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# Wavelet lifting over information-based EEG graphs for motor imagery data classification

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**Abstract.** The imagination of limb movements offers an intuitive paradigm for the control of electronic devices via brain computer interfacing (BCI). The analysis of electroencephalographic (EEG) data related to motor imagery potentials has proved to be a difficult task. EEG readings are noisy, and the elicited patterns occur in different parts of the scalp, at different instants and at different frequencies. Wavelet transform has been widely used in the BCI field as it offers temporal and spectral capabilities, although it lacks of spatial information. In this study we propose the use of a tailored second generation wavelet to extract features from these three domains. This transform is applied over a graph representation of motor imaginary trials, which encodes temporal and spatial information. This graph is enhanced using per-subject knowledge in order to optimise the spatial relationships among the electrodes, and to improve the filter design. The resulting method improves the performance of classifying different imaginary limb movements maintaining the low computational resources required by the lifting transform over graphs. By using an online dataset we were able to positively assess the feasibility of using the novel method in online BCI contexts.

## 1 Introduction

Brain signal analysis, applied for the control of computerised devices, determines a human-machine interaction paradigm known as brain-computer interfacing (BCI). This new way of communication not only has a direct positive impact on motor disabled users in terms of quality of life and interaction with their surroundings, but also opens new modes of operation for healthy users to interact with their environment.

The human brain responds to different stimuli with alterations in its neural activity. These alterations which manifest as changes in the electrical activity in the cortex and changing blood oxygen levels in different regions, can be measured with appropriate technology. In terms of electrical activity, these responses provoke oscillations known as event related potentials (ERP) that can be measured with electroencephalographic (EEG) devices and detected with appropriate algorithms [11].

One example of an ERP is the imagination of limb movements, which is commonly known as motor imagery (MI). Depending on the limb involved, the brain activity derived from MI tasks produces changes in neural activity on different parts of the brain cortex, at different rates and with different temporal behaviour.

These changes produce a series of short lasting amplifications and attenuations in the EEG data known as event related desynchronisation (ERD) and event related synchronisation (ERS) [17].

The study of ERS/ERD has proven to be a hard task. EEG data is noisy and of low amplitude, there is no inter-subject pattern consistency, and features that make the ERS/ERD patterns recognisable appear at different time intervals, different scalp locations and different frequency bands.

Wavelets have been profusely applied in the BCI domain as they allow a meaningful temporal-spectral analysis of the EEG data. Shifts and dilations of a mother wavelet function provide a series of orthogonal subspaces resulting in what is known as multi-resolution analysis (MRA) [10]. First generation wavelets present a major disadvantage of difficult design. Researchers usually make use of well established wavelet families, even though the wavelet function features may not completely fulfil the needs of the domain of study.

On the other hand, during the last decade a new wavelet framework has become popular for building tailored wavelets. Wavelet lifting, often referred to as second generation wavelets, offers a simple framework to construct wavelet functions that have fewer restrictions and can be applied in a natural way to many situations that first generation wavelets are incapable of handling [23] [24].

The possibilities offered by the lifting transform open new interesting ways to look at BCI signal processing. In [1] a new MRA system for BCI data analysis was proposed using lifting scheme over graphs to fully explore the three domains involved in ERS/ERD patterns evolution. Graph EEG data representation is a natural way of describing the spatio/temporal relations among electrode readings. The purpose of this study is to extend the static graph representation by automatically building an enhanced graph in which the connections represent meaningful relationships among different electrodes. For this purpose we used mutual information as it provides a measurement of how much information one channel shares with another channel.

The paper is organised as follows: data acquisition is detailed in Section 2.1, Section 2.2 explains the lifting scheme over graphs, Section 2.3 describes how the graphs are built, Section 2.4 focuses on the feature extraction technique, pattern description and classification methods, and the experimental methodology is described in Section 2.5. The obtained results along with discussions are presented in Section 3. Finally, the conclusions are drawn in Section 4.

## **2 Methods**

### **2.1 Data Acquisition and Preprocessing**

The first dataset was recorded at the BCI Laboratory at the University of Essex. The protocol was set up as follows: The electrode placement followed the 10-20 international system and 32 channels were recorded with a sampling frequency of 256 Hz. During the recording session the subject was sitting on an arm-chair in front of a computer screen. A fixation cross was showed at the beginning of the trial at  $t = 0s$ . At  $t = 2s$  a cue was shown indicating the imaginary movement class to perform. The end of the trial was marked when the fixation cross and cue disappeared at  $t = 8s$ . The subjects were asked to perform 120 trials of each of the three imaginary movements (right hand, left hand and feet). A total of 12

subjects participated in the recording sessions, half of them were naive on the use of BCI systems, 58% of the subjects were female, and the ages ranged from 24 to 50. During the result analysis these subjects were identified by the prefix *E-X*, with *X* being the subject number.

