular, targets that require extension of the elbow are always more challenging. For this reason, median values over the targets of a same movement set were used for subsequent analysis.

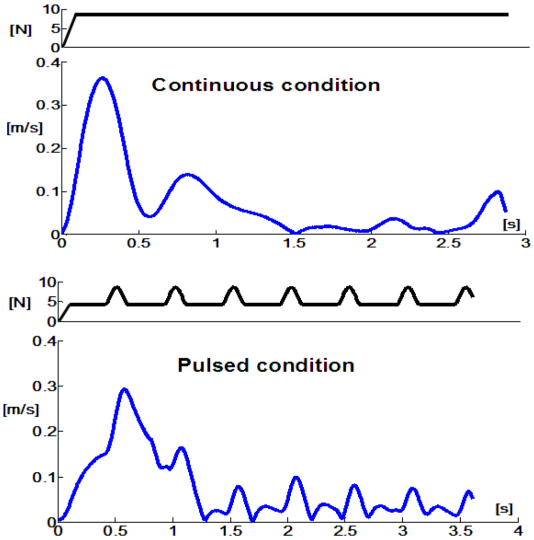

Fig. 4. Force [N] (black curve) and velocity profile [m/s] (blue curve) in condition ***C*** (top panel) and ***P*** (bottom panel) for subject S1.

In order to evaluate if any beneficial effect of a single-session training was present, we compared the percentage of variation of each performance indicator in the last movement set with respect to the first one. A performance variation was considered relevant if exceeding 5%. Table II reports the result. Colored squares represent performance improvement ranking:

1. Red squares represent a positive trend in all the 5 movement directions. This covers the case in which the subject reached and kept the maximum performance for a certain target throughout the session;
2. Orange squares represent a positive trend in 4 directions out of 5;
3. Grey squares indicate conditions when more than 1 direction fell behind the threshold. We report that in all but 3 cases (i.e. S1-VP for mean speed and end-point error, and S2-C for end-point error alone) no more than 2 directions displayed a non-significant performance improvement;
4. Blank regions account for maximum level of performance in all trials of the movement set (0 variation).

TABLE II. INTRASESSION VARIABILITY

SUBJ	MEAN SPEED				END POINT ERROR				T-RATIO				AC	
	VC	C	VP	P	VC	C	VP	P	VC	C	VP	P	VP	P
S1	Grey	Orange	Grey	Grey	Grey	Orange	Grey	Orange	Grey	Red	Grey	Orange	Grey	Orange
S2	Orange	Red	Red	Orange	Orange	Grey		Orange	Orange	Orange	Orange	Grey	Orange	Grey
S3	Orange	Orange	Grey	Red	Red	Orange	Red	Red	Red	Orange	Red	Red	Grey	Red
S4	Orange	Grey	Red	Orange	Red	Orange		Grey	Red	Orange	Red	Grey	Red	Grey
S5	Red	Red	Red	Red	Red	Red	Orange	Red	Red	Red	Red	Red	Red	Red

Subjects (SUBJ); improvement ranking between first and last movement set in all 4 testing conditions.

We report that in all but 3 cases (i.e. S1-**VP** for mean speed and end-point error, and S2-***C*** for end-point error alone) no more than 2 directions displayed a non-significant performance improvement. Fig. 5 compares median and relative interquartile range values of the performance indicators across movement sets for S5. In accordance with the former result, all the three indicators considered (*mean speed, endpoint error*, and *T-ratio*) show a substantial improvement of the median value in the last trial compared to the first one of the test session. The data highlight that subject S5 in condition ***C*** exhibits the highest performance variation. We think that this improvement is hardly imputable to subject's voluntary control, but is probably due to the remarkable reduction in the arm stiffness (see F_A in Table I) not balanced by a reduction of the assistance level.

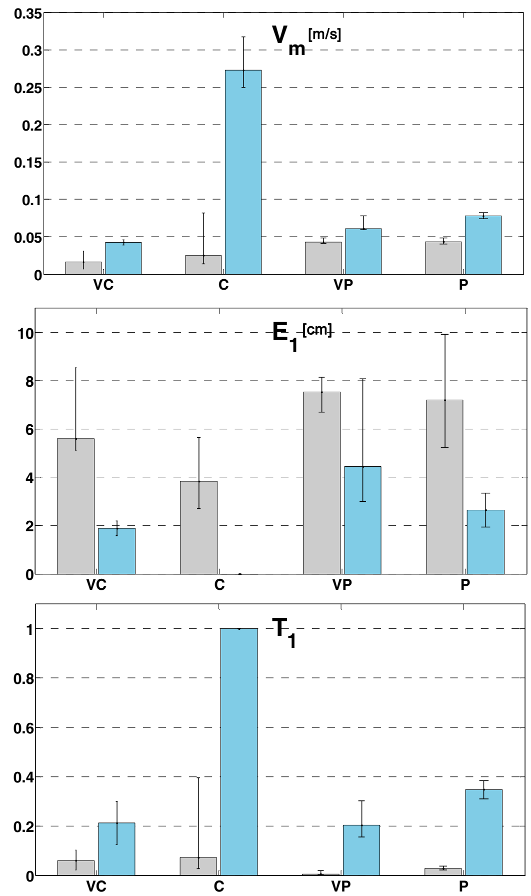

Fig. 5. Median values with the corresponding interquartile range of subject S5 during the first (grey bars) and last (light blue bars) target set for the four training conditions: ***VC***, ***C***, ***VP***, ***P***. Top panel: Mean Speed [m/s]. Center panel: Endpoint Error after first sub-movement [cm]. Bottom panel: T-ratio [a-dimensional].

