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Commanding a robotic wheelchair with a high-frequency steady-state visual evoked potential based brain–computer interface

Pablo F. Diez [a,b,*], Sandra M. Torres Müller [c], Vicente A. Mut [b], Eric Laciar [a], Enrique Avila [b], Teodiano Freire Bastos-Filho [d], Mário Sarcinelli-Filho [d]

[a] Gabinete de Tecnología Médica (GATEME), Universidad Nacional de San Juan (UNSJ), Argentina
[b] Instituto de Automática (INAUT), Universidad Nacional de San Juan (UNSJ), Argentina
[c] Computer Engineering Department, Federal University of Espirito Santo (UFES), Sao Mateus-ES, Brazil
[d] Electrical Engineering Department, Federal University of Espirito Santo (UFES), Vitoria-ES, Brazil

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ABSTRACT

This work presents a brain–computer interface (BCI) used to operate a robotic wheelchair. The experiments were performed on 15 subjects (13 of them healthy). The BCI is based on steady-state visual-evoked potentials (SSVEP) and the stimuli flickering are performed at high frequency (37, 38, 39 and 40 Hz). This high frequency stimulation scheme can reduce or even eliminate visual fatigue, allowing the user to achieve a stable performance for long term BCI operation. The BCI system uses power-spectral density analysis associated to three bipolar electroencephalographic channels. As the results show, 2 subjects were reported as SSVEP-BCI illiterates (not able to use the BCI), and, consequently, 13 subjects (12 of them healthy) could navigate the wheelchair in a room with obstacles arranged in four distinct configurations. Volunteers expressed neither discomfort nor fatigue due to flickering stimulation. A transmission rate of up to 72.5 bits/min was obtained, with an average of 44.6 bits/min in four trials. These results show that people could effectively navigate a robotic wheelchair using a SSVEP-based BCI with high frequency flickering stimulation.

1. Introduction

A brain–computer interface (BCI) is a system that allows an individual to command a device, such as a wheelchair [1], a web navigator [2], a speller [3] or writing with a robot [4] using only brain signals and without any muscle movements [5]. Generally, this brain activity is detected in the electroencephalographic (EEG) signals, which are registered with electrodes placed directly on the surface of the scalp.

It is well known that a BCI system can control a mobile robot and/or a robotic wheelchair using BCI paradigms, such as mental tasks [6–8], motor imagery [9,10] or P300 potentials [11–14]. Specifically, Leeb et al. were the first to show that a quadriplegic subject could control a wheelchair using motor imagery inside an avatar-populated virtual street [10]. Millán et al. used the imagination of a left hand movement, word associations, arithmetic operations and relaxation. Those mental commands were associated with the three steering functions of the wheelchair: turn left, turn right and move forward [8]. A P300 BCI was proposed by Rebsamen et al. which allows the user to select a destination item on a menu and then drive the wheelchair to the corresponding target following a pre-defined path [14]. Another BCI driving a wheelchair using P300 and navigation algorithm based on laser sensor information was proposed by Iturrate et al. [13]. On the other hand, a wheelchair was commanded using electrooculography (EOG) [15], surface electromyography (sEMG) [16] and both EOG/sEMG [17–19]. Furthermore, a BCI can be inserted into hybrid or multimodal systems, using EEG and/or other biomedical signals for commanding a wheelchair [20,21].

A BCI must classify the ongoing brain activity constantly for controlling a wheelchair, i.e., the BCI should be able to detect if the user intentionally decides to perform some specific tasks or does not generate any command. This is called a self-paced or asynchronous BCI [22].

A BCI to control robotic systems can also be based on steady-state visual-evoked potentials (SSVEP). A SSVEP is a resonance phenomenon arising mainly in the visual cortex when a person is focusing the visual attention on a light source flickering with a

* Corresponding author at: Universidad Nacional de San Juan, Av. Libertador 1109 (oeste) Capital, San Juan (J5400ARL), Argentina. Tel.: +54 0264 4211700x359.
E-mail addresses: pdiez@gateme.unsj.edu.ar (P.F. Diez), sandramuller@ceunes.ufes.br (S.M. Torres Müller), vmut@inaut.unsj.edu.ar (V.A. Mut), laciar@gateme.unsj.edu.ar (E. Laciar), eavila@inaut.unsj.edu.ar (E. Avila), teodiano@ele.ufes.br (T.F. Bastos-Filho), mario.sarcinelli@ele.ufes.br (M. Sarcinelli-Filho).

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frequency above 4 Hz [23]. In EEG signals the SSVEPs are periodic signals, with a stationary distinct spectrum showing characteristic SSVEP peaks, stable over time [24].

Visual evoked based BCI systems (such as SSVEP or P300) can be used by most users with relatively little training, in contrast to other BCI systems that do not rely on SSVEP or P300 signals [25–27]. Moreover, SSVEP-BCI systems tend to outperform other BCIs in terms of information transfer rates [27]. SSVEP signals are triggered by external stimuli, which are more robust and easier to control than internally generated signals (such as sensorimotor rhythms) [24].

According to [23], a SSVEP can be elicited in three ranges of frequency, namely, low (4–12 Hz), medium (12–30 Hz) and high frequency range (>30 Hz). In general, SSVEP in low frequency range has larger amplitude responses than in the medium and high ranges. Moreover, the larger the amplitude of the SSVEP, the easier its detection is. The weakest SSVEP is found in the high frequency range. This is perhaps the main reason why high-frequency SSVEP-based BCIs are not broadly used. However, low and medium frequency stimulation interfere with the alpha rhythm, consequently, near the spontaneous EEG alpha peak the SSVEP detection is difficult [24]. Moreover, low and medium frequency stimulation could cause epileptic seizures as well [28]. On the other hand, high frequency stimulation has the advantage of greatly decreasing visual fatigue, caused by flickering, so that these stimuli can be used in order to develop a more comfortable SSVEP-based BCI [29–33]. However, this statement was not reliably confirmed to present day.

There are few studies about wheelchair navigation using a SSVEP-based BCI. For example, three non-impaired subjects commanded a simulated wheelchair using a finite state machine controlled by a SSVEP-based BCI [34]. In the research presented by Mandel et al., eight healthy subjects (non-handicapped) commanded a wheelchair using a SSVEP-based BCI, requiring ten minutes of preparation [35]. Finally, in some previous work developed by one of co-authors of this paper, a robotic wheelchair was commanded by six healthy subjects with SSVEP extracted with a statistic test and classified with a rule-based classifier [1]. However, none of these three projects used high-frequency stimulation.