The second dataset is from the BCI Competition IV (dataset 2a) and follows a similar acquisition protocol. The full experiment description can be found in [5]. The data covers four different types of MI movement data: right-hand, left-hand, feet and tongue recorded at 250Hz. There are a total of 288 trials recorded, for each of the nine subjects. The subjects belonging to this dataset are identified by the prefix *C-X*.

The calibration trials of the third dataset were acquired following an identical protocol to the one described for the first dataset. The validation trials were recorded while the subject controlled a BCI system in real-time with continuous feedback, even though the results for the online classification are not the ones described here, the data was reprocessed for this study. Twelve healthy subjects took part in the experiment, all of whom were naive to the use of online BCI. The subjects were aged from 24 to 32 and 50% were female. The results for this dataset are labeled with the prefix *O-X*.

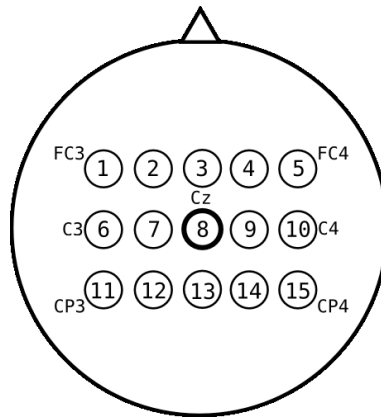


Fig. 1: Numbering of the 15 electrodes used during the experimentation, which were allocated from FC3 to FC4, C3 to C4, and CP3 to CP4.

For this study we utilised a subset of 15 electrodes, covering the major area of the motor cortex ( Figure 1). The original data was filtered from 8 to 30 Hz in order to attenuate external noise and artifacts. Each trial  $X_i$  of  $T$  samples was scaled by applying  $X_i = \frac{1}{\sqrt{T}} X_i^{orig} (I_t - 1_t 1_t')$ , where  $I_t$  is the  $T \times T$  identity matrix and  $1_t$  is a  $T$  dimensional vector with ones in it.

The competition data was already divided into training and evaluation subsets. The off-line data from the University of Essex was split using the first two acquisition runs (180 trials) as training data and the last two runs (180 trials) as evaluation set. The data from the online experiment contained 160 trials for calibration and 90 trials for validation purposes.

## 2.2 Wavelet Lifting over EEG Graphs

First generation wavelets represent signals in terms of shifts and dilations of the basis function known as mother wavelet. The design of this function obeys a set of restrictions assuring an accurate orthogonal decomposition of the original data. The main benefit of wavelet analysis over other orthogonal systems, such as the Fourier transform or the cosine transform, is its multiscale capability. Wavelets allow to analyse the data not only in the frequency domain, but also in the temporal domain at different levels. [10] [12].

The use of first generation wavelets is pervasive in many domains where signal processing is involved, and BCIs are no exception.

In P-300 based BCIs is common to find wavelet decomposition used as method of signal analysis. In [4] they use non-parametric statistics from the different levels of a Mexican hat common wavelet transform. The Daubechies wavelet in its discrete variant was applied in [16] using the thresholded coefficients as feature vectors for classification.

There are also several examples of the use of the wavelet transform in MI-based BCIs. In [18] Morlet wavelets are used to decompose signals of imaginary limb movements. In [6] discrete wavelet transform is applied using different families (Daubechies, Coiflets and Symlets), sub-band power and the standard deviation of the analysis and detail coefficients at different levels are used as feature extraction method. In [26] Coiflet decompositions of fifth order are used, as they claim that its shape is the closest to the ERD/ERS signals.

It is rarer to find the wavelet transform in the study of SSVEP BCIs but it has been applied, as in [25] where continuous Morlet wavelets were used to study the EEG data.

One major issue to cope with when working with the wavelet transform is that wavelet function design is an extremely complex task, and therefore, researchers apply common families in their studies despite the mother wavelet may not be suitable for the domain of study. The introduction of second generation wavelets, also known as wavelet lifting [22], alleviates this problem making the design of complete multiresolution systems more straight forward. The wavelet lifting is capable of handling data where Fourier analysis is not suitable (and therefore first generation wavelets either) such as unevenly sampled data, surfaces, spheres [20], trees [21] and graphs [14] [13].

