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Upper and lower body coordination in FES-assisted sit-to-stand transfers in paraplegic subjects - A case study

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Abstract

The objective of the work presented is to improve functional electrical stimulation (FES) assisted sit-to-stand motion in complete paraplegic individuals by restoring coordination between the upper part of the subject's body, under voluntary control, and the lower part of the body, under FES control. The proposed approach is based on the observation of trunk movement during rising motion and a detection algorithm, which triggers a pre-programmed stimulation pattern. We present a pilot study carried out on one T6 paraplegic subject. We validated the ability of the subject to produce repeatable trunk acceleration during sit-to-stand transfers under FES and, the ability of the system to trigger the stimulator at the desired instant in time. We also analyzed the influence of the timing of leg stimulation, relative to the trunk acceleration profile, on upper limb efforts applied during sit-to-stand motion.

Keywords

FES · sit-to-stand · paraplegia

1. Introduction

Standing up is a common daily activity and a prerequisite to standing or walking. This frequently executed task is one of the most biomechanically demanding activities [1]. The biomechanics of standing up, in both able bodied and paraplegic individuals, has been extensively studied and described [2]-[10].The ability to rise from a sitting to a standing position is very important for individuals with paraplegia in order to achieve minimal mobility and has functional and therapeutic benefits related to bone loading, joint extension, cardio-circulatory stimulation, and pressure sore prevention [11]. One method which has been widely investigated is functional electrical stimulation of the lower extremities. The principle of FES is to apply electrical impulses to the sensory-motor system via electrodes which are either placed on the skin, or implanted [12]-[14].

The sit-to-stand method, which is widely used in clinical practice, involves open-loop stimulation of knee extensors activated by hand switches, as proposed by Kralj and Bajd [13] and Guiraud et al. [14]. This technique works adequately in many cases [15]; however, in applying this strategy, stimulation starts without reference to the upper body movement. Hence, the whole-body motion is not optimal and requires a high velocity of the joints and large upper limb forces during the rising motion [10], which may cause both damage of joint tissues and shoulder complications.

A number of closed-loop strategies have been proposed to solve some of those problems. Ewins et al. [16] proposed a proportional-integral-derivative (PID) closed-loop knee controller. The results showed smoother trajectories, but neither reduced upper limb efforts nor lower terminal velocities in the knee were reported. Because of the nonlinear dynamics of muscles and the sit-to-stand motion, a PID controller does not work well [17]. Some other strategies, such as ON/OFF [18] or switching curve [19] controllers were successful in reducing terminal knee velocity during sit-to-stand manoeuvres, however, greater arm force was required when compared with open-loop stimulation. Davoodi and Andrews developed and compared gain scheduling PID and fuzzy logic controllers [20]. These authors used a genetic algorithm (GA) as an optimization approach for tuning the controllers' parameters. Both controllers, when compared with a PID or ON/OFF controller, result in smooth rising manoeuvres and the average electrical stimulation required for successful motion was reduced. They were also able to reduce the arm force but the level of required arm support was still greater than the forces needed in the open-loop FES method. Further, the number of GA trials required during calibration makes it inconvenient for practical use. The same group of authors proposed a fuzzy logic controller based on reinforcement machine learning for controlling FES of lower limbs during rising manoeuvres [15]. Three simulation scenarios were successfully tested: learning to compensate for weak arm forces, learning to minimize arm forces, and learning to minimize the terminal velocity of the knee and arm forces. Although this method appears to be promising, only its theoretical feasibility has been tested.

