Abstract

It is believed that human primary visual cortex (V1) increases activity with increasing temporal frequency of a visual stimulus. Two kinds of visual stimulus were used in the previous studies, one is patterned-flash stimulus with a fixed onset period and an increasing average luminance with the increase of temporal frequency, the other is contrast reversing flickering checkerboard or grating with a constant average luminance across different temporal frequencies. That hemodynamic responses change as a function of reversal frequency of contrast reversing checkerboard is at odds with neurophysiological studies in animals and neuroimaging studies in humans. In the present study, we addressed the relationship between reversal frequency of contrast reversing checkerboard and hemodynamic response in human V1 using an event-related experimental paradigm and found that the transient characteristics of blood oxygenation level dependent response in human V1 depended very little on the reversal frequency of a contrast reversing checkerboard.

Introduction

A prevailing view from neuroimaging experiments is that human visual cortex increases activity with increasing temporal frequency of a visual stimulus. Using a patterned-flash with a fixed stimulus-on period (5 ms), Fox and Raichle [1] found that the regional cerebral blood flow (rCBF) measured with positron emission tomography (PET) changes in human visual cortex as a linear function of stimulus rate between 0 and 8 Hz, peaked at ~8 Hz, and decreased at still higher rates. Similar results were also reported by a number of investigators using both PET and functional magnetic resonance imaging (fMRI) [2–10]. Subsequently, this stimulus rate (or temporal frequency) dependency was generalized to other types of visual stimuli. In a follow-up study, Fox and Raichle [11] found that changes in rCBF as a function of stimulus rate up to ~8 Hz were nearly identical for both patterned-flash and contrast reversing checkerboards. As a result, it is often implicitly assumed in functional imaging studies that human visual cortex is optimally activated at a temporal frequency of 8 Hz.

While the stimulus rate dependency with a patterned-flash stimulus can be accounted for by a confound between the average luminance and the stimulus rate, a temporal frequency dependency is surprising for the contrast reversing checkerboard because the average luminance is constant across different reversal frequencies. That hemodynamic responses change as a function of reversal frequency of contrast reversing checkerboard is at odds with neurophysiological studies in cat [12–15], ferret [16,17], and macaque monkey [18,19] indicating that primary visual cortex (V1) contains two major populations of neurons (low-pass and band-pass neurons). How these two groups of neurons are distributed in the cortex is still under debate [20], with one claiming that they form fine (~0.5 mm in cat V1) columnar structures [21] while the other suggesting that no clear spatial cut between them [19]. In either case, neuroimaging studies would be expected to sample the activity of both populations (at least for low spatial resolution studies) and therefore show similar responses to both high and low temporal frequencies.

In fact, similar activations to low and high temporal contrast reversing frequencies have been observed in many other neuroimaging studies [22–37] using either gratings or checkerboard. Because the measurement of brain activation was made mainly by a block-design and was averaged over a certain period of time in these studies, it is still unclear whether there are any differences in the transient aspects of brain activation across different stimulus frequencies. We have addressed this issue by using fMRI with an event-related paradigm and found that there is no clear stimulate rate dependency in the transient characteristics of blood oxygenation level dependent (BOLD) signal on contrast reversing frequencies.
Materials and Methods

Subjects

Eight subjects (5 males and 3 females, ages 28–35) with no past history of psychiatric or neurological diseases participated in the experiments, which were approved in advance by the RIKEN Functional MRI Safety and Ethics Committee. The subjects were selected after morphological screening of high-resolution three-dimensional anatomical MR images (1×1×1 mm³). To cover targeted cortical area by limited number of slices, the calcarine sulcus of the selected subjects follows a relatively straight course and the cortex around their calcarine V1 is relatively flat and less sulcated in at least one hemisphere, which was then used as the target subsequent study. All subjects had normal or corrected-to-normal vision and gave their written informed consent before each experiment.

Imaging hardware

All experiments were conducted on a Varian Unity Inova 4 T whole-body MRI system (Varian NMR Instruments, Palo Alto, CA) equipped with a Magnex head gradient system (Magnex Scientific Ltd., Abingdon, UK). High-resolution three-dimensional T1-weighted anatomical MR images were acquired with a birdcage radio-frequency (RF) coil. A 5-inch transmit/receive butterfly quadrature RF surface coil was used to acquire functional and co-registered anatomical images in the functional experiments. The surface coil was mounted on a bakelite support frame attached to the patient table.