Recently, two high frequency SSVEP-based BCIs were presented in which none of the volunteers controlled a wheelchair [33,36]. In the first one, a miniature robot was moved on fixed positions in a maze; whereas in the second one an object was moved on the screen through a maze. Nevertheless, commanding a wheelchair remains to be quite a different task, since the user is seated on the mobile object.

In this paper, the operation of a real robotic wheelchair using an asynchronous high-frequency SSVEP-based BCI is reported. The implemented system was evaluated with both healthy and people with disabilities, commanding the wheelchair through different scenarios.

2. Materials

The robotic wheelchair used in this research was developed in the Federal University of Espirito Santo (UFES), Brazil, and is shown in Fig. 1. It was developed starting from a modified commercial motorized wheelchair. A dual channel digital motor controller, model AX3500, from RoboteQ Inc. (www.roboteq.com), is responsible for controlling the velocity of the two driven wheels. A compact and low-power mini-computer (EPIA, with a 1 GHz processor and 1GB of RAM) performs all functions, such as EEG processing, communication with the AX3500 board, algorithms execution, etc.

A 12″ tactile LCD screen was attached to the robotic wheelchair structure and the stimuli system was fitted on the sides of that screen. The flickering frequencies were 37 Hz (top), 38 Hz (right), 39 Hz (bottom) and 40 Hz (left). The stimuli temporization is precisely controlled using a FPGA Xilinx 3E on a Nexys board (Digilent Inc.). Each stimulus is composed of a 2.5 cm × 2.5 cm square box, illuminated by high efficiency green Light-Emitting Diodes (LEDs). The stimulation frequencies were chosen according to resonance phenomena on 40 Hz [37] and 1 Hz separation between stimuli was wider than minimal requirement of 0.2 Hz [38].[1]

The wheelchair velocities were empirically chosen in order to generate safe movements that would not endanger the subject seated on it (high speeds were avoided). On the other hand, low speeds could lead subjects to lose their motivation in using the BCI, since they could consider it slow, boring or tedious. Having these two points in mind, the wheelchair velocities were held constant at 20 cm/s for advancing and 14°/s for turning.

Three bipolar EEG channels were recorded (Ch1: O_1-F_3; Ch2: O_Z-F_Z; Ch3: O_2-F_4) using a BrainNet BNT-36 signal acquisition equipment, whose analog filter was set for 3–100 Hz band, and the signals were digitalized at 240 Hz. This configuration (three bipolar EEG channels, filters bandwidth and sampling frequency) was the most similar to the one used in our previous work in high frequency SSVEP [36]. Moreover, bipolar derivations are more robust to noise and have higher performances in SSVEP detections than monopolar ones [42]. The EEG is transferred from the BrainNet BNT-36 signal acquisition equipment to the computer once per second.

3. Experimental trials

Fifteen subjects (24 ± 7 years, 14 m – 1 f) participated of the experiments, thirteen of them being healthy individuals. One subject with disabilities is male and paraplegic, with severe paralysis of upper limbs due to a lesion at the fifth cervical vertebra. Another subject with disabilities is a female, 34 years old, with tetra-paresis due to a lesion at cervical (C4-C5 vertebra) level. No one had previous experience in using a BCI. The experiments were performed according to the rules of the Ethics Committee of the Federal University of Espirito Santo, under registration number CEP-048/08.

Two subjects (the female with disabilities and one healthy subject) were unable to control the wheelchair and did not perform the proposed trials. They were excluded from evaluations and were referred to as SSVEP-BCI illiterates. BCI illiteracy is a well-documented phenomenon and refers to a non-negligible portion of subjects that are unable to achieve effective control of a BCI [33,39].

Firstly, a baseline EEG signal was acquired, for equalization purposes, due to the low power EEG spectrum associated with higher frequencies [36] (as an example, a SSVEP at 38 Hz has higher power than another SSVEP at 40 Hz). This baseline was recorded during 60 s, while the subject was gazing the center of the screen, and without focusing on any stimuli (even though the stimuli were on).

Subjects performed a "training session" before the wheelchair navigation. They were seated on the wheelchair, however the BCI did not convey any control signal to the wheelchair (the motors were kept off, for safety) and, hence, subjects were not concerned about wheelchair movements. This was appropriate, since it was the first time that such subjects used a BCI system. This trial was performed in order to evaluate the detection rate of SSVEP of BCI. Specifically, the training session consisted of gazing at each stimulus during a lapse of 20 s (all stimuli were flickering), beginning with the top stimulus and switching to the next one in clockwise direction (i.e., top, right, bottom, left). Thus, the trial was finished

[1] In the current work, the Power Spectral Density (PSD) was computed using the periodogram function with a 2s (N = 480 samples) long rectangular window, as it is stated in Section 4.1. Due to EEG signals are sampled at f_S = 240 Hz, the frequency resolution $\Delta f = f_S/N = 240/480 = 0.5$ Hz is lower than 1 Hz separation used between the stimulation frequencies.

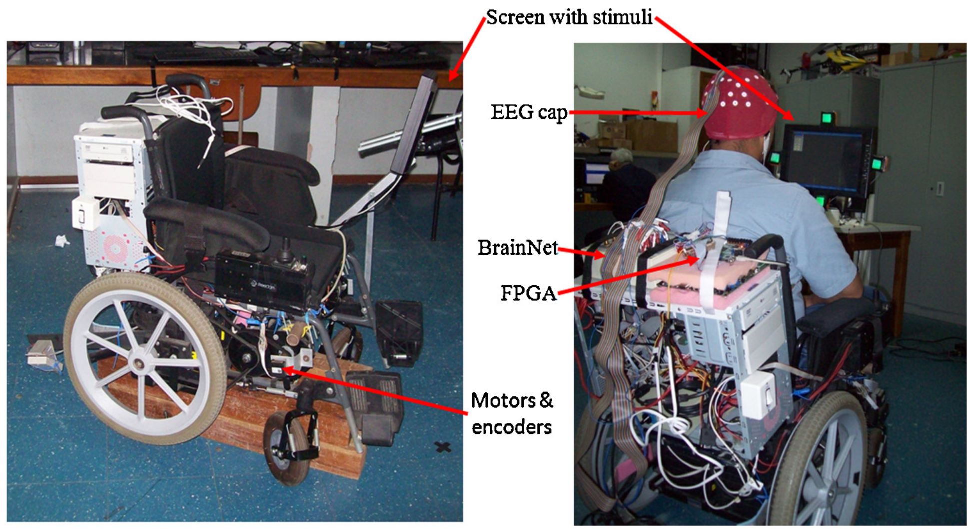

Fig. 1. Robotic wheelchair used in the experiments (left) and a subject seated on it using the signal acquisition equipment and the stimulation system (right).

in 80 s. On-line feedback was presented to the subjects, by means of an arrow on the screen. The EEG signal was processed with the method described in Section 4.1.

