Spreading Codes Enables the Blind Estimation of the Hemodynamic Response with Short-Events Sequences

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Finite impulse response (FIR) filters are considered the least constrained option for the blind estimation of the hemodynamic response function (HRF). However, they have a tendency to yield unstable solutions in the case of short-events sequences. There are solutions based on regularization, e.g. smooth FIR (sFIR), but at the cost of a regularization penalty and prior knowledge, thus breaking the blind principle. In this study, we show that spreading codes (scFIR) outperforms FIR and sFIR in short-events sequences, thus enabling the blind and dynamic estimation of the HRF without numerical instabilities and the regularization penalty. The scFIR approach was applied in short-events sequences of simulated and experimental functional magnetic resonance imaging (fMRI) data. In general terms, scFIR performed the best with both simulated and experimental data. While FIR was unable to compute the blind estimation of two simulated target HRFs for the shortest sequences (15 and 31 events) and sFIR yielded shapes barely correlated with the targets, scFIR achieved a normalized correlation coefficient above 0.9. Furthermore, scFIR was able to estimate in a responsive way dynamic changes of the amplitude of a simulated target HRF more accurately than FIR and sFIR. With experimental fMRI data, the ability of scFIR to estimate the real HRF obtained from a training data set was superior in terms of correlation and mean-square error. The use of short-events sequences for the blind estimation of the HRF could benefit patients in terms of scanning time or intensity of magnetic field in clinical tests. Furthermore, short-events sequences could be used, for instance, on the online detection of rapid shifts of visual attention that, according to literature, entails rapid changes in the amplitude of the HRF.

Keywords: BOLD, fMRI, spreading codes, FIR, hemodynamic response.

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1. Introduction

The accurate estimation of the hemodynamic response function (HRF) is a challenging topic justified by clinical inference derived from the morphologic analysis. In the past, many researchers have developed abundant methods for this purpose. Here, we briefly mention some significant works, such as Refs. 1–3. In Ref. 4, the wavelet transform was presented as a rich source of new techniques to enhance analysis of fMRI data. More recently, in Ref. 5 a method based on nonparametric statistics proved to require significantly less computation and the same order of robustness than other methods. Currently, various approaches have been used that typically fit into parametric and nonparametric models.6

In parametric approaches, the HRF is synthetically rebuilt based on the expected similarity to one among a set of basis functions.7–9 Models are given the use of specific shapes as templates (e.g., Poisson, Gamma or Gaussian), which are tuned by means of few parameters, namely onset time, time-to-peak, peak amplitude and main to secondary amplitude ratio. This facilitates their estimation but models based on the assumption that HRFs across subjects share the same functional shape for a given region of interest (ROI) and stimulus8 will fail in the case of altered blood-oxygen-level dependent (BOLD). Studies have shown that the HRF can be absent,10,11 reduced,12,13 negative,14,15 delayed,16 with latencies to peak longer than expected,14,17 or with deeper initial dips.18 These considerations provide a justification for the use of blind methods for HRF estimation, such as the nonparametric models.

In nonparametric models, there are no assumptions about the HRF shape. Among them, the Finite impulse response (FIR) filters are considered the least constrained option and much more flexible than parametric models.18 Unfortunately, estimation of the filter coefficients (beta coefficients) is not exempt from drawbacks. For instance, as the filter order increases, the degrees of freedom decrease with a potential risk of overfitting. Another problem arises when using the standard approach to estimating beta coefficients, as follows:

\[
\beta_{FIR} = (X^T X)^{-1} X^T Y, \tag{1}
\]

where \(X\) is the design matrix containing the stimulus sequence and \(Y\) is the vector with the observed values (see Ref. 18 for a complete description of \(X\) and \(Y\)). In (1), matrix \(X^T X\) must be inverted. The inversion of this matrix causes numerical instability when the ratio of the number of observations to the filter order is small.18 Finally, another problem is that FIR solutions tend to be very noisy without high-quality data as many separate coefficients must be estimated.19 The latter necessarily gives rise to long-events sequences for a better performance. In order to mitigate this problem, a solution based on matrix regularization has been proposed.7,14,20 The sFIR model (2) consists on the addition in (1) of a smoothing or regularization term that introduces priors about a standard HRF response, thus biasing the beta coefficients, deteriorating the estimation in case of the existence of an altered BOLD and breaking the blind principle. Please, confront (7) for the definition of the variables of the regularization term.