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Mahboodi and Towhidkhah [17] investigated FES-assisted standing up, using the nonlinear model predictive control approach. This theoretical study shows good tracking behaviour for lower-limb joint angles, but arm efforts during the motion have not been analyzed.
Contrary to these "control-driven" methods, other approaches based on an inverse dynamic model have been proposed [21]- [23]. Here the action of the FES controller is adjusted to the voluntary contribution of the patient, i.e. to the hand forces or body posture. The controllers aim to reduce upper body efforts during standing up. Feedback to the system consisted of joint positions or hand reaction forces, fed into the inverse dynamic model in order to predict the stimulation pulse duration needed for the movement. These strategies require very accurate and realistic models that are often difficult to estimate [11] and their practical real-time application remains limited [12].
In our present work, we propose a "patient-driven" FES method that would coordinate motion of the trunk, which is under voluntary control of the patient, and motion of the lower limbs, which are under FES control. The proposed approach, unlike other "patent-driven" approaches, does not require a complex model of paraplegic standing up and is based on observation of trunk movement during rising motion and a detection algorithm, which triggers a pre-programmed stimulation pattern.
It has been shown that, in able-bodied individuals, trunk orientation and acceleration present low inter and intra-variability and therefore can be a good characteristic signature of the sit-to- stand task [24]. To be efficient, bending forward of the trunk should precede leg movement and last throughout knee extension [24],[25]. In a previous study, we have shown by simulation that such coordination of hip, knee and ankle joints during sit-to-stand motion may reduce arm effort and lower-limb joint torque during the motion [25].
The goal of the present study is to demonstrate an FES closed-loop system for sit-to-stand transfer in paraplegic subjects; this system should automatically trigger leg stimulation at the optimal moment with respect to the trunk motion, in order to decrease arm participation during this motion.

2. Method

2.1. Approach

We have developed a method based on observation of trunk acceleration in the sagittal plane during rising motion, including a detection algorithm that triggers a pre-programmed stimulation pattern. Trunk acceleration was acquired by a single-axis wireless accelerometer positioned on the subject's back. The accelerometer measures acceleration of movement in the horizontal plane with a sampling frequency of 100 Hz. An example of a trunk acceleration signal is shown in Fig. 1.
The detection algorithm consists of an online comparison of the acceleration of the ongoing motion with the reference pattern (a typical pattern characterizing the sit-to-stand transfer for each subject) using Pearson's correlation coefficients. The reference pattern is built from one or more recorded accelerometer signals in the sagittal plane from the same subject. The signal is truncated by defining a time window of the desired length (WL) terminating at the instant when the stimulator should be triggered (WE). Knowing that the rising part of sit-to-stand motion in paraplegic subjects lasts around 500 ms, the length of the window was set to 300 ms. An example of a reference pattern is shown in Fig. 1. The goal of this study was to estimate the impact of the value of WE, i.e. the instant when the stimulator is triggered, on arm participation during sit-to-stand motion of a paraplegic subject. Provided that the reference pattern contained N samples, the correlation between the last N samples of the ongoing signal and the N reference samples was computed using the following equation:

$$C(k) = \frac{1}{N}\sum_{n=1}^{N}\frac{(x(k-N+n)-\bar{x})(y(n)-\bar{y})}{\sqrt{\sigma(x)^2\sigma(y)^2}} \qquad (1)$$

For k⩾N. Where x is the measured signal and y the reference pattern, σ(x), σ(y) are standard deviations of the x and y signal, and $\bar{x}$, $\bar{y}$ are mean values of the x and y signal, respectively. N is the number of samples of a reference pattern; in this study N was set to 30 (0.3 x 100).
When the movement begins, the correlation coefficient starts to change. As the measured signal approaches the point from which the reference was defined, the correlation coefficient increases. Its maximum value should be close to 1 if the ongoing acceleration signal matches the reference pattern. When the coefficient reaches the threshold, the motion is recognized as the sit-to-stand pattern and a command signal for beginning the stimulation is sent to the pre-programmed stimulator. The threshold value for this study was set to 0.85. As part of this strategy, the subject is instructed to project his trunk forward before sit off begins in order to use trunk inertia during the motion [?].
To summarized, once the trunk acceleration signal in sagittal plane has been recorded and reference pattern has been computed, the protocol for our algorithm is described below:

1. Acquire trunk acceleration signal in the sagittal plane.
2. Compute correlation coefficient, C, for the given reference pattern, for last N samplings of a measured signal.
3. Compare C with the threshold value.
4. If C is lower than threshold value go back to 1. If C is bigger than threshold value go to 5.
5. Command signal sent to stimulator.