A lifting scheme consists of iterations of three basic operations [7]:

- **Split:** Separate the original signal  $x$  into two subsets, referred as odd ( $x_o$ ) and even ( $x_e$ ) elements.
- **Predict:** The error of predicting  $x_o$  in base of  $x_e$  using a *predictor operator*  $\mathcal{P}$  conforms the wavelet coefficients  $d$ .
- **Update:** The coarser approximation of the original signal is calculated by combining  $x_e$  and  $d$  using an *update operator*  $\mathcal{U}$ .

A lifting transform over graphs can be defined as follows [14]. Let us consider a graph  $G = (V, E)$  where  $V$  is the node set of size  $N = N_o + N_e$  and  $E$  the edges linking those nodes.  $V$  is divided into the even and odd sets and  $E$  is represented using the adjacency matrix  $Adj$ . We rearrange  $V$  and  $Adj$  so that the odd set of nodes (a vector  $V_o$  of size  $N_o \times 1$ ) is gathered before the even set (a vector  $V_e$  of size  $N_e \times 1$ ), obtaining the following structure:

$$\begin{aligned}\tilde{V} &= \begin{pmatrix} V_o \\ V_e \end{pmatrix} \\ \tilde{A}j &= \begin{pmatrix} F^{N_o \times N_o} & J^{N_o \times N_e} \\ K^{N_e \times N_o} & L^{N_e \times N_e} \end{pmatrix}\end{aligned}\quad (1)$$

The submatrices  $F$  and  $L$  in  $\tilde{A}j$  in Equation (1) are discarded as they link elements within the same node sets. The block matrices  $J$  and  $K$  contain only edges linking nodes from different node sets.

The lifting transform functions are defined using a weighted version of the block matrices  $J$  and  $K$ :

$$\begin{aligned}D &= V_o - J^{\omega} \times V_e \\ A &= V_e + K^{\omega} \times D\end{aligned}\quad (2)$$

where the prediction and update functions are defined as a matrix product:  $\mathcal{P} = J^{\omega} \times V_e$  and  $\mathcal{U} = K^{\omega} \times D$ , where  $J^{\omega}$  and  $K^{\omega}$  are the weighted adjacency block matrices and their actual values depend on the domain of application.

We repeat the process described in Equation(2) in each level  $l + 1$  assigning the approximation coefficients  $A$  in level  $l$  to  $V$ .

### 2.3 Automatic EEG Graph Building and Filter Design

In [1], a static EEG data graph representation was introduced. This representation had the benefit of keeping a channel oriented structure although no extra information was used to optimise the inter-channel links. In order to establish which channels should be connected for each subject we made use of the mutual information of every pair of channels [9][15].

Mutual information measures the amount of information that one random variable  $Y$  contains about another random variable  $Z$  and is given by:

$$I(Y;Z) = \sum_{y \in Y} \sum_{z \in Z} p(y,z) \log \frac{p(y,z)}{p(y)p(z)} \quad (3)$$

where  $p(y,z)$  is the joint probability mass function and,  $p(y)$  and  $p(z)$  is the marginal probability mass function.

Consider a set of MI trials  $X^{T \times C}$  of  $T$  samples and  $C$  channels. In order to establish the relationships among the spatial locations we compute the mutual information  $M(r,s) = I(c_r; c_s)$  for every pair of channels  $c_r, c_s$  with  $r \in \{1 \dots C\}$  and  $s \in \{1 \dots C\}$ . Note that the diagonal elements of  $M$  are set to zero (the mutual information of a channel with itself is ignored) and rest of non-zero elements normalised between zero and one.

The symmetric matrix  $M$  describes how all the channels are related to each other and this information can be used to build a specific graph representation for each subject.

Let us assume that the graph  $G^x = (V^x, E^x)$  is embedding a trial  $X$ , where  $V^x$  defines the nodes and the edge set  $E^x$  is represented by a weighted adjacency matrix  $Adj^x$ :

$$Adj_{ij}^x = \begin{cases} a_{ij} & \text{If } v_i^x \text{ is connected to } v_j^x \\ 0 & \text{Otherwise} \end{cases} \quad (4)$$

For convenience, the odd set will correspond to the elements of  $X$  at odd values of  $t$ , and the even set at even values of  $t$ . Therefore, we obtain two different node vectors  $v_o^x$  and  $v_e^x$ .