During the experiment, the subject was made to lie supine on the patient table and to rest the back of the head on the surface coil. Head motion was restricted using a bite-bar as well as padding spongy rubber around the subject’s head. The subject’s heartbeat was monitored with a pulse oximeter, and respiration was monitored with a pressure sensor placed on the abdominal region. Both signals were recorded along with the timing of RF pulses for later correction of physiological fluctuations.

Visual stimulation

The presentation of visual stimuli was controlled by a custom-built software package run on a PowerMac G4 computer. Stimuli were delivered to the subject’s eyes via an optic fiber goggle system (Avotec, Inc., Jensen Beach, FL) that subtended 25°×19° of visual angle. The image on the goggles had a resolution of 800×600 pixels and a refresh rate of 60 Hz. The subject adjusted two red/green pairs of spectacles, each with a pair of filters, to align the display with his or her visual field. The calibration of the stereo optical recording was performed for each subject.

Since our custom-built program presented the stimulus exactly at the beginning of each frame (with a refresh rate of 60 Hz), the highest achievable reversal frequency with precise timing using our stimulation presentation device was 15 Hz (containing 30 contrast transitions). Second, because signal fluctuations caused by cardiac pulsations (~1 Hz) were removed in the data post-processing, a low frequency of 0.75 Hz, instead of 1 Hz, was used. Third, for comparison with previous studies, a very low frequency was introduced. Because the brief stimulus presentation time (3 s) as required by the event-related design, the lowest contrast reversal frequency can be achieved is 0.17 Hz, corresponding to the appearance and disappearance of the checkerboard. Finally, it should be pointed out that the liquid crystal display (LCD) projector or goggle, compared with conventional cathode ray tube (CRT) monitors, has relative slow rise- and fall-times [38]. As a result, the luminance contrast of our display (LCD goggle) decreases as temporal frequency increases. The rise and fall-times of our display were measured and this confounding tendency was corrected accordingly [28], and a fixed contrast level (33%) was used for all the frequencies.

We also made efforts to restrict our analyses to the gray matter of a well-defined, circumscribed portion of V1. In all six scans, the checkerboard was confined to a quadrant of the visual field. This stimulus configuration, combined with the meridian mapping that demarcated V1/V2 borders of individual subjects (see below), allowed us to localize activation in the expected portion of V1.

Throughout the experiment, the subject was instructed to fixate on the central fixation cross (in unsaturated red). To help maintain fixation and the level of arousal, the subject performed a detection task on the fixation cross. Every 2–5 s, the luminance of the fixation cross was increased for 500 ms, and the subject was to report that event with a button press. Data would be discarded once the subject missed to press the button more than 5%. We nevertheless monitored the subject’s eye position using iView (SensoMotoric Instruments, Boston, MA) and the eye tracker built into the goggle system. Offline analysis confirmed that the subject performed the task correctly more than 99% and maintained fixation throughout the experiment and therefore no data were discarded.

Imaging parameters and experimental paradigms

In addition to screening the morphology around the calcarine sulcus, we also determined the V1/V2 borders of individual subjects using a standard method for mapping the representation of the vertical meridian [39,40]. The V1/V2 borders were determined in a separate meridian and retinotopic mapping experiment. In the meridian and retinotopic experiment, there were two blocked-design runs in which a wedge (15°) of checkerboard was presented 6 times at vertical or horizontal meridian with 30 s on and 30 s off periods and six traveling wave design runs in which a wedge (90°) of checkerboard with rotating speed in 24 s/cycle was presented 12 times. Based on the mapped V1/V2 borders, in the main functional experiment, the slices were placed parallel to the calcarine sulcus on the target hemisphere to completely contain the dorsal V1, with one optimally covering the
dorsal bank of the calcarine sulcus. Prior to each functional experiment, T1-weighted anatomical images at the same slice positions as functional images were acquired with a four-segment inversion recovery FLASH pulse sequence (slice thickness = 3 mm; FOV = 24×24 cm²; in-plane resolution = 0.94×0.94 mm²; volume TR = 1 s; TE = 10 ms; number of averaging = 2).