The ability of the algorithm to detect sit-to-stand motion and trigger a stimulator, as well as differentiating between other similar motions, such as grasping, has been successfully tested in able-bodied subjects [26],[27]. Here, we present a pilot study involving just one paraplegic subject in order to evaluate the feasibility of a new approach to FES assisted sit-to-stand. Three questions were addressed: 1) Is a T6 complete paraplegic subject able to produce repeatable trunk movement? 2) Does the timing of leg stimulation have an impact on upper limb efforts? And finally, 3) is the proposed closed-loop system able to automatically trigger leg stimulation?

2.2. Protocol

Experimental data were collected from one T6 complete paraplegic subject. Approval was obtained from the local ethical committee to perform these tests. The subject had experience in FES usage. The subject's characteristics are presented in table 1 . The subject undertook one muscle mapping session, two muscle training sessions and two measurement sessions.

Mapping session

We tested the condition of the following muscles of the subject: quadriceps (vastus medialis and vastus lateralis) and biceps femoris. The subject's muscle strength was calibrated on the MRC (Medical Research Council) scale of 0-5 (grade 0: no movement is observed,

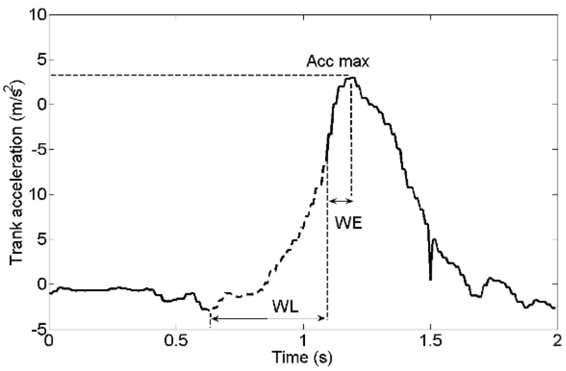

Figure 1. *Trunk acceleration signal in sagittal plane. Acc max is maximum value of acceleration signal, WE and WL are window end and window length respectively. Dotted line represents an example of a reference signal built from this acceleration signal.*

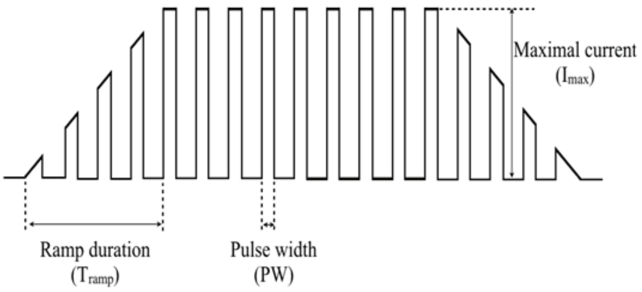

Figure 2. *Shape of stimulation train.*

Table 1. *Subject's caracteristics.*

Age [years]	40
Heigh [m]	1.71
Weigh [kg]	70
Gender	male
Level of the lesion	T6
Post injury period [years]	25
MRC Quadriceps right	3
MRC Quadriceps left	3
MRC Biceps femoris right	3
MRC Biceps femoris left	3
Imax Quadriceps right [mA]	120
Imax Quadriceps left [mA]	130
Imax Biceps femoris right [mA]	120
Imax Biceps femoris left [mA]	120

grade 5: muscle contracts normally against full resistance). Electrodes were positioned on the skin over the motor point of the muscles to be contracted. During the mapping session, we defined the maximum stimulation amplitude (Imax) that would be capable of inducing muscle contraction and ensuring joint locking (table 1). Stimulation parameters were pulse width (PW) of 300 μs and frequency of 30 Hz. These parameters remained approximately the same throughout the study. A PROSTIMTM stimulator was used: a programmable stimulator manufactured by MXM, France. The device provides 8 bipolar channels delivering current controlled biphasic stimulation pulses, with a capacitive secondary pulse. The stimulator is CE marked. Rectangular, self-adhesive, multi-use surface electrodes (50x90mm) were positioned over the muscles.