\[
\beta_{sFIR} = \left( X^T X + \sigma^2 \Sigma^{-1} \right)^{-1} X^T Y. \tag{2}
\]

Spreading codes are an alternative to FIR and sFIR in case of short-events sequences. The use of spreading codes in fMRI studies (e.g. m-sequences) is not new. In Ref. 21, m-sequences showed better efficiency than random sequences. In Ref. 22, m-sequences were used to obtain a high-sensitivity estimates of HRF. In Ref. 23, the same author used m-sequences to minimize the impact of nonlinearities on the estimation of the HRF. In Ref. 24, a genetic algorithm was used to find the optimal experimental design for a multi-objective experiment and designs were compared with m-sequences. However, as far as we know, there are no studies that propose spreading codes as an alternative to FIR and sFIR for the blind estimation of the HRF with short-events sequences.

An area of application of sFIR could be the online detection of shifts of visual attention with short-events sequences. It has been proved that visual spatial attention modulates the shape of the HRF in regions of the cortex involved in this cognitive task. For instance, in Ref. 25 changes in the amplitude of the HRF were observed in an eccentricity-dependent manner. Also, in Ref. 26 variations on the latency of the HRF response were studied in contra-lateral and ipsilateral visual cortex. Despite shifts in visual attention is a cognitive ability that humans perform instantaneously, the detection in these studies was performed offline by using
long-event sequences. Our scFIR approach could enable the online detection by means of short-events sequences in Brain–computer interface (BCI) applications.

In this study, we present and alternative to FIR and sFIR for the blind estimation of the HRF based on short sequences of spreading codes. In this case, FIR and sFIR yield unstable and biased results, respectively. Therefore, scFIR could be an efficient option when the clinical testing requires short scanning time, online analysis of functional capacities and the intensity of the scanner is limited. Another advantage of scFIR is that, while FIR and sFIR require independent and identically normally distributed (i.i.d.) noise, the only restriction of scFIR is the uncorrelatedness of noise and the spreading code sequence. The latter makes unnecessary the use of pre-whitening filters as in FIR and sFIR case.

2. Materials and Methods

2.1. Underlying principles of scFIR

Let \( i[n], o[n] \) and \( h[n] \) denote the input, output and impulse response respectively of a Linear Time Invariant (LTI) system. The output of this LTI system, which is contaminated with noise \( e[n] \), represents the BOLD signal, the input represents the sequence of spreading codes and the impulse response represents the HRF. Equation (3) shows the relation between these signals.

\[
o[n] = h[n] * i[n] + e[n].
\]  

(3)

Let \( \hat{h}[n] \) be our estimation of the impulse response \( h[n] \) computed by direct circular cross-correlation of \( o[n] \) with \( i[n] \). That is,

\[
\hat{h}[n] = o[n] \odot i[n].
\]  

(4)

Taking the correlation with respect to \( i[n] \) in (4)

\[
\hat{h}[n] = o[n] \odot i[n] = h[n] * r_o[n] + e[n] \odot i[n].
\]  

(5)

Assuming (i) uncorrelated input and noise and (ii) the auto-correlation function of the input, namely \( r_o[n] \), approaches the unit impulse function \( \delta[i] \), we have

\[
\hat{h}[n] = h[n] * r_o[n] \approx h[n].
\]  

(6)

Notice that, unlike FIR and sFIR models, no matrix inversion is involved. The only algebraic operation to perform is the free-of-instabilities correlation of vectors. Regarding the presence of noise, the only condition to eliminate the presence of noise from (5) to (6), is the uncorrelatedness of the noise \( e[n] \) and the sequence of spreading codes \( i[n] \) used as input.