The *predict* and *update* filters are computed in terms of the matrices  $M$  and  $Adj^x$ . The following steps are carried out in order to set the adjacency matrix values:

1. Apply a threshold  $th$  to the matrix  $M$  such that  $M(r, s) = 0$  if  $M(r, s) < th$ , so only those channels with high mutual information values will be linked, and normalise the non-zero values between zero and one.
2. Set  $Adj^x$  such that for a given channel  $c$  and instant  $t$  it will be connected to the previous  $t - 1$  and following  $t + 1$  time instants with a weight  $a_{ij} = 1$ .
3. For all the other channels  $c_r$  and adjacent temporal values  $t + 1$  and  $t - 1$  set the weight  $a_{ij} = M(c, c_r)$  in the corresponding entry of  $Adj^x$ , if  $M(c, c_r) > 0$ .

The resulting adjacency submatrices of  $F^x$  and  $L^x$  from  $Adj^x$  are empty. The *predict* and *update* matrices  $J^{\omega x}$  and  $K^{\omega x}$  (weighted versions the submatrices  $J^x$  and  $K^x$ ) are computed row-wise as  $J_{ij}^{\omega x} = \frac{J_{ij}^x}{\sum_{k=0}^J J_{ik}^x}$  and  $K_{ij}^{\omega x} = \frac{J_{ij}^x}{2 * \sum_{k=0}^J J_{ik}^x}$ . It is noteworthy to mention that the obtained lifting filters are weighted Laplacian graph filters, and the design explained here ensures that those channels that share high mutual information will contribute more to the detail coefficients than those that share low mutual information.

## 2.4 Feature Extraction and Classification

One of the main drawbacks in the use of multiresolution analysis for signal classification is the large number of coefficients generated during the transform. In order to overcome this problem we use common spatial patterns (CSP) as a method for feature extraction and dimensionality reduction. CSP is an extension to PCA where two different classes of data are taken into account.

Assuming that the trials contains data from the label (+) and the label (-), the set of samples  $X$  is divided into  $X^{(+)}$  and  $X^{(-)}$ . Their simultaneously estimated covariance matrix decomposition is given by [3]:

$$\begin{aligned}\Sigma^{(+)} &= W \Lambda^{(+)} W^T \\ \Sigma^{(-)} &= W \Lambda^{(-)} W^T\end{aligned}\quad (5)$$

where  $\Sigma^{(+)}$  is the estimated covariance matrix for the trials belonging to class (+) and  $\Sigma^{(-)}$  is the covariance matrix for the trials belonging to class (-).  $\Lambda^{(+)}$  and  $\Lambda^{(-)}$  are diagonal matrices with the eigenvalues corresponding to the decomposition of  $\Sigma^{(+)}$  and  $\Sigma^{(-)}$ . A large eigenvalue  $\Lambda_{jj}^{(+)}$  means that the corresponding eigenvector from matrix  $W$ ,  $w_j$ , leads to high variance in the projected signal in the positive class and low variance in the negative one (and vice-versa). The CSP projection results in  $Y_i = W^T X_i$ .

The different detail  $D^l$  and approximation  $A^l$  sets at different levels  $l$  from the MRA were projected onto their own CSP subspaces  $Y^{D^l} = W_{D^l}^T \times D^l$  and  $Y^{A^l} = W_{A^l}^T \times A^l$ . For clarity, we will refer to  $Y^{D^l}$  and  $Y^{A^l}$  using  $\bar{Y}$ .

For every  $\bar{Y}$ , we extracted the rows which maximised and minimised the variance between the two different classes (namely, the first  $m$  rows and last  $m$  rows) and calculated every feature as  $f_k = var(\bar{y}_k)$  with  $k = \{1, 2, \dots, m, C - m, C - (m - 1), \dots, C\}$ , obtaining a total of  $F = 2 * m$  features. In order to scale down

the difference among the feature values, the logarithm  $f_k^{log} = \log(\frac{f_k}{\sum_{j=1}^m f_j})$  was computed [19].

For this study,  $m \in 2, 3, 4$  was chosen using cross validation as explained in Section 2.5.