Training session

The subject undertook two training sessions. During these sessions he became familiarized with the experimental setup. At the beginning of each session, he was sitting on the chair. The quadriceps muscle group and biceps femoris were stimulated, starting with a stimulation amplitude of 30 mA and increasing until Imax was reached. This was repeated until the subject was able to perform sit-to-stand motion and maintain standing posture for a couple of seconds. To ensure smoother muscle contraction, the stimulation train was ramped. The duration of the ramp (Tramp) was set to 300 ms (see Fig. 2). The experimental setup is presented in Fig. 3. Initial and final positions of the subject are presented in Fig. 3a). The paraplegic subject of the experiment is depicted in Fig. 3b).

Measurement session

The subject undertook two measurement sessions within one week. The kinematic data were acquired by a single-axis wireless accelerometer positioned between the shoulder blades of the subject. Technical characteristics of the accelerometer are following: nonlinearity $\pm 2\%$ of full scale; 0g offset accuracy ± 0.04g; maximal value of 0g offset long term accuracy $\pm 1\%$. The accelerometer was mounted on a harness that was fixed at the same position for all the experimental sessions. The wireless system is described in [28],[29]. ATI Industrial Automation's force sensors with six degrees of freedom were mounted on handles on a set of parallel bars in order to record arm efforts. The sampling frequency of the accelerometer and force sensors was 100 Hz. A video camera recorded the trial. A video projector was connected to the camera and positioned in front of the subject, who could therefore see his profile. The stimulation current amplitude was set to Imax. For the same reason as in training session, the stimulation train was ramped with a ramp time of 300 ms (Fig. 2). The same muscles were stimulated as in training session. At the beginning, the subject was sitting on the chair with arms resting on the handles (Fig. 3). The positions of the handles were adjusted according to the subject's height and preference. The subject was instructed to keep his trunk straight and vertical, and to perform the rising motion, following the experimenter's signal, by propelling his trunk forward before the sit off phase commenced. At the beginning of each measurement session, the subject performed a sit-to-stand motion using only the arm support. This was so that the hand forces recorded during these trials could be compared with those recorded during FES-assisted motions. For the second rising motion, stimulation was triggered manually by the experimenter. The trunk acceleration measured in this trial was used to build the reference signal. Subsequent trials were performed using our detection algorithm. The number of trials depended on the subject's ability (fatigue and muscle response to FES) to repeat the sit-to-stand motion. As discussed above, the goal of this study was to find the optimal value of WE. In a previous publication, it has been shown that in able bodied people motion of the legs starts at around maximum trunk acceleration [?]. Considering that our stimulation pattern includes a 300 ms rising ramp and that there is a time delay between sending the command signal