In the functional experiment, five contiguous slices (slice thickness = 3 mm; FOV = 24×24 cm²; in-plane resolution = 3.75×3.75 mm²) were collected. Functional images were acquired with a two-segment centric-ordered EPI pulse sequence (volume TR = 0.5 s; TE = 25 ms; average FA = 45°). Twelve trials, each consisting of a 3 s stimulation period with a checkerboard reversing at one of the six frequencies and a 30 s baseline period with a homogeneous gray screen, were repeated. Before the first trial, a 33 s baseline was inserted. In total, 858 volumes over 429 seconds were acquired in a scan. Six scans, each with a reversal frequency chosen in pseudo-randomized order, were collected from each subject. In addition, before the six event-related scans, a region of interest (ROI) scan was collected with identical imaging parameters and slice prescription on each subject. During the scan, in which 360 volumes over 180 seconds were collected, the subject viewed three cycles of a 2 Hz checkerboard and a gray scan. The ROI identified in this scan was used for quantifying BOLD responses in subsequent event-related scans.

Data processing and analysis

In all functional experiments, longitudinal magnetization was allowed to reach steady state before EPI images were collected. EPI distortions were minimized using a reference volume (without phase encoding) acquired at the beginning of each experiment [41]. The first echo in each segment was a navigator echo, which was used to correct inter-segment phase and amplitude variations [42]. After EPI images were reconstructed, cardiac and respiratory fluctuations were removed from time series images using a retrospective estimation and correction method which employed pulsation and respiration data recorded during image acquisition [43]. Two-dimensional motion correction was also applied [44]. Both corrections were applied in k-space. A high-pass filter was used to suppress baseline signal drifts. This filter was set to have a 3dB drop-off cutoff frequency of 0.00085 cycles/sec but to retain the DC level. No other spatial or temporal smoothing was applied. Statistical analyses were performed using the software package STIMULATE developed at the University of Minnesota [J.P.Strupp, NeuroImage 3, S607, 1996]. Activated voxels (displaying positive signal changes) were identified by the Student’s t test comparing images acquired during stimulation periods and those acquired during control periods on a voxel-by-voxel basis. To account for the hemodynamic delay, the first image acquired during each period in the block-design experiments was excluded.

Due to large geometrical variations around V1 between the subjects, quantitative analyses for data from the experiments were performed on a single-subject basis. In the event-related experiment, a time course (3 s checkerboard stimulation and 30 s gray screen for each trial, 12 repetitions) for each of the six frequencies was obtained by averaging BOLD responses from all the voxels within the ROI defined in the preceding ROI scan. The baseline (0% BOLD response) was determined from the 33 s control period before the first trial. To quantify the transient BOLD response to each frequency, we first fit each time course using a gamma function:

\[
A \left( \frac{t - \text{lag}}{\tau(n-1)} \right)^{n} e^{-\left( \frac{t - \text{lag}}{\tau(n-1)} \right) \frac{1}{\tau(n-1)}}
\]

where A is the amplitude; n is a shape parameter that was allowed to take on integer values from 4 to 6; and t roughly corresponds to the width of the response and was allowed to take on values from 0.5 to 2 seconds. The lag value controls when the gamma function begins relative to stimulus onset and take on values ranging from 0 to 4 seconds. When t-lag was negative, the function was set to 0. A difference of gamma functions, in which the second gamma function was allowed to have t values between 0.5 and 4 s and a lag 2–8 s, was used to capture the delayed negative response of the hemodynamic function. Levenberg-Marquardt optimization was used to find the parameters of the two gamma functions that minimized the mean squared difference between the estimated hemodynamic response and the actual time course. From the estimated hemodynamic response for each of the six frequencies, we extracted four parameters, that is, the maximum BOLD response, full-width at half-maximum (FWHM) of BOLD response curve, integrated area under the BOLD response curve (for this calculation, we only used positive responses; post-stimulus undershoots were not used) and time to reach maximum BOLD response, and compared them across the eight subjects. All data analyses mentioned above and further statistical analyses were performed using in-house developed and built-in functions in Matlab (The MathWorks, Inc., Natick, MA).