Afterwards, the subjects were able to command the robotic wheelchair. Four scenarios were proposed for the system evaluation, as depicted in Fig. 2. The tasks consisted of commanding the wheelchair from the start zone toward the goal zone, which is close to the doorway. Fig. 2 also presents the desired paths to be followed by the wheelchair. The room dimensions are 8.75 m long by 7.07 m wide. This room is located on the first floor. Due to the lack of accessibility, the trials with subjects with disabilities were performed in another room with similar dimensions and the same obstacle configuration, now located in the ground floor. The subjects defined their paths to get to the goal zone.

Trial 1 consisted in commanding the wheelchair from the start zone to the doorway of the room without any obstacle in its trajectory, as shown in Fig. 2. During this trial the subject had to become familiarized with the BCI–wheelchair system. In Trial 2, the user could choose between two possible paths to evade an obstacle in the middle of the room, in order to reach the goal zone.

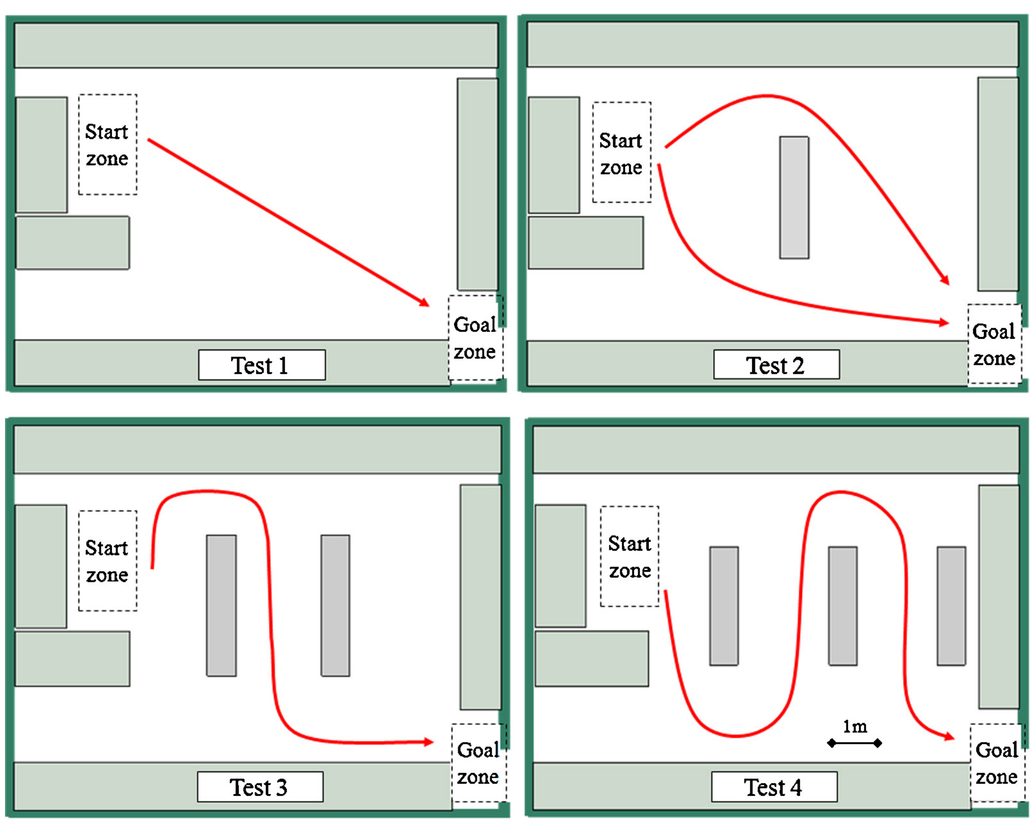

Fig. 2. Four proposed scenarios used in the experiments. Light green rectangles represent desks and cabinets. Gray rectangles are the different obstacles between the start zone and the goal zone. The red line indicates the suggested path. In Trial 2, there are two possible paths. (For interpretation of the references to color in the artwork, the reader is referred to the web version of the article.)

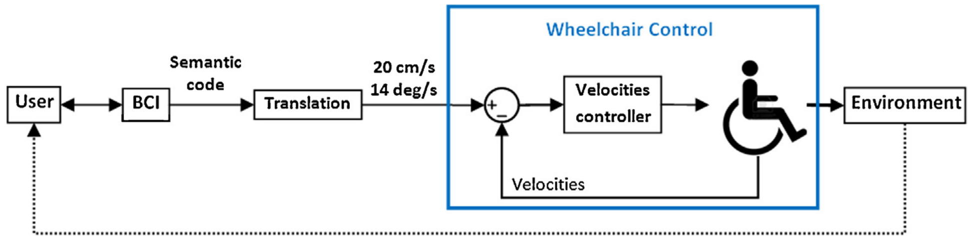

Fig. 3. Scheme of BCI–wheelchair interaction. When the BCI detects that the user is gazing at stimulus, a command is generated as a semantic code ('advance', 'right', 'stop', 'left'). This code is translated into velocities (20 cm/s advance, 14°/s clockwise turn, 0 cm/s stop, 14°/s counterclockwise turn, respectively), which are the reference velocities of the wheelchair control system. Finally, the user looks at the environment and decides (or not) to send a new command to the wheelchair through the BCI.

Trial 3 is more complex: the subjects had to be able to command the wheelchair through a corridor and avoid colliding with the desks at the entrance and exit of the corridor. To do that, they had to perform three small turns of 90° in order to get to the goal zone. The corridor is 1.5 m wide and the wheelchair is 0.6 m wide, which leaves only 0.45 m on each side of the wheelchair along the corridor. Consequently, the wheelchair could collide with the wall of the corridor in case of erroneous command detection by the BCI. For safety precautions, all the obstacles and the corridor were built with common chairs, aimed at simulating the corridor in such a way that the subjects would not be hurt in case of a collision.

The trials depicted in Fig. 2 are increasing in their level of difficulty from Trial 1 to Trial 4, i.e., once a subject accomplished Trial 1, he/she was invited to perform the Trial 2 and so on.