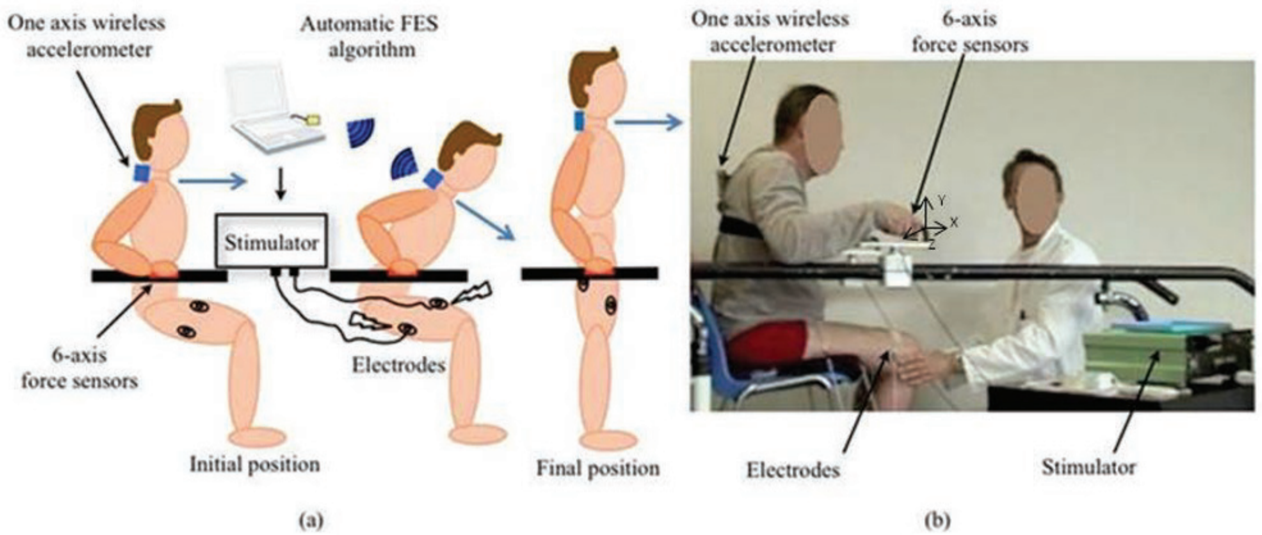

Figure 3. Initial and final positions of the patient and protocol description.

to the stimulator and muscle contraction, the motion most similar to that of able bodied subjects would be one in which stimulation starts around 400 ms before maximum acceleration of the trunk. Reduced hand forces are expected at these values. In order to find the optimal timing for the start of the stimulation, we tested our algorithm for different values of WE: 400 ms, 300 ms, 200 ms and 0 ms, (i.e. the delay before trunk acceleration reaches its maximum value). Each measurement continued for a few seconds after standing posture was reached.

3. Results

As stated above, the goal of this study was to answer the following questions:

1. Is a T6 complete paraplegic subject able to control his trunk motion sufficiently to perform reproducible trunk acceleration profiles?
2. Is there an influence of the timing of leg stimulation relative to trunk acceleration on upper limb efforts?
3. Is the proposed closed-loop system able to automatically trigger leg stimulation?

3.1. Trunk motion reproduction

Pearson's correlation coefficients, between acceleration signals from the first (manually triggered) and subsequent FES-assisted sit-to-stand trials, are calculated and presented in Table 2. It can be concluded that, despite the very low amount of training of this subject, the correlation between signals is relatively high (0.77 +/- 0.14). From twelve trials,

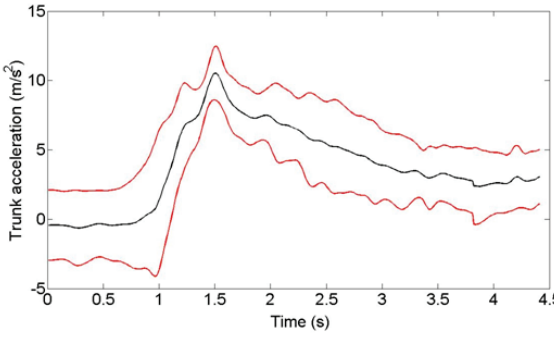

Figure 4. Rising phase of all sit-to-stand trials: mean value and standard deviation.

ten exhibited a correlation coefficient with the chosen reference pattern above 0.73. In Fig. 4, mean and standard deviation values of trunk acceleration for the rising phase of each trial are also shown.

3.2. Impact of stimulation time on upper limbs effort

Using the forces recorded under patient's hands, sum of the left and right, resultant hand forces were calculated using the following equation:

$$F = F_{left} + F_{right} = \sqrt{Fx_{left}^2 + Fy_{left}^2 + Fz_{left}^2} + \sqrt{Fx_{right}^2 + Fy_{right}^2 + Fz_{right}^2} \quad (2)$$

Table 2. Description of the sit-to-stand trials reaslized during both measurement sessions.