A further analysis has been conducted to test for differences across conditions both in the first and last trial. Table III summarizes the results of the Friedman ANOVA test (significant differences are marked in red) on *mean speed*, *endpoint error* and *T-ratio* indicators, respectively. Conditions that differ significantly according to the average rank post hoc test are also shown. It can be noted that no significant difference was found neither between the two vision conditions (***VC***, ***VP***) nor between pulsed or continuous haptic feedback (***C***, ***P***). If a difference among vision and no-vision condition is present in the beginning trial, it appears that it is less pronounced or vanishes in the conclusive one (see S2, S3, S4, in particular). This means that if a subject initially relies more on vision, at the end of the session the reliance on proprioception is considerably increased.

TABLE III. INTER-CONDITION VARIABILITY

SUBJ	TR	MEAN SPEED		END POINT ERROR		T-RATIO	
		p	rank>2.1541	p	rank>2.1541	p	rank>2.1541
S1	first	0.1176		0.2394		0.2658	
	last	0.0018	[VP,C]	0.1228		0.0072	[VP,C]
S2	first	0.0070	[VC,P]	0.0035	[VP,P]	0.0041	[VP,C];[VP,P]
	last	0.0263	[VC,P]	0.0099	-	0.0347	-
S3	first	0.0070	[VP,C]	0.0035	[VP,P]	0.0040	[VC,P];[VP,P]
	last	0.1176		0.2298		0.2518	
S4	first	0.0503		0.2123		0.2472	
	last	0.0694		0.0318	-	0.0627	
S5	first	0.0406	-	0.2658		0.0503	
	last	0.0070	[VP,C]	0.0048	[VP,C]	0.0046	[VC,P]

Subjects (SUBJ); trial (TR); Friedman ANOVA p-values and significantly different condition pairs according to post-hoc analysis between conditions in first and last trial with p = 0.05. Significant differences are marked in red.

This observation is supported by the parallel positive variation of the *AC* index reported in Table IV. Only S2 shows a significant reduction of the index that also might explain the reduction in **P** performance of Table II.

TABLE IV. ACTIVE CONTRIBUTION INDEX

SUBJ	TC	ACTIVE CONTRIBUTION			
		first		last	
S1	VP	0.5653	[0.1031 0.3871]	0.5812	[0.1756 0.2092]
	P	0.3763	[0.0416 0.0926]	0.6235	[0.1751 0.0642]
S2	VP	0.8408	[0.0862 0.1592]	0.8646	[0.0726 0.0880]
	P	0.3643	[0.1295 0.0690]	0.3092	[0.0563 0.1680]
S3	VP	0.7568	[0.0799 0.1527]	0.7654	[0.0928 0.0776]
	P	0.4788	[0.0546 0.0271]	0.6435	[0.0284 0.0739]
S4	VP	1.0000	[0.2148 0.0000]	1.0000	[0.0000 0.0000]
	P	0.7012	[0.2962 0.1155]	0.5405	[0.0441 0.1638]
S5	VP	0.0252	[0.0056 0.0107]	0.2372	[0.0404 0.0446]
	P	0.0811	[0.0417 0.0080]	0.3671	[0.1227 0.1227]

Subjects; Training Conditions: **VP** (visual and pulsed haptic feedback and **P**, pulsed haptic feedback); Active Contribution: median values with interquartile range during the first and last target set.

IV. DISCUSSION

There are two elements of novelty in this preliminary study: 1) the use of pulsed force assistance; 2) the automatic evaluation of the minimal assistance level, via the measurement of the holding force. As regards the latter point, a similar type of evaluation was used in [15], but with a different purpose: proprioceptive assessment instead of adaptive selection of assistance. For future developments, we think that it is certainly a good idea to integrate both procedures in the same protocol, thus avoiding the danger of slacking and providing, at the same time, a robust measurement of proprioceptive awareness: this is crucial for recruiting the neural plasticity that is the driving force for functional recovery in stroke patients.

As already suggested in the introduction, we think indeed that in order to promote functional recovery it is necessary to re-establish the causal relationship between intended actions and the corresponding feedback reafference. Pulsed assistance may indeed enhance proprioceptive awareness in a better way than continuous assistance. Both types of assistance are characterized by null internal mechanical impedance of the robot therapist, as perceived by subjects, and this is a fundamental prerequisite for facilitating the emergence of voluntary control because it does help movement, does not force it. Pulsed assistance is likely to be more efficient than continuous assistance in promoting proprioceptive awareness because it activates the phasic channel in addition to the tonic one. One may wonder whether pulsed assistance could have negative effects on performance in spastic subjects, but this does not seem to be the case because also the subjects with rather high values of the ASH index do exhibit improvements in the active contribution index. In addition, our results showed that after the training the holding force is generally steady or decreasing, suggesting that the protocol did not have a negative impact on spasticity. Moreover, pulsed assistance, as demonstrated by this preliminary study, allows to obtain a performance level that is comparable to the continuous paradigm with about half the average assistive force level in a period T. This difference is relevant for the haptic interaction between the robot and the subject, in the sense that the lower the assistance level the higher the chance of promoting intentionality and voluntary control.

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PM and JZ conceived the pulsed assistance paradigm. DDS, VS, JZ carried out the experiments and analyzed the data. LM contributed to the experimental setup and MC to data analysis and interpretation. PG selected the patients and contributed to clinical interpretation. AR performed the clinical tests. DDS, JZ, and PM wrote the manuscript.

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