Originally there were only three trials (Trials 1–3); however, the experiments demonstrated that the subjects could execute the proposed trials successfully. Therefore, a new trial was proposed during the experimental sessions, in order to evaluate the developed system in a more complex environment, originating the fourth trial. Then, Trial 4 was only executed by the last four subjects that had participated in the experimental sessions.

4. Methods

This section presents the EEG signal processing method, followed by the wheelchair control, and the evaluation procedures used to evaluate the performance of the subjects and the proposed system.

4.1. EEG processing method

The EEG signal processing method was developed by our group in previous work [36]. Firstly, the EEG signal is digitally filtered with a 6-order Butterworth band-pass filter with cut-off frequencies set at 32 and 45 Hz. Then, the power spectral density was estimated using the periodogram function with a 2 s long rectangular window. This window was moved in 0.25 s steps. Then, the normalized power, $P(f_i)$, at each stimulus frequency f_i (37, 38, 39 and 40 Hz) is computed as:

$$P(f_i) = \sum_{\mathrm{ch}=1}^{M} \frac{\hat{S}_{\mathrm{ch}}(f_i \pm 0.25\mathrm{Hz})}{\widehat{BL}_{\mathrm{ch}}(f_i \pm 0.25\mathrm{Hz})} / M \quad (1)$$

where M is the total number of channels ($M=3$); ch is the channel number, varying from 1 to $M=3$; $\hat{s}_{\mathrm{ch}}(f)$ is the periodogram of the EEG signal and $\widehat{BL}_{\mathrm{ch}}(f)$ is the periodogram of the baseline EEG signal (computed with a 2 s long rectangular window, with 50% of overlapping). This calculation was performed four times per second, according to the window movement steps (0.25 s). Then, the maximum $P(f_i)$ is extracted as

$$\mathrm{class}(f_i) = \max\left\{P(f_i)\right\} \quad (2)$$

A SSVEP is labeled as one of the four possible classes (top, right, bottom or left) when a class (f_i) is maintained for at least 2 s. In other words, when eight consecutive class (f_i) corresponding to the same f_i, are detected. Finally, whenever this condition is not satisfied, the EEG segment is classified as belonging to an undefined class.

Once a class is detected, the corresponding command is sent to the wheelchair once per second, keeping the corresponding movement of the wheelchair as long as this class is detected. Upon changing the class, the movement of the wheelchair is changed accordingly.

4.2. Wheelchair command

As discussed in the previous subsection, the developed BCI can detect five classes: top, bottom, left, right and undefined. The first four classes correspond to the four stimuli and the last class is related to the case when none of the stimuli frequencies is detected. The programmed actions are: the wheelchair goes forward if the BCI detects the top stimulus, turns left (or right) if the BCI detects the left (or right) stimulus and stops if the BCI detects the bottom stimulus. Additionally, when an undefined class is detected, the wheelchair stops as well (for safety reasons). Online feedback was provided to the user, consisting of a blue arrow on the screen indicating the detected stimulus. Upon detection of an undefined class, a red circle is shown in the center of the screen.

The intention of the user should be translated into wheelchair movement, as shown in Fig. 3. Therefore, the BCI detects the user intention and generates a semantic code, i.e., 'advance', 'right', 'left' or 'stop', and this code has to be translated into velocities. As mentioned before, such velocities were set to 20 cm/s for advancing and 14°/s for turnings. Then, according to the intention of the user, these reference velocities are the input of the wheelchair velocity control system. Finally, the user seated in the wheelchair closes the loop observing the environment and eventually makes a new decision.

No external sensors were used onboard the wheelchair during the experimental sessions, therefore, high level control algorithms were not implemented in order to avoid obstacles, to travel along a certain path, to travel through a doorway or to accomplish any related task, as was implemented in [9] and [13]. Consequently, the user had to be in complete control of the wheelchair. Hence, the user conveys different commands, using the BCI, which are translated into wheelchair velocities. The wheelchair moves and then the user observes the surrounding environment and makes a new decision about the next movement of the wheelchair, whenever necessary. The BCI sends a command to the wheelchair control system every second, a rate that has shown to be suitable during the execution of the proposed navigation trials. This time constraint is due to BrainNet BNT-36 signal acquisition equipment.

4.3. Evaluation procedures

Evaluation procedures were performed on the information obtained from the experimental sessions. This information consists of the SSVEP power calculated in real time, the commands sent to the wheelchair control system and the path described to get to the goal zone. Each experimental session was video-recorded.

Generally, videos allow engineers to determine the subject intention, i.e., which stimulus is gazed at. Later, by comparing the information obtained from the experimental sessions (SSVEP power and the corresponding detected commands) with the videos recorded during the trial, it is possible to classify the commands as 'Correct' (C), 'Incorrect' (I) or 'Non-Detected' (ND). Correct commands are those which are according to the proposed path and incorrect commands are those not in such accordance. For instance, when the subject should turn to the right, a different command is an incorrect one. Finally, non-detected commands correspond to the undefined classes detected by the BCI. It is difficult to determine when subjects start/finish gazing at a stimulus due to asynchronous mode. Hence, the calculated correct and incorrect values could under/overestimate the true values and they should be analyzed as an estimation of the real ones.

The Information Transfer Rate is a standard measure of communication systems which contemplates accuracy, the number of possible selections, and the time required to make each selection [40]. In this work, the ITR is calculated as [7]

$$\mathrm{ITR} = (1-p_\mathrm{u})\left[\log_2 N + (1-p_\mathrm{e})\log_2(1-p_\mathrm{e}) + p_\mathrm{e}\log_2\left(\frac{p_\mathrm{e}}{N-1}\right)\right] \quad (3)$$

where p_u is the probability of undefined cases, p_e is the probability of incorrect detected cases and N is the number of targets (in our case $N = 4$). The ITR in (3) is expressed in bits/commands. To obtain it in bits/min, it is necessary to multiply the result from (3) by the selection speed of the BCI, expressed in commands/min. As was previously mentioned, the developed BCI executes 1 command per second. Therefore, the selection speed is 60 commands/min.

The ITR is computed based on the p_u and the p_e values. The p_u value is calculated as the rate between the number of non-detected commands and the total number of commands ($p_\mathrm{u} = \mathrm{ND}/(\mathrm{C}+\mathrm{I}+\mathrm{ND})$). A similar criterion was used to calculate p_e, that is, $p_\mathrm{u} = \mathrm{I}/(\mathrm{C}+\mathrm{I}+\mathrm{ND})$. The calculated ITR is an estimation of the capacity of the communication channel among brain and computer when subjects command the wheelchair.