2.2. Simulated fMRI data

We evaluated scFIR and compared the performance with FIR and sFIR by using simulated fMRI data. We performed the blind estimation of a target HRF, namely HRF\(_2\), as well as a dynamic estimation of a changing-in-amplitude target HRF, namely HRF\(_1\). The blind estimation was focused on the ability of scFIR to estimate without priors target HRFs. In the dynamic estimation, the goal was to show the ability of scFIR to estimate dynamic changes in the HRF response as a consequence of simulated shifts of visual attention. In both, we used short-events sequences of spreading codes and the results were compared with FIR and sFIR. Equation (7) defines the regularization term included in (2):

\[
\sum_{ij} = \nu \exp \left(-\frac{\psi^2}{2(i-j)^2}\right); \quad \text{with}
\]

\[
g = \frac{1}{(\text{TR}/\psi)^{\nu}}, \quad \sigma^2 = 1,
\]  

(7)

where \( \sigma^2 \) is the noise level, \( \nu \) defines the strength of the prior, \( \exp() \) is the operation \( e^x \), \( g \) is the smoothness factor, \( i \) and \( j \) are the row and column respectively of the regularization term and TR is the repetition time. In sFIR, the parameters of the regularization term were set up to values considered optimal in previous relevant studies \( \psi = 0.5 \) or \( \psi = 2 \) and \( \nu \) conveniently equals the noise level.

In order to compare FIR and sFIR, normalized correlation coefficients (\( R \)) were calculated in a run with large number of trials. The normalization of \( R \) was performed by dividing the covariance of the estimation and the target HRF by the product of their standard deviations. The correlation of random signals is not normally distributed. Therefore, Fisher transformation was applied and then the mean and confidence intervals of the normalized \( R \) computed (the confidence intervals are based on the normal distribution of the Fisher Transform). Afterwards, the inverse Fisher transform was applied for the representation in the results section.
The estimation of the confidence interval was at a significance level of $\alpha = 0.05$. Since scFIR was compared with both FIR and sFIR, the Bonferroni correction was taken into account.

Among several families of spreading codes we chose m-sequences because they have the best auto-correlation properties. However, the auto-correlation function of short m-sequences does not approximate a $\delta[n]$ function required in (7), thus causing distortion in the estimation of HRF. In order to mitigate this effect the output of scFIR was low-pass filtered by a second-order Butterworth filter that was executed forward and reverse to cancel phase shifts. Note that the net effect is a fourth-order filter with 6dB of loss at the original 3dB cutoff frequency. This filtering cannot be considered a consequence of priors about the HRF shape, but rather a direct consequence of the use of realistic m-sequences with non-ideal correlation properties. Therefore, this filtering does not invalidate the blind principle. For this reason, the cutoff frequency was arbitrarily set to a third of sampling rate that, in turns, was arbitrarily set to 1 s.

Numerous simulated BOLD signals were generated by using multi-combinations of m-sequences, various levels of signal-to-noise ratio (SNR) and two target HRF1 and HRF2. Each combination was obtained by the convolution of one of the m-sequences by one of the HRFs plus the addition of white noise, thus giving rise to simulated BOLD signals of different quality.

In order to obtain ground truth in our simulations, two target HRFs were synthesized. The first one (HRF1) was generated by means of two Gamma functions

$$h[n] = A \left( \frac{\Gamma(\alpha_1 - 1 / 2) \alpha_2^\alpha_1 - 1 / 2}{\Gamma(\alpha_2)} - \frac{\Gamma(\alpha_2 - 1 / 2) \alpha_2^\alpha_1 - 1 / 2}{\Gamma(\alpha_2)} \right)$$

with standard parameters ($\alpha_1 = 6, \alpha_2 = 16, \beta_2 = 1, c = 1/6, A = 1$) as proposed in literature\(^7\) (see Fig. 1, left). We also emulated a real hemodynamic response (HRF2) obtained from a patient suffering of aphasia.\(^8\) Both target HRFs are composed of 30 beta coefficients. Y-axis in arbitrary units.

real world with totally different envelop as the classical HRF\(_1\). This would reinforce the blind character of the estimation.

The SNR levels, namely $-6$ and $6$ dB were chosen as representative of BOLD signals with poor and good quality respectively because typical fMRI signals likely fit within this interval. Furthermore, simulation of BOLD signals with good quality points out the performance justified only by the method without affection of signal with low quality. The sequence lengths (SLs) were chosen as representative of short (15, 31), medium (63) and large (127) events sequences. In this study, we are interested in the short-events sequences. Medium and large sequences are shown in order to see the equivalence of the three methods in terms of performance with the increase in the number of events.