The features obtained from the CSP were classified using LDA as it provides a fair compromise between resource consumption and classification performance [2]. In order to measure the classification performance Kappa value [8] was used instead of the classification ratio. Kappa value gives an accurate description of the classifier's performance, taking into account the per class error distribution. The Kappa value was computed as  $\kappa = \frac{p_o - p_c}{1 - p_c}$ , where  $p_o$  is the proportion of units on which the judgement agrees (based on the output from the classifier and the actual label), and  $p_c$  is the proportion of units on which the agreement is expected by chance.

## 2.5 Experimental Methodology

After the data preprocessing, a temporal sliding window of one second with a fifth of second overlap was applied over each trial. The segmented data was then transformed using a lifting scheme over graphs (See Section 2.2 and Section 2.3) to the sixth level. The transformation resulted in twelve different coefficient sets, which were further processed to obtain the feature sets by selecting different number of CSP features (See Section 2.4).

The MRA approaches used for comparison were:

- Graph lifting scheme with static graphs [1]. The static graph is built by linking the elements from the surrounding channels as shown in Figure 2. The filters are calculated analogously as explained in Section 2.3 but by setting the weights of the Laplacian filters to one.

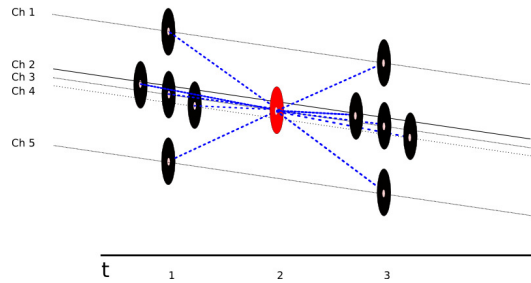


Fig. 2: Detail of the graph after the even/odd split for the static approach. The even element (in red) is linked to the surrounding odd elements (in black) adding spatial information to the decomposition.

- Graph lifting scheme and mutual information driven graph building.

Each detail and approximation coefficient sets from the different temporal segments were classified with a separate LDA model after applying CSP. This led to a total of  $n_s * l * 2$  LDA outputs, with  $n_s$  being the number of segments and  $l$  the



number of levels. A majority voting approach was carried out in order to obtain the final classification output for each trial.

A cross-validation step using five folds was performed over the training data in order to select the two free parameters involved: the threshold applied to the mutual information matrix in Section 2.3 and the number of CSP features as explained in Section 2.4.

### 3 Results and Discussion

From the analysis of the mutual information matrix  $M$  for the different subjects we learn that, in general, the standard deviations of the paired calculation do not differ much when compared among classes (two orders of magnitude smaller than the mean). Therefore, instead of computing a matrix  $M$  and a different graph to process one class against the others, we just use the whole set of data to generate the mutual information based adjacency matrix. This simplifies the model decreasing the execution time. It is also noteworthy that the performance of the transform calculation does not vary although the values of the filters applied changed.

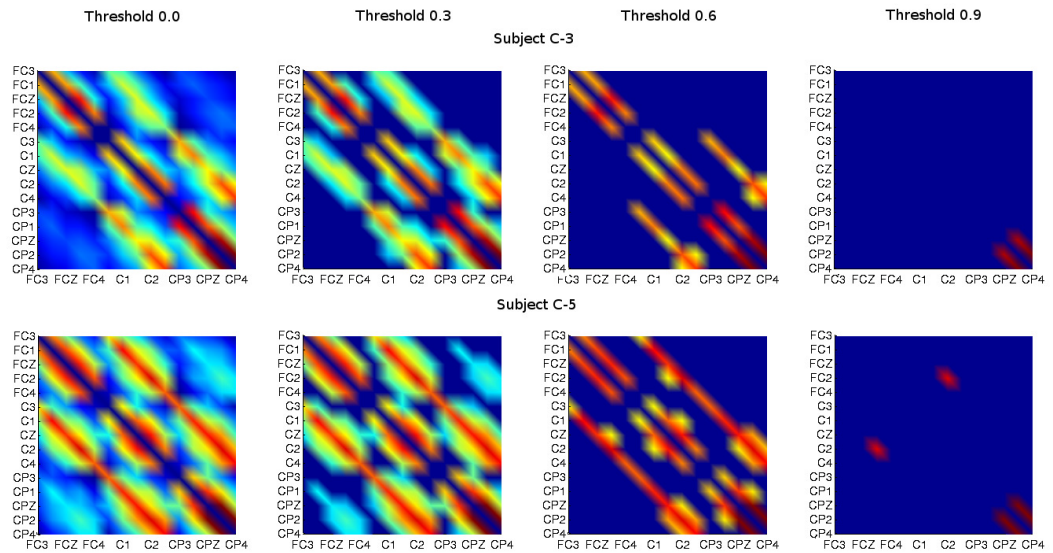


Fig. 3: Representation of the values of  $M$  for subjects C-3 and C-5 applying different thresholds