The Positive Predictive Value (PPV) is the proportion of detections that are correct detections. It is calculated as:

$$\mathrm{PPV} = \frac{\mathrm{TP}}{\mathrm{TP}+\mathrm{FP}} \quad (4)$$

where TP is True Positive values (i.e. a SSVEP correctly detected) and FP is False Positive (i.e. a SSVEP incorrectly detected). Sensitivity and Specificity were not calculated since False Negative (FP) and True Negative (TN) values cannot be separately determined because both values are considered as non-detected cases.

At the end of the experiments, each subject answered a questionnaire about his/her own experience with BCI system usage. The questions were as follows:

1) Are you tired?
2) Did the screen oscillations interfere with your concentration? This question is related to screen oscillations due to the wheelchair movements, especially when it ends or begins a movement.
3) Was the green color of the stimuli annoying?

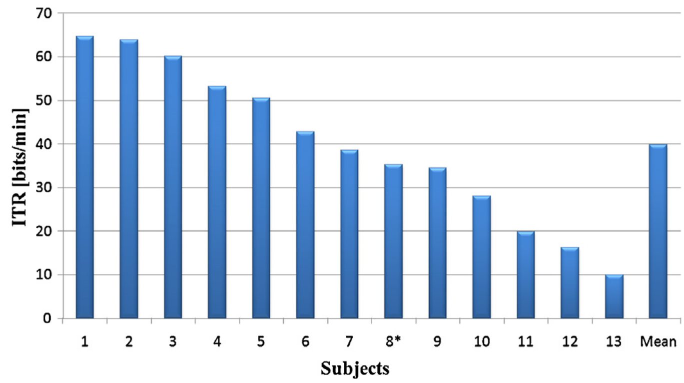

Fig. 4. Information Transfer Rate (ITR) per subject measured in bits × min^{-1} obtained in the training session. Subjects were sorted in descending order of performance. *Subject with disability.

4) Do you think that the position and separation of the stimuli are correct?
5) Was it exhausting for you to wear the EEG cap?

Except for question number 4 (whose answer should be Yes, No or Medium), they had to answer with the following options:

1: None.
2: A little.
3: Medium.
4: Quite.

5. Results

Fig. 4 shows the ITR reached by each subject in the training session (the phase before commanding the wheelchair, in which the BCI was active and the wheelchair was inactive). The subjects were sorted in decreasing ITR order, i.e., subject 1 has the best performance and subject 13 has the worst performance.

Additional information is supplied with this paper. Video #1 shows all the subjects performing the proposed tasks. Videos #2 and #3 show subjects 1 and 6 executing three and four tasks respectively. Video #4 shows the subject with disabilities performing all tasks.

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.medengphy.2012.12.005.

In Fig. 5a the SSVEP power is presented for each stimulus (according to (1)), whereas in Fig. 5b the corresponding detected classes is presented for the subject 9 executing Trial 2. These classes are translated into commands to the navigation of the wheelchair. Moreover, Fig. 5c shows, the path effectively followed by the wheelchair during this trial. In addition, a few time stamps were included in Fig. 5c, which are coincident with the time axis of Fig. 5a and b. As an example, in Fig. 5a, at $t = 17$ s the SSVEP corresponding to right stimulus is maximum; then in Fig. 5b, at $t = 19$ s, a command 'right' is triggered, and the wheelchair turns to the right. This is marked in Fig. 5c with a stamp. Finally, in Fig. 5a, at $t = 22$ s, the maximum SSVEP swaps from 'right' to 'top', the command 'right' is not sent and the wheelchair stops.

Fig. 5b shows an incorrect command ('right') sent at 39 s. In the video recorded during such trial the subject was looking at the up stimulus, consequently the command 'right' is an incorrect one. Moreover, the next commands, after this incorrect one, were 'advance'. Consequently, the subject was trying to advance and other commands different from 'advance' are incorrect commands.

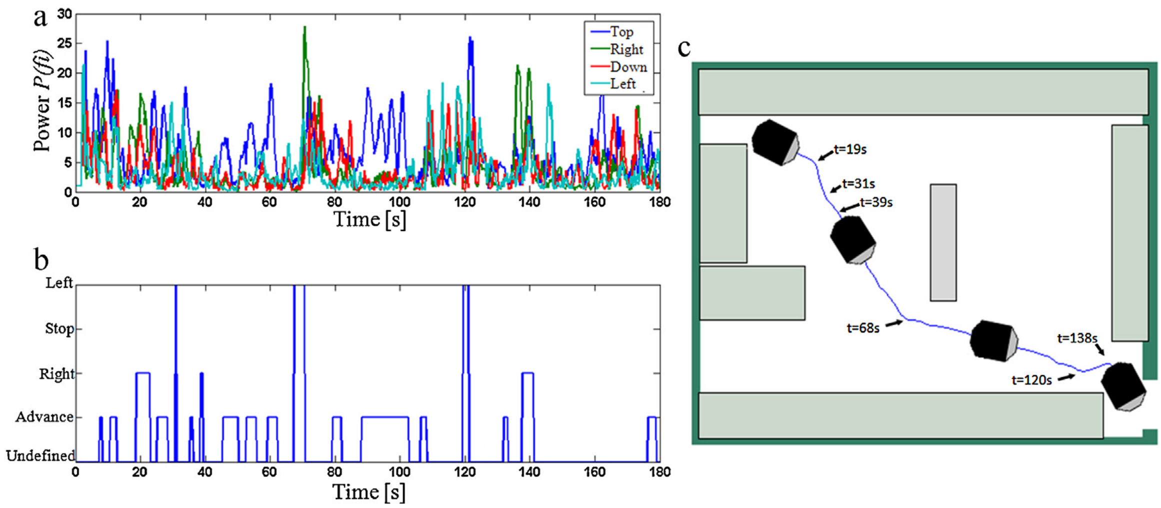

Fig. 5. Results of Trial 2 performed by Subject 9. (a) SSVEP power calculated according to Eq. (1); (b) Commands sent to the wheelchair based on the SSVEP powers showed in (a); (c) Path described by the wheelchair with time stamps corresponding to (b).

Table 1
Results obtained in Trial 1 and 2.