Control experiment

In addition to the main experiment described above, a control experiment was run with a different group of subject, different settings of imaging hardware, and a different experimental paradigm. Specifically, four subjects (3 males and 1 female, ages 19–28) with no past history of psychiatric or neurological diseases participated in the experiments, which were approved by the Tsinghua Functional MRI Safety and Ethics Committee. The experiment was conducted on a Phillips Achieva 3 T MRI system at the Center for Biomedical Imaging Research, Tsinghua University. High-resolution three-dimensional T1-weighted anatomical MR images, functional and co-registered anatomical images were acquired with a 32-channel RF coil. Visual stimulation was delivered to the subject’s eyes via a mirror and a screen system (Invivo, Gainesville, FL). Same contrast level across different temporal frequency was achieved using a psychophysical procedure [45] based on the prior measurements of rising and falling characteristic of LCD monitors [28]. In the control experiment, there were three long runs. In each long run, twenty-four trials were conducted and one trial consisted of a 5 s stimulation period and a 30 s baseline. Six reversing frequencies were repeated 4 times in random order in the long run. Before the first trial, a 33 s baseline was inserted. To avoid selection bias, 6 Hz instead of 2 Hz was used in the localizer experiment, which was a different frequency from that used in the main experiment and in the middle range of all the six temporal frequencies. In the control experiment, seven contiguous slices (slice thickness = 3 mm; FOV = 22×22 cm²; in-plane resolution = 2.75×2.75 mm²) were collected. Functional images were acquired with a single-shot EPI pulse sequence (TR = 0.5 s; TE = 35 ms; FA = 90°). Functional data were pre-processed (i.e., head motion correction and baseline drifts suppression) using AFNI (http://afni.nimh.nih.gov/afni/).
Results

In the fMRI experiment, we investigated several transient aspects of BOLD responses to the reversal frequency by quantifying four parameters, namely, the maximum BOLD response, full-width at half-maximum (FWHM) of BOLD response curve, integrated area under the BOLD response curve and time to reach maximum BOLD response in a circumscribed region of V1 following a brief exposure (3 s) to a black/white checkerboard whose contrast was reversed at 0.17, 0.75, 2, 4, 8, or 15 Hz. Each frequency in the experiment was studied in a separate scan.

Before the main event-related experiment, a functional localizer experiment was conducted with a block-design for each subject using the frequency of 2 Hz that is in the middle of used frequencies. Based on the statistical t-map \((p < 0.01)\) obtained in the functional localizer experiment, a region of interest (ROI), which extended across 2 to 3 slices, depending on the morphology of V1 of individual subjects, was determined. The same ROI was then used for quantitative analysis for all the frequencies in the event-related experiments. Overall, about 8 to 15 activated voxels were found for each subject by this procedure.

To quantify the dependency of the transient BOLD response on reversal frequency, we obtained the estimated hemodynamic response by averaging the time course of all the voxels inside ROI for each of the six frequencies (Figure 1), from which we extracted four parameters, the maximum BOLD response, full-width at half-maximum (FWHM) of BOLD response curve, integrated area under the BOLD response curve and time to reach maximum BOLD response (Table 1). A one-way repeated-measures ANOVA revealed no frequency effect \((p = 0.699,\) maximum BOLD response; \(p = 0.213,\) FWHM of BOLD response curve; \(p = 0.877,\) integrated area under the BOLD response curve; and \(p = 0.429,\) time to reach maximum BOLD response). Please note that almost same amplitudes of BOLD response were already observed at low frequencies as that at high frequencies (e.g., 0.17 Hz elicited \(\sim 93\%\) of response by 8 Hz).

Discussion

Using a spatially confined black/white contrast reversing checkerboard that activated a circumscribed portion of early visual cortex, we found that the transient characteristics of BOLD response in human V1 depended very little on the reversal frequency. The results from the present and our former study [28] appear to be at odds with the prevailing thought that hemodynamic responses depend on the temporal frequency of visual