2.2.1. Blind and dynamic estimation

The blind estimation consisted on the estimation of the target HRF\(_2\) (see Fig. 1 right) from various simulated BOLD signals with good and bad quality (SNR = $-6$ and $6$ dB respectively) generated by short-events sequences of 15, 31, 63 and 127 events.

For the dynamic estimation, we simulated sequences of on/off attention to visual stimuli. According to literature,\(^9\) the amplitude of the HRF can change 60% and 20% for eccentricities smaller and greater than $2.5^\circ$ and $8.5^\circ$, respectively in V1 region between the Attend and the Ignore condition in covert attention paradigms. Since changes in attention can be performed instantaneously by human, dynamic estimation would account for the
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Fig. 2. Shifts in visual attention to stimuli were simulated by means of changes in the amplitude of the HRF. The figure represents a sequence of conditions A-I-A and the corresponding target HRF. The dotted line corresponds to the measurable “level” of attention.

RAPID TRANSITION BETWEEN THE ATTEND/IGNORE CONDITIONS. Shifts in visual attention to stimuli were simulated by means of changes in the amplitude of the HRF of 60% of the maximum value when the stimulus is under the Ignore condition (see Fig. 2). The figure represents a sequence of two conditions Attend-Ignore-Attend and the corresponding HRF.

The goal of the three methods was the dynamic and accurate estimation of the amplitude of the target HRF. In this section, we used the target HRF to generate the BOLD signals. We considered the 6th beta coefficient as the estimated amplitude (see amplitude of HRF in Fig. 1, left).

In order to generate the “level” of attention with values 100% and 60% (see Fig. 2), a continuous estimation of the HRF was performed by feeding the three FIR, sFIR and scFIR methods with a sliding window of simulated BOLD signals (window size = the SL; window shift = 1 event). The circular correlation properties of the m-sequences, guaranteed the best correlation properties in any of the sliding window. The duration of each condition was approximately 100 times the duration of the SL in event units. Note that we used a sampling rate (or event unit) of 1 s and 30 beta coefficients. That means that each target HRF lasted 30 s. The latter guaranteed to reach the permanent regime much before a condition transition. Finally, the output of the three methods was smoothed with the same low-pass filtering. Since the “level” of attention remains constant within each A-I-A block (see dotted line in Fig. 2), this filtering caused the elimination of a priori unnecessary high frequency components in the sliding estimation of the HRF amplitude. Therefore, this filtering does not invalidate the blind principle of the estimation. It is only the consequence of prior knowledge of the fMRI block design and not of that of the HRF.

2.3. Experimental data

An experimental data set from previous fMRI studies was used to test our method. Please refer to Ref. 33 for a comprehensive explanation about the stimulus characteristics, data acquisition and preprocessing methods. A brief summary follows below.
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MRI data were obtained with a 3.0-Tesla Siemens MAGNETOM Trio A Tim scanner located at the ATR Brain Activity Imaging Center (Japan). The acquired fMRI data were slice-timing and motion corrected using SPM2. Basic pre-processing such as linear trend removal and normalization was applied. The main objective of Ref. 33 was the reconstruction of binary-contrast $10 \times 10$-patch images from voxel activity. For this purpose, a random-block design was used with 20 runs of 22 different blocks per run and one random image per block, thus giving rise to 440 different random images (see Fig. 3 left) and three volumes per block. Each block (both active and inactive) lasted 6 s and contained three volumes (TR = 2 s). In each block, an image consisting of $10 \times 10$ random binary-contrast patches was shown. Each of the 100 patches was either flickering at 6 Hz (“1” or active condition) or a homogeneous gray (“0” or inactive condition) with equal probability.

The objective of this section is the estimation of the experimental target HRFs from various voxels arranged in iso-eccentricity sets. The retinotopy principle links specific visual field locations to specific cortical location, thus forming a mapping from the visual field to the cortical voxels. Due to the magnification factor, this spatial mapping is one-to-many for central patches and many-to-one at the periphery (i.e. the activity of voxels mapped to central and periphery patches is likely caused just by one and many stimuli, respectively). In other words, the visual receptive field of receptors in the central area is smaller than that of the periphery. That means that stimuli from central patches are expected to cause activations in more voxels than those coming from the periphery.