Subject	Trial 1						Trial 2					
	Time	Commands			ITR	VPP	Time	Commands			ITR	VPP
		C	I	ND	(bits/min)	(%)		C	I	ND	(bits/min)	(%)
1	2′58″	52	7	119	33.7	88.1	2′42″	72	3	87	51.0	96
2	1′51″	49	0	62	53.0	100	2′51″	65	0	106	45.6	100
3	1′28″	55	2	31	70.3	96.5	2′47″	81	1	85	57.1	98.8
4	2′16″	60	1	75	51.8	98.4	2′26″	67	0	79	55.1	100
5	2′23″	67	0	76	56.2	100	4′20″	92	4	164	41.2	95.8
6	2′20″	38	2	100	32.1	95	2′25″	64	0	81	53.0	100
7	1′40″	43	1	56	50.3	97.7	4′20″	56	8	196	25.9	87.5
8[a]	1′48″	56	1	51	60.5	98.2	2′54″	65	1	108	44.2	98.5
9	1′32″	45	0	47	58.7	100	2′59″	55	1	123	36.4	98.2
10	3′	43	0	137	28.7	100	3′50″	59	6	165	30.3	90.8
11	5′24″	45	9	270	17.7	83.3	5′5″	67	12	226	26.4	84.8
12	4′57″	45	7	245	18.9	86.5	4′40″	56	1	223	23.9	98.2
13	2′45″	48	0	117	34.9	100	6′33″	71	2	320	21.7	97.5
Mean	2′38″	49.7	2.31	106.7	43.6	95.7	3′41″	66.9	3.0	151.0	39.4	95.9

C: correct, I: incorrect and ND: non-detect number of commands.
[a] Subject with disabilities.

Table 1 presents the results of all subjects during Trial 1 and 2. It shows the elapsed time, the amount of commands sent by the BCI to the wheelchair, and the ITR achieved by each one. Table 2 summarizes the results of Trial 3 for all subjects. Subject 7 did not perform this trial because he had some detection problems with the left stimulus (40 Hz), i.e., the BCI could not detect the SSVEP associated to this stimulus with a good performance. Hence, subject 7 chose not to perform this task.

As mentioned before, Trial 4 was devised after some experimental sessions, since a high performance was observed in commanding the wheelchair. Therefore, just four subjects were invited to participate in it. The results obtained in this trial are shown in Table 3.

Fig. 6 shows the ITR achieved by each subject during the training session and during the trials with the wheelchair (Trials 1–4) in a single graph. In this way, it is simpler to compare the ITR achieved under all experimental conditions.

Table 4 details the answers of each subject to the survey questions about the tiredness they experienced after using the system. In the following columns the number of tasks accomplished by each subject and the duration of time they spent wearing the EEG cap for executing these tasks are observable.

Table 2
Results obtained in Trial 3 execution.

Subject	Trial 3					
	Time	Commands			ITR	VPP
		C	I	ND	(bits/min)	(%)
1	2′24″	82	2	60	65.5	97.6
2	3′42″	91	3	128	47.6	96.8
3	2″50	83	1	86	57.5	98.8
4	3′56″	85	6	145	41.4	93.4
5	2′34″	93	0	61	72.5	100
6	4′6″	78	3	165	37.3	96.3
8[a]	3′	94	0	86	62.7	100
9	4′34″	98	3	173	41.9	97
10	4′20″	69	6	185	31.2	92
11	6′45″	84	5	316	24.9	94.4
12	6′	80	8	272	26.6	90.9
13	6′	67	6	287	22.5	91.8
Mean	4′11″	83.7	3.58	163.7	44.3	95.8

C: correct, I: incorrect and ND: non-detect number of commands.
[a] Subject with disabilities.

Table 3
Results obtained in Trial 4 execution.

Subject	Trial 4					
	Time	Commands			ITR	VPP
		C	I	ND	(bits/min)	%
2	3′57″	114	2	121	56.3	98.3
4	5′47″	142	3	202	48.0	98
6	5′38″	112	6	220	38.6	95.2
8[a]	4′30″	142	2	126	61.6	98.6
Mean	4′58″	127.5	3.3	167.3	51.1	97.5

C: correct, I: incorrect and ND: non-detect number of commands.
[a] Subject with disabilities.

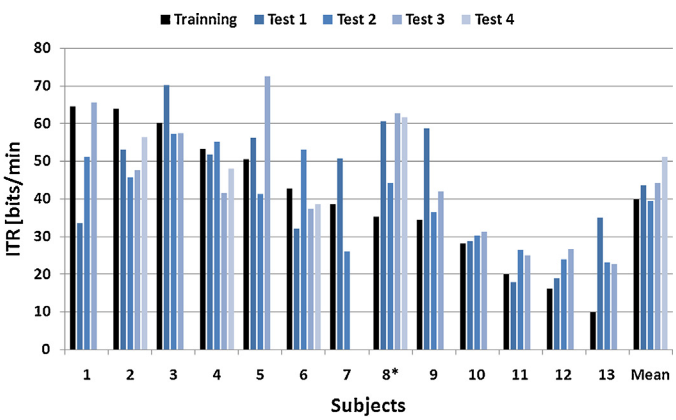

Fig. 6. The ITR obtained in the training session (black bar) and the four Trials (blue bars) on the wheelchair achieved by each subject. (For interpretation of the references to color in the artwork, the reader is referred to the web version of the article.)

6. Discussion

First an evaluation of the system and subject performances is presented, and then fatigue effects are analyzed. Finally, a comparison with other control systems is presented.

6.1. Evaluation of the system and subject performances

The ITR measured in the training session is different from that obtained with the trials onboard the wheelchair, as observed in Fig. 6 (Fig. 4 and Tables 1–3, as well). For example, subject 7 obtained 38.5 bits/min in the training session, 50.7 bits/min during Trial 1 but in Trial 2 this value dropped to 25.9 bits/min. Finally, the subject did not perform Trial 3 because he was upset by the low detection performance for the left stimulus. On the other hand, subject 8 achieved a performance similar to subject 7 in the training session (35.2 bits/min) and achieved 60.5 bits/min in Trial 1, and 44.1 bits/min in Trial 2. Then, in Trials 3 and 4, he obtained higher performances (62.7 and 61.6 bits/min, respectively).