Figure 3, Figure 4 and Figure 5 are graphical representations of the values of the matrix  $M$  for different users and with different thresholds. In Figure 3 we find two examples to show that there exists a clear correlation between the electrode spatial location and the mutual information, the parallel lines crossing the figure diagonally corresponds to high mutual information values between adjacent

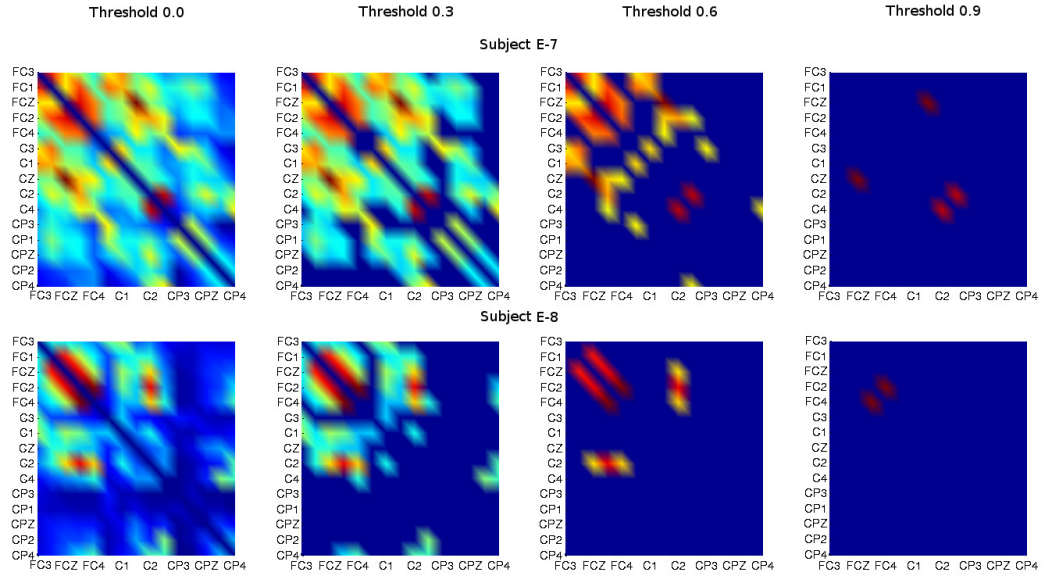


Fig. 4: Representation of the values of matrix  $M$  for subjects E-7 and E-8 applying different thresholds

electrodes. This correlation is more evident if we compare it with with Figure 6, which corresponds to the matrix  $M$  of the static approach. Although from Figure 4 to 5 this effect is still noticeable, it is more obvious how, for these specific subjects, the inter-electrode information is more prominent in particular locations of the motor cortex, concretely, in the frontal and central lobes for subject E-7 and 0-2, eminently central for subject E-8, and more scattered for subject 0-7.

After applying the experimental methodology described in Section 2.5 we can analyse the impact of the automatic graph building on the classification results. Figure 7, Figure 8 and Figure 9 show how the median Kappa values change when different threshold values are applied. It is clear that the behaviour of the method is dependant on the subject of analysis. Some subjects, such as E-8, C-4 and C-2, are not significantly affected by the change of the threshold value, although, on the other hand, we find subjects where the Kappa value fluctuates around 0.1 depending on the threshold value such as in subjects C-3, C-7, E-9, E-11, O-1 and O-4.

After selecting the two parameters, the mutual information threshold and the number of features used in CSP, from the cross-validation results we can compute the classification performance on the evaluation dataset. Table 1 shows the Kappa values and mean accuracy for the Essex dataset, Table 2 for the competition dataset, and Table 3 for the online dataset.

For 78.8% of the 32 subjects the proposed method achieved a higher Kappa value. For some of the subjects this improvement rises the Kappa value by 0.1 when compared to the static approach.