Conversely, voxels that according to retinotopy respond to stimuli from the periphery will cause fMRI signals associated to more than one patch, so these voxels are contaminated with noise caused by neighboring stimuli. The immediate consequence is that BOLD signals extracted from voxels activated by periphery will have components correlated with more stimuli than those activated by central vision. Based on this knowledge we would expect voxels activated by patches situated at a certain eccentricity to have a HRF with similar amplitude. For this reason we chose three sets of patches arranged by crescent eccentricity, namely ecc0, ecc1 and ecc2. In each set, the patch number closer to the iso-eccentricity circle was included (see Fig. 3, right). For each patch of each iso-eccentricity set, the voxel with the higher SNR was found and associated to the patch. The SNR of each voxel related to a patch, was computed as the energy of the sequence label of that patch divided by the variance of the BOLD signal once the sequence label was removed. The BOLD signal of each voxel was composed of 440 blocks. It was divided in training data (380 blocks or 2280 events) and evaluation data (60 blocks or 360 events). For each voxel, we estimated the HRF with the training data. It was considered the target HRF. The target HRF was estimated as the mean value of sFIR and scFIR estimations (FIR was not taken into account due to numerical instability). Then, we estimated the target HRF from the evaluation data. Since voxels from each iso-eccentricity set are located in different brain areas (upper-lower and left-right hemispheres), each of the estimated HRF could not share the same shape. For this reason, we discarded voxels within each iso-eccentricity set whose HRF was clearly different from the others. It corresponds to voxels associated to patches filled in gray in Fig. 3, right. Afterwards, the number of remaining voxels in each iso-eccentricity set was four, nine and twelve for ecc0, ecc1 and ecc2 respectively (see patches within unfilled circles in Fig. 3, right). The evaluation data were submitted to FIR, sFIR and scFIR methods to estimate the HRF with small number of samples. Normalized $R$ and mean-square error was computed. As with the simulated data, Fisher transform was used before mean values of $R$ were obtained. Also, Bonferroni correction was applied ($\alpha = 0.05$).

2.4. Structured versus white noise

In the simulated section (blind and dynamic estimation) white noise was added to the BOLD signal. We did it since FIR and sFIR require white noise for an optimum performance, thus favoring them in the comparison with scFIR. There are multiple sources of low-frequency structured noise in fMRI signals (e.g. hardware imperfections, heart rate, and respiration, movement artifacts, etc.). These contributions give rise to a certain level of autocorrelation in the residuals of the fMRI signal, thus invalidating the statistical analysis. In the presence of structured noise, some pre-whitening filters (or high pass
filters) are typically used for optimal performance of FIR and sFIR. However, this restriction does not apply for scFIR, whose performance only relies on the uncorrelatedness of noise and the spreading codes (see Eqs. (5) and (6)). This is a much more flexible restriction that permits the typical presence of structured noise in fMRI signals without the need of pre-whitening filters.

3. Results

3.1. Blind estimation

This section shows beta coefficients estimated by FIR, sFIR and scFIR with simulated BOLD signals. BOLD signals were simulated by several combinations of SNRs (SNR = −6 and 6 dB), SLs (SL = 15, 31, 63 and 127) and the target HRF$_2$. Plots without curves (e.g. in FIR column, SL = 15 and 31) means inability to estimate coefficients due to numerical instability. Figure 4 shows a representative example of an estimation of the target HRF$_2$ (small circles).

For statistical purposes, we estimated the target HRF$_2$ 1000 times for each combination of BOLD signals (except for SL = 127, for which we performed 4000 estimations). Figure 5 shows the mean normalized R and confidence intervals (see Sec. 2.2 for further details about their computation). The high number of estimations guaranteed hardly visible confidence intervals, thus facilitating the search for significant differences in performance by visual inspection. Notice that for SL = 15 and 31, FIR bars are missing due to numerical instability.

3.2. Dynamic estimation

This section shows the estimation of the “level” of attention (dotted line in Figs. 2 and 6) by FIR, sFIR and scFIR. BOLD signals were simulated by multiple combinations of SNRs (SNR = −6 and 6 dB).