Measurement section	Number of trials	Type of trials	Correlation coeficient
	1	No stimulation	—
F	2	Manual	Reference signal
I	3	Automatic	0.868
R	4	Automatic	0.789
S	5	Automatic	0.864
T	6	Automatic	0.939
	7	Automatic	0.875
	1	No stimulation	—
S	2	Manual	Reference signal
E	3	Automatic	0.427
C	4	Automatic	0.738
O	5	Automatic	0.784
N	6	Automatic	0.796
D	7	Automatic	0.796
	8	Automatic	0.564
	9	Automatic	0.793

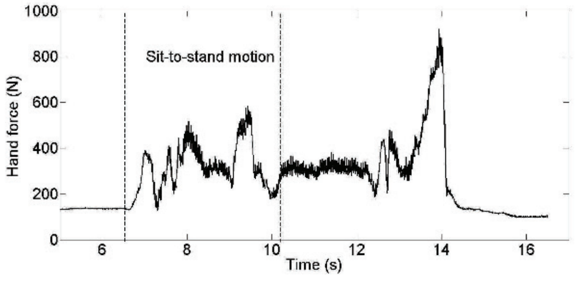

Figure 5. Example of sum of the left and right resulatant hand forces for one sit-to-stand trial recorded during the experiments.

Mean (M) and maximum (Max) values of F were calculated for each rising phase of each sit-to-stand trial. In Fig. 5, an example of the calculated hand forces for one sit-to-stand trial is presented.
Fig. 6 a) and Fig. 6 b) show, respectively, the M and Max values for each rising phase of each sit-to-stand trial. The first two bars represent trials during which the subject stood up using only the arm support. The other bars represent sit-to-stand trials performed using FES. Numbers on the x axis represent the stimulation time in seconds with respect to the maximum trunk acceleration. Negative values indicate that the stimulator was triggered before Accmax. Similarly, positive values mean that stimulation started after Accmax, i.e. after sit off. Observing these figures, we see that there the timing of stimulation application (WE) has an influence on the measured upper limb support forces. The best compromise between lowest mean and maximum value of the hand forces seems to be achieved for the trials where the command signal was sent to the stimulator at 0.42 s, 0.33 s, 0.26 s and 0.24 s before maximum trunk acceleration. The subject also reported a better feeling during those trials. As depicted in Fig. 6, if the stimulation trigger appears after the maximum value of trunk acceleration, i.e. after the sit-off phase, the stimulation can be seen as an external perturbation creating additional hand forces. Here, it should also be mentioned that in the case when stimulation started 0.19 s before Accmax, the subject had strong spastic contractions of the leg muscles; hence ,the forces recorded in that case should probably not be taken into consideration.

Table 3. Detection algorithm perfprmace.

Measurement section	Type of trials	WE desired [s]	WE "real" [s]	Detection error [s]
F	Automatic	0	-1.23	1.23
I	Automatic	0	-0.15	0.15
R	Automatic	-0.3	-0.19	0.11
S	Automatic	-0.3	-0.42	0.12
T	Automatic	-0.3	-0.24	0.06
S	Automatic	-0.4	*	—
E	Automatic	-0.4	-0.33	0.07
C	Automatic	-0.4	+0.76	1.16
O	Automatic	-0.4	-0.85	0.45
N	Automatic	-0.3	-0.26	0.04
D	Automatic	-0.3	-0.03	0.27
	Automatic	-0.2	-0.42	0.22

3.3. Detection of sit-to-stand motion

Table 3 shows the desired stimulation time (WEdesired), the stimulation time actually achieved (WE "real") and the difference between them (Detection error) for all sit-to-stand trials performed in so-called "automatic mode" during the two measurement sessions. Again, negative values indicate that the stimulator was triggered before Accmax and positive values mean that stimulation started after sit off had begun. The values for the mean and standard deviation of the detection error are 0.35 +/- 0.43 s. The importance of the sensor location for the sit-to-stand motion recognition is highlighted by table 3. For the first automatic trial of the second measurement session, the reference signal which was used to detect the motion was the one recorded and used during the first measurement session. In that case, correlation between signals was low, due to the slightly different accelerometer position, and detection of the motion failed. We then used the acceleration recorded in the manual trial as the reference signal. In fact, it can be seen that when the reference signal was correctly constructed, as described in section 2.1, and had the same sensor location, the stimulator was triggered automatically by our system. It is also important to mention that the subject used the system during two sessions only, for a total of 12 trials. We believe that, in the case of this specific subject at least, with suitable training the subject would improve his trunk motion performance, so that acceleration signals would be more repeatable and the overall ability of the system to trigger the stimulator at the desired instant would increase.