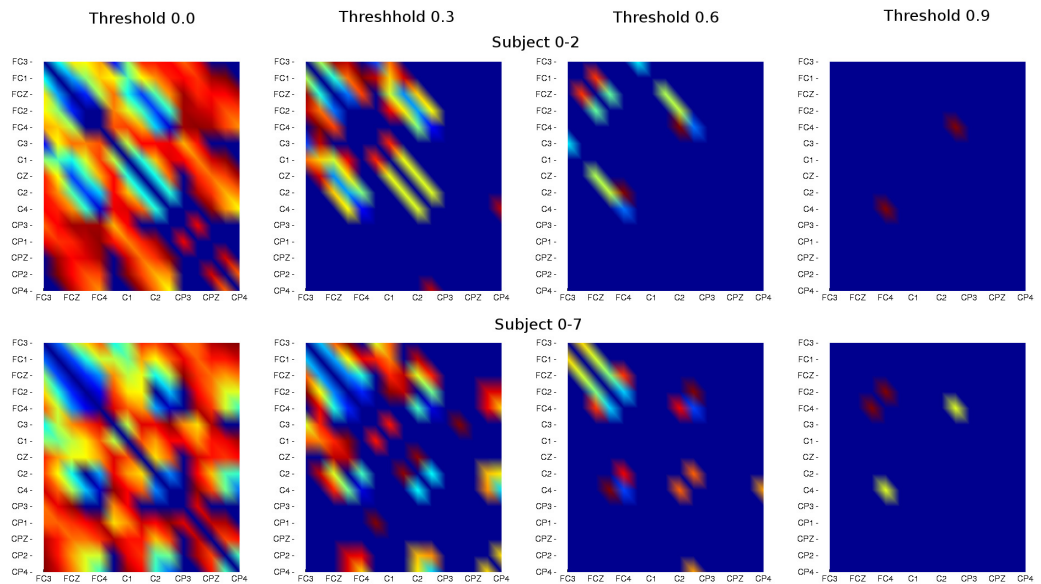


Fig. 5: Representation of the values of matrix  $M$  for subjects O-2 and O-7 applying different thresholds

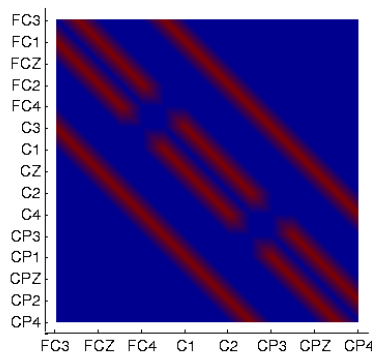


Fig. 6: Values of the matrix  $M$  for the static graph approach.

There exists a difference between the Essex and the online datasets in terms of classification accuracy, even though the acquisition protocol was practically identical. This decrement in the classification performance is due to effect of stress suffered by the subjects during the online experiments.

As a final remark we can mention that the proposed method obtains a Kappa value of 0.586 using the competition dataset, while the winner achieved 0.57.

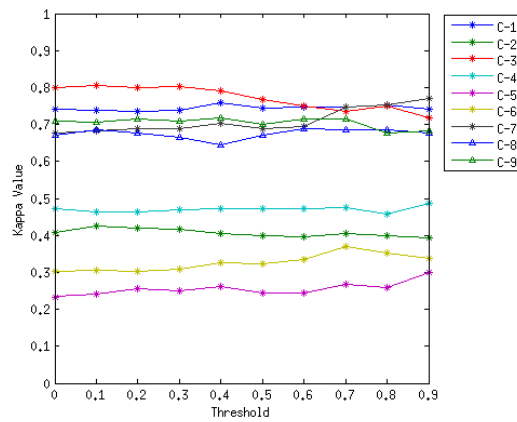


Fig. 7: Median of the Kappa value in function of the threshold applied to the matrix  $M$  for the competition dataset

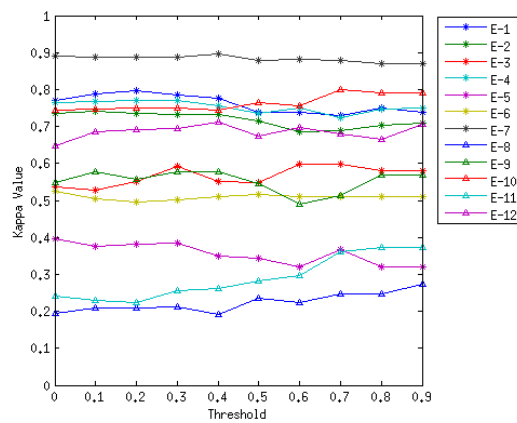


Fig. 8: Median of the Kappa value in function of the threshold applied to the matrix  $M$  for the Essex dataset

The small number of subjects in the competition data does not allow us to carry out a definitive significance test to compare both approaches.