![Fig. 4 Beta coefficients estimated by FIR, sFIR and scFIR. In each plot, the circles represent the target HRF$_2$ (see Fig. 1, right), the thin and thick lines represent the estimated beta coefficient with bad and good BOLD signal quality (SNR = −6 and 6 dB, respectively). SL: sequence length in events units.](image)

![Fig. 5 Normalized R and confidence intervals for the estimation of target HRF$_2$ by FIR, sFIR and scFIR. Several simulated BOLD signals were generated as a combination of different SLs (SL = 15, 31, 63 and 127), different signal qualities (SNR = −6 and 6 dB) and two target HRFs (HRF$_1$ and HRF$_2$). For SL = 15 and 31, FIR was unable to compute the beta coefficients and sFIR achieved a nearly uncorrelated estimation of the target HRF$_2$. For all values of SL, the normalized R of scFIR significantly outperformed that of FIR and sFIR. SL: sequence length in events units.](image)
SLs (SL = 15, 31, 63 and 127) and target HRF. The “level” of attention was considered the 6th estimated beta coefficient, which corresponds to the amplitude of target HRF \(1\). The "level" of attention was considered the 6th estimated beta coefficient, which corresponds to the amplitude of target HRF \(1\). We simulated shift of visual attention by means of modulation of this amplitude between values 100% and 60% (Attend and ignore conditions, respectively). Plots without curves (e.g. in FIR column, SL = 15 and 31) means inability to estimate coefficients due to numerical instability. Figure 6 shows a representative example of estimation of target HRF \(1\).

For statistical purposes, we estimated the “level” of attention by estimating the amplitude of the HRF more than 6000 times per condition (sequence A-I-A).

Table 1. Normalized R and confidence intervals \((\alpha = 0.05)\) for the estimation of the “level” of attention. Cells with gray backgrounds mean values of \(R\) significantly better than the other two methods. Note that FIR was unable to compute the beta coefficient due to numerical instability for SL values up to 31, while scFIR estimated them with an extraordinary high correlation \((R = 0.96)\).

<table>
<thead>
<tr>
<th>SL</th>
<th>SNR (dB)</th>
<th>xFIR</th>
<th>(R_{\text{min}})</th>
<th>(R_{\text{max}})</th>
<th>sFIR</th>
<th>(R_{\text{min}})</th>
<th>(R_{\text{max}})</th>
<th>scFIR</th>
<th>(R_{\text{min}})</th>
<th>(R_{\text{max}})</th>
</tr>
</thead>
<tbody>
<tr>
<td>15</td>
<td>−6</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
<td>NaN</td>
</tr>
<tr>
<td>31</td>
<td>−6</td>
<td>NaN</td>
<td>0.01</td>
<td>0.02</td>
<td>NaN</td>
<td>0.01</td>
<td>0.02</td>
<td>NaN</td>
<td>0.01</td>
<td>0.02</td>
</tr>
<tr>
<td>127</td>
<td>−6</td>
<td>0.83</td>
<td>0.83</td>
<td>0.84</td>
<td>NaN</td>
<td>0.83</td>
<td>0.84</td>
<td>NaN</td>
<td>0.83</td>
<td>0.84</td>
</tr>
</tbody>
</table>

in Figs. 2 and 6). Table 1 shows the normalized \(R\) and confidence intervals between the simulated “level” of attention and the estimated one. Gray backgrounds represent significant differences. Notice that for SL = 15 and 31, FIR was unable to compute the HRF due to numerical instability.

3.3. Experimental fMRI data

Figure 7 shows statistical results from experimental fMRI data. FIR results were very poor in comparison with sFIR and scFIR and were not plotted. In this case, the use of sFIR is the simplest workaround to the numerical instability of FIR.
4. Discussion

4.1. Blind estimation

Figure 4 shows single-trial estimations of the target HRF for different values of SNR and SL. In this way, the shape of each estimation can be visually inspected without benefitting from any denoising process caused by trials averaging. Notice that for the shortest SLs (SL = 15 and 31), FIR was unable to compute the HRF, sFIR approximately yielded an uncorrelated shape and scFIR reported an extraordinary normalized R, that, in the case of BOLD signal with good quality (SNR = 6 dB, SL = 31) exceeded 0.9 (see Fig. 5, upper right plot). For the medium and long SLs (SL = 63 and 127), the difference in performance between FIR, sFIR and scFIR gradually disappear. In the best scenario (SL = 127, SNR = 6 dB), the difference in performance between the three methods is very small, and despite it is significantly different in favor of scFIR, it suggests that longer SLs would cause total convergence in performance.