4. Discussion and Conclusion

The objective of this study was to demonstrate the approach that we have developed relating to functional rehabilitation techniques of lower extremities. Unlike "control-driven" approaches, our system takes into account the contribution from trunk inertia to sit-to-stand motion and does not require a predefined reference input for lower limb trajectories, which would need to be adjusted to each subject. In addition, due to the task dynamics and the nonlinearities of the postural system, the development of a closed-loop control law remains challenging. In our approach, generation of the motion in paralyzed limbs is driven by the

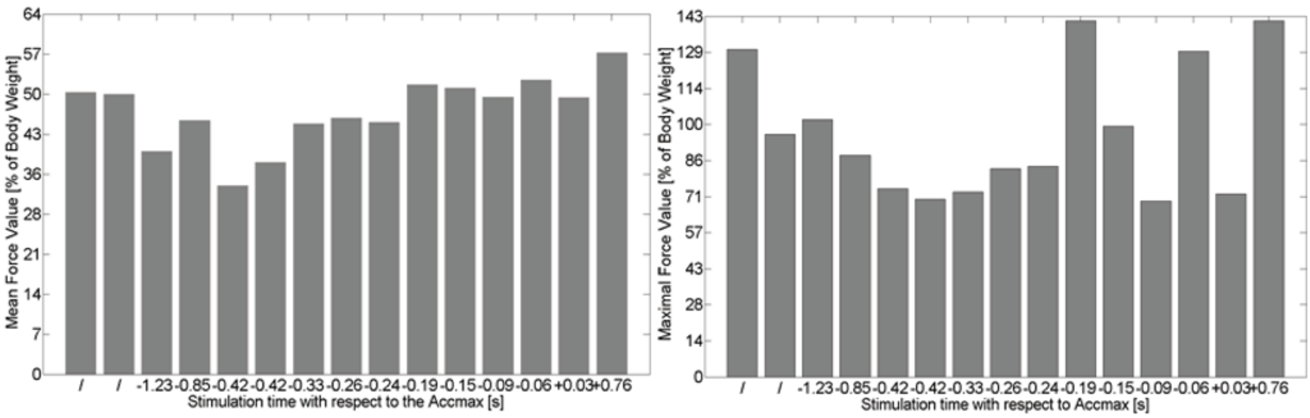

Figure 6. 6a) Mean value of sum of the left and right resultant hand forces measured during experiments calculated over each sit-to-stand trial. 6b) Maximal value of sum of the left and right resultant hand forces measured during experiments calculated over each sit-to-stand trial. The first two bars in both graphs represent trials during which the subject stood up using only the arm support.

patient's voluntary trunk motion. Compared with other "patient-driven" approaches, our system does not require complex modelling. Finally, only one accelerometer, which is easy to use in clinical applications, is required. From the results we presented here we can conclude, for this specific paraplegic subject, that he is able to produce repeatable trunk motion and that, in the cases where the acceleration and reference signal are similar, our algorithm is able to recognize sit-to-stand motion and to properly trigger leg stimulation at the desired instant. Also, we have shown that there is an influence of stimulation timing on applied hand forces during the motion. The best results were achieved for trials in which motion was similar to that of able bodied subjects in terms of trunk motion and the beginning of the leg motion with respect to the trunk acceleration signal. We believe that with extra training, it would be possible to improve the performances of the paraplegic subject and, at the same time, to improve performance of the system with respect to stimulation. The results of this pilot study are encouraging, though further subject tests are necessary. In the future we plan to include several complete paraplegic subjects in our study in order that more generally applicable conclusions can be drawn. Moreover, we plan to improve robustness of the system by improving sensor positioning so that the reference signal, once built, can be used for various sessions. The same approach will also be tested in the near future on transfers from wheel-chairs to car seats or beds, by complete and incomplete paraplegic subjects. The ability to achieve these transfers with minimal participation of the upper limbs would greatly improve daily life for paraplegic individuals and would help to preserve long-term shoulder integrity.

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