## 4 Conclusions

In this study we have proposed a novel method to improve the EEG data representation based on static graphs by using the mutual information among the different channels, which constitutes an example of the possibilities offered by second generation wavelets in the BCI domain. This new strategy for building

Table 1: Results on the Essex dataset in terms of Kappa values. The mean accuracy is included at the bottom.

Subject	GLS	GLS + Mutual Information
E-1	<b>0.757</b>	0.741
E-2	0.736	<b>0.744</b>
E-3	0.539	<b>0.644</b>
E-4	<b>0.730</b>	0.712
E-5	0.392	<b>0.393</b>
E-6	<b>0.529</b>	0.488
E-7	0.883	<b>0.891</b>
E-8	0.210	<b>0.263</b>
E-9	0.565	<b>0.581</b>
E-10	0.757	<b>0.774</b>
E-11	0.237	<b>0.321</b>
E-12	0.648	<b>0.707</b>
Mean Kappa	0.582	0.605
	±	±
	0.21	0.19
Mean Acc	0.723	0.737
	±	±
	0.14	0.13

Table 2: Results on the competition dataset in terms of Kappa values. The mean accuracy is included at the bottom.

Subject	GLS	GLS + Mutual Information
C-1	0.754	<b>0.763</b>
C-2	0.410	<b>0.419</b>
C-3	0.800	<b>0.805</b>
C-4	0.484	0.475
C-5	0.243	<b>0.257</b>
C-6	0.317	<b>0.364</b>
C-7	0.629	<b>0.758</b>
C-8	0.661	<b>0.707</b>
C-9	0.698	<b>0.721</b>
Mean Kappa	0.555	0.586
	±	±
	0.19	0.20
Mean Acc	0.666	0.689
	±	±
	0.15	0.15

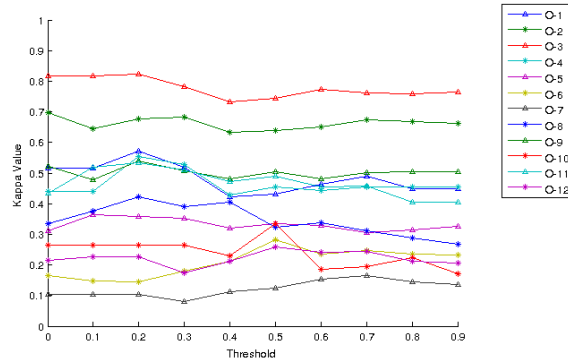


Fig. 9: Median of the Kappa value in function of the threshold applied to the matrix  $M$  for the online dataset

the graph also has an impact on the filter design, allowing an automatic way to weight the contribution of the different spatial locations.

After applying the proposed methodology on the three datasets, the resulting Kappa value was increased for an 78.8% percent of the subjects obtaining for several subjects an improvement of 0.1.

From the results achieved on the online dataset we can state that the proposed method is suitable for its use in online BCI systems, although the calibration time needed to obtain competitive results is relatively high, as the number of features used by CSP and the threshold have to be set via cross-validation.

Comparing the mutual information matrices from different subjects we can observe how the initial static graph approach, where the surrounding electrodes were linked together, was appropriate as close electrodes tend to share similar information.

These positive results encourage us to explore new ways for optimising the graph representation of EEG data. Although mutual information has helped to improve the classification rate, other techniques such as Granger causality, which could be more robust when coping with non-stationarity, should be examined in future work.

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Table 3: Results on the online dataset in terms of Kappa values. The mean accuracy is included at the bottom.

Subject	GLS	GLS + Mutual Information
O-1	<b>0.494</b>	0.454
O-2	0.640	<b>0.711</b>
O-3	0.777	<b>0.789</b>
O-4	0.432	<b>0.541</b>
O-5	0.334	<b>0.376</b>
O-6	<b>0.306</b>	0.278
O-7	0.107	<b>0.165</b>
O-8	0.248	<b>0.376</b>
O-9	<b>0.534</b>	0.504
O-10	0.222	<b>0.320</b>
O-11	0.434	<b>0.532</b>
O-12	0.179	<b>0.263</b>
Mean Kappa	0.392	0.443
	±	±
	0.18	0.17
Mean Acc	0.59	0.629
	±	±
	0.12	0.11

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