There were two subjects who could not command the wheelchair: one healthy and one with disabilities. The healthy subject obtained an ITR of 16.37 bits/min in the training session. However, this value is higher than those corresponding to subjects 12 and 13, who could reach an effective control of the wheelchair. However, that subject could not stay concentrated in a stimulus when the wheelchair was moving. The subject with disabilities (female, 34 years old, suffering of tetra-paresis level C4-C5) was very drowsy and could not concentrate as required by the stimuli. In the training session she achieved only 7.3 bits/min. Unfortunately, she could not come back another day. A requirement for using a BCI system is concentration, i.e. the user must focus in the performed task. Possibly, a SSVEP based BCI is less demanding than other BCI paradigms such as mental tasks or motor imagery. However, concentration is still an essential requirement.

Bibliography has reported that approximately only 65% of people are able to operate a BCI based on high frequency SSVEP [33], meaning that the remaining 35% of people cannot achieve effective control of this kind of BCI system. However, in our trials only 2 out of 15 subjects (about 13%) were not able to operate our SSVEP-based BCI. A Proportion Test (binomial distribution, null hypothesis: $P = 0.35$, alternative hypothesis $p \neq 0.35$) was performed. As a result the two-tailed p-value of 0.123 was obtained. Therefore, the null hypothesis cannot be rejected and the two aforementioned subjects are included within that 35%.

Summarizing, it is hard to establish the performance of a subject in commanding a wheelchair a priori, since the performance of a subject depends on many factors, e.g. the task itself which is to be accomplished, the previous experience the subject had with the same or with a similar system, and the ability of the subject to learn how to use the system. Moreover, motivation and concentration are essential requirements for good performance, so that if the subject is drowsy or tired his/her performance will certainly low.

During Trial 1 and 2 (see Table 1) the highest ITR was obtained by subject 3. The subject with disabilities had similar performance. Subject 11 had the worst performance in Trial 1. The average time on Trial 2 was incremented in approximately 1′, compared to Trial 1, because of the increased complexity.

It is expected that high ITR should correlate with short trial competition time, but this is not observed in the experiment results The completion time of the task depends on, the different number of

Table 4
Results obtained from questionnaire.

Subject	1. Fatigue	2. Screen movement	3. Color fatigue	4. Stimuli separation	5. Fatigue for EEG cap	Cap wearing time (min)	Number of performed trials
1	2	2	1	Yes	1	23	3
2	1	1	1	Yes	2	37	4
3	2	1	1	Yes	2	21	3
4	1	1	1	Yes	1	37	4
5	2	1	2	Yes	1	27	3
6	2	2	1	Yes	1	42	4
7	1	1	1	Medium	1	32	2
8[a]	2	1	1	Yes	1	37	4
9	2	2	2	Yes	3	44	3
10	1	2	1	Medium	1	39	3
11	2	3	1	Yes	4	40	3
12	1	1	1	Yes	1	38	3
13	1	1	1	Yes	2	32	3
Mean	1.54	1.46	1.15		1.62	34.5	

1. None; 2. A little; 3. Medium; 4. Quite.
[a] Subject with disabilities.

commands (ND, C and I commands rate) used to calculate the ITR, the path described by the wheelchair, the time spent by the subject to evaluate his path and to decide what stimulus to gaze at, among other factors. For example, subject sends many correct commands (high ITR) to the wheelchair, but the wheelchair describes a longer path to the goal zone (high elapsed time).

Although the subjects were just getting familiarized with the system in Trial 1, their average ITR (43.6 bits/min) was greater than the one corresponding to the second Trial (39.4 bits/min). This can be justified because in Trial 2 the subjects had the additional task of evading an obstacle. Then, the subjects focused on the obstacle, instead of focusing on the screen and on the stimuli. This certainly produced an increment of non-detected cases. For instance, in Trial 1 there was an average of 106.7 non-detections, whereas in Trial 2 the average value was of 151. Moreover, there was an increment in the number of correct commands, which was expected, since it was necessary to send more commands in order to evade the obstacles.

In Trial 3, as shown in Table 2, the highest ITR was attained by subject 5. The subject with disabilities (subject 8) obtained a similar performance, and subject 13 obtained the lowest performance, but he was able to accomplish this third task, although in a longer time

The proposed BCI system was accurate for user command detection. The PPV values reported in Tables 1–3 indicate that when a command is detected, very probable that the correct one is. For example, the average number of correct commands in Trial 3 was greater than the average value of incorrect commands (83.7 and 3.58, respectively). This is an important characteristic of the designed system, because a high rate of false detection would lead the subject to lose his interest in using the BCI. For instance, the highest number of false detection in Trial 3 was obtained by subject 12 (8 commands), but he sent 80 correct commands. In Trial 1 2 the amount of incorrect commands was low as well. In Trial 3, a higher average ITR (44.3 bits/min) was reached, compared to Trial 1 (43.6 bits/min) and Trial 2 (39.5 bits/min), even performing a more difficult task, indicating that the subjects had become better familiarized with the system, which was expected.

In Trial 4, the subjects commanded the wheelchair through two corridors, which were narrower than the one presented in Trial 3. This means that the subjects should not make excessive incorrect command selections, in order to accomplish the Trial, because an excess of incorrect command selections could cause a collision. Although the Trial 4 is much more complex, the average elapsed time used to complete the Trial (4′58″) was just one minute greater than the elapsed time for Trial 3 (3′52″) (such comparison only considers the subjects who participated in both Trials). This shows an improvement in the ability of subjects to command the wheelchair, accompanying the increased experience in using the BCI, since they could complete a more complex task in a similar time. In Trial 4, subject 8 (the one with disabilities) obtained the highest ITR (61.6 bits/min) with a time of 4′30″. Subject 2 accomplished the Trial in a similar time, but with a lower ITR (56.3 bits/min).

In this Trial (the fourth one), the subjects with lower performances in the previous Trials (subjects 11–13) did not participate. However, they possibly could accomplish the task, although spending much more time. It is worth mentioning that they were not refused for this Trial, because this Trial was designed after their participation in the experiments, as previously explained.

6.2. Fatigue evaluation

Fatigue directly affects the performance of the subjects. Then, if the system leads the user to a fatigue state swiftly, the performance will be low. Hence, some items affecting the fatigue were analyzed. It can be observed in Table 4, that there was no (or even minimal) fatigue induced during the Trials. This evidence was obtained from the answers to question 1 of the questionnaire. Also, the green color of stimuli was not displeasing, as checked from the answers to question 3.