In summary, and despite we favored sFIR by adjusting the ψ parameter, we state that in terms of normalized R, scFIR performed significantly better than FIR and sFIR with short-events sequences and simulated BOLD signals.

4.2. Dynamic estimation

We can state similar appreciations as in the discussion of the blind estimation. In summary, for short sequences (SL = 15 and 31), FIR and sFIR were unable to compute the amplitude of the HRF with accuracy (see Fig. 6, first and second columns, first and second rows), whereas scFIR was able to compute the amplitude with accuracy (e.g. normalized R between 0.80 and 0.96, see Table 1). The estimation of the HRF has a delay of exactly the SL (e.g. for SL = 127, the three methods would output...
Figure 7 shows the experimental target HRF (first 4.3. Experimental fMRI data

4.4. White versus structured noise

In scFIR, the uncorrelatedness of noise and the input sequence is assumed. This represents an extraordinary advantage of scFIR in comparison with FIR and sFIR. These two methods require i.i.d. noise for an optimal performance. In broad words, they need white noise. In practice, this leads to the use of pre-whitening filters (typically high-pass filters) that, somehow, means the enhancing of the level of noise. With the scFIR approach this is not necessary because uncorrelatedness is less restrictive than independency. As an example of the latter, we could state that scFIR performs optimal with any fMRI signal contaminated by structured (nonwhite) noise as long as the noise and the input sequence are uncorrelated to each other. In practical terms, the latter can always be assumed. In this case, both FIR and sFIR will perform suboptimal since structured noise is not i.i.d. noise. For this reason the fMRI signal typically are high-pass to equalize low and high spectral bands of the noise, thus adopting the shape of white noise. This operation is not needed for scFIR. Furthermore, this operation is completely irrelevant for scFIR as long as the uncorrelatedness of noise and the input sequence remains unaltered and, a priori, there is no reason to think so.

The results showed in Secs. 3.1 and 3.2 were obtained with simulated BOLD signals contaminated by white noise, thus giving rise to levels SNR = −6 and 6 dB. As explained before, the use of white noise is necessary for the optimal performance of FIR and sFIR. This is not the case for scFIR, which perfectly tolerates the presence of structured (nonwhite) noise providing that it is uncorrelated with the input sequence. Even in this optimal scenario for FIR and sFIR, scFIR demonstrates much better performance with short sequences.

Some preliminary tests were executed with structured noise (that is noise with a certain level of autocorrelation), obtained by simply low-pass filtering of white noise and the results were, as expected, even better than those presented here with white noise.

5. Conclusion

In summary, in the present study we have presented and alternative to FIR and sFIR for the blind and dynamic estimation of the HRF with short-events sequences based on spreading codes. Whereas FIR
and scFIR yielded unstable solutions and biased solutions, respectively. scFIR presented significant difference in normalized \( R \) and mean-square error with short-events sequences.

The scFIR approach can be considered a blind and improved version of FIR and sFIR that performs better with small number of observations, reduces computational complexity, and lacks ill-posedness without the use of a regularization term. It is achieved because in scFIR the estimation of the beta coefficients relies on simple correlation of vectors.

The scFIR approach can be applied in scenarios such as for the blind estimation of BOLD (e.g. in cerebral-vascular impaired patients), when the clinical test requires short scanning time or the intensity of the scanner is limited, and for the online detection of changes in the amplitude of the HRF.

A specific area of application of scFIR could be the online detection of shifts of visual attention, or more generally, the online detection of attention with BCIs. In the case of visual attention, BCIs are designed to detect endogenous attention (covert attention) without muscular movement, thus providing additional information to that provided by other devices such as eye-trackers based on shifts of gaze. In some cases, when a user is unable to shift the gaze correctly (e.g. users with locked-in syndrome or Amyotrophic lateral sclerosis), covert attention is the only way to detect the user visual attention. However, EEG-based BCIs have proved extremely bad performance in detecting visual covert attention. For instance, from the conclusion section of Ref. 40, we can infer that the BCI could only detect visual covert attention at a rate of on one per minute approximately. In this regard and given the delays discussed in Sec. 4.2 and performance of Table 1, we could roughly expect a HRF-based BCI to take less than 10s in the detection of visual covert attention, thus constituting a novel and promising paradigm.

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