Studies on the flickering effects on visual fatigue due to high frequency stimulation were not reliably assessed. A state of the art on high frequency SSVEP based BCI was presented in [36]. In related articles [29–33] (included in such state of the art) have suggested the convenience of using high-frequency stimulation instead of low-frequency stimulation. However, neither of those work has effectively confirmed the decreasing of the visual fatigue of high frequency stimulation. Although, this statement is accepted in BCI researcher's community, it should be studied further. Comparatively with the statements from subjects participating in our previous experiments with low [1,41] and medium [42] frequency stimulation, the high frequency scheme is less discomforting. On the other hand, Allison et al. using flickering stimuli at 8 and 13 Hz, report that such frequencies were not that displeasing [43]. In the current work, it can be stated that high frequency stimulation was not annoying since fatigue effects were not reported by the subjects.

In this work, the average time of using the EEG cap was 35.5′ (SD: 7.1′) and according to the answers to question 5, the EEG cap did not cause too much discomfort. However, a person (with disability or not) using a BCI system for performing his/her daily tasks should wear an EEG cap many hours a day. For this reason, the designed system uses just three EEG channels, which allows using single electrodes instead of an EEG cap. The oscillation of the stimulation screen (question 2) did not affect the subject concentration.

6.3. Comparison with other systems

Generally, systems based on sEMG or EOG are faster and accurate than BCI systems, i.e., higher ITR can be achieved. However, user should retain neuromuscular control over his eyes [15], arms [16] or both eyes and facial muscles [17–19] for command a wheelchair. Users who not retain this neuromuscular control cannot use sEMG or EOG based systems and can benefit from BCI approach. Note that a SSVEP-based BCI would require less gaze control than an EOG system. On the other hand, some people have residual activity of their muscles and can benefit from hybrid approaches, i.e. a combination of a BCI with other biomedical signal such as sEMG or EOG [44]. Finally, the implemented system will depend on the user capabilities and limitations.

As mentioned before, a BCI system can also be based on other paradigms, such as motor imagery [10], mental tasks [8] or P300 potential [13,14]. However, such systems generally require a training step that can necessitate some minutes, hours or even days. In addition, the user is submitted to mental efforts that could produce fatigue and sometimes these systems are slow and cannot achieve high ITRs. On the other hand, SSVEP-based BCI systems do not have these problems, although the stimulation procedure can cause visual fatigue. In order to avoid this drawback, a SSVEP-based BCI system using high frequency stimulation was adopted in this work.

In all related work where a SSVEP-based BCI was used for commanding a wheelchair, low and medium frequency ranges stimuli were used [1,34,35]. Specifically, Teymourian et al. presented a simulated wheelchair commanded by three healthy subjects using stimuli between 5 Hz and 20 Hz and six EEG channels [34]. In the work presented by Mandel et al., eight healthy subjects commanded a real wheelchair using stimuli between 13 Hz and 16 Hz and minimum energy combination of six EEG channels for SSVEP detection [35]. Moreover, in [35] eight of nine subjects (one BCI illiterate) were able to command the robotic wheelchair in a path describing a figure of an eight. They required ten minutes for preparation and could send between an average of 14.5 and 20 commands, achieving accuracies higher than 90%. Finally, in the research presented by Torres Müller et al., ITR values up to

101.6 bits/min with an accuracy of 96% were achieved [1]. In that work, six healthy subjects commanded a wheelchair in order to reach the next room. Stimuli flicker between 5.6 Hz and 8.0 Hz was used, the detection of the SSVEP was based on a statistic test, and a rule-based classifier was implemented, working with the signals coming from twelve EEG channels.

In this paper, a BCI based on a simpler method (power spectral estimation of only three EEG channels) was used to extract the weakest SSVEP, which is located in the high frequency range (>30 Hz). This BCI allowed 13 subjects (1 with disabilities) to command a real wheelchair using high frequency stimulation. These stimuli did not produce (or produced only minimal) visual fatigue, which is crucial in long term BCI operation since it is difficult for tired subjects to achieve a high performance.

The stimuli screen in front of the user partially obstructs his/her visibility, although the monitor used in this work was quite small. Hence, another advantage of the present approach is that the monitor could be removed since it is used only for feedback purposes. This feedback could be presented to the user with another LED next to each stimulus, as an audible form, or even both. The stimulation system could be mounted on a transparent structure (such as an acrylic platform or something similar), thus increasing the visual field of the BCI user.

A SSVEP-based BCI is usually defined as a dependent BCI, because it requires some neuromuscular control of head and/or eyes. Moreover, stimuli must always be within the visual field of the user. This leads to the following drawbacks:

1) The BCI can detect a SSVEP and generate a command when the user does not want it. This problem is known as “Midas Touch Effect” [45].
2) The flicker stimulation can produce visual fatigue.

The developed BCI system is an approach trying to reduce these two drawbacks. The first issue is reduced since the BCI has a low error rate, avoiding the majority of the unwanted commands. The second issue is reduced thanks to the high frequency stimulation.

7. Conclusions

This research presents a BCI based on high frequency SSVEP for commanding a robotic wheelchair. The system was evaluated by 15 subjects (13 healthy and 2 with disabilities) who performed different navigation tasks, but only 13 subjects (12 healthy ones) could reach effective control. Specifically, they had to command the wheelchair from a start zone to arrive at a goal zone located next to the entrance to the room, at the same time that they evaded obstacles, traveled through corridors and avoided collisions with desks and other furniture, in four different scenarios. As a result, an overall average ITR of 44.6 bits/min was obtained along the four Trials, with a maximal ITR of 72.5 bits/min.

The developed BCI system achieves high ITR and low error command rate (average PPV 95%), thus allowing commanding a robotic wheelchair in only few minutes. Moreover, the subjects did not express any discomfort or fatigue due to the stimulation in a high frequency range. Hence, it is advisable that SSVEP-based BCI uses high frequency stimuli instead of other frequency ranges.

8. Future work

In this work, the wheelchair control strategy to avoid obstacles and collisions was performed by the user, but it is desirable to have an auxiliary autonomous navigation system, to be able to help the user in the accomplishment of such tasks, for security reasons and to prevent the user from becoming overly tired from using the BCI for long time intervals.

The performance of each subject depends on many factors, such as concentration and individual willingness, tiredness previous to the experiment, fatigue induced by the system, intrinsic parameters of the subject, as well as the signal processing methodology and stimulation characteristics (size, color, etc.). These factors need more profound study and research as to how they influence the performance when using a BCI.

In a future work, a control system that supports the user in the wheelchair navigation will be implemented. This includes a system to evade obstacles, avoid collisions, and pass through a doorway, among other tasks.

Conflicts of interest

The authors declare that they have no conflict of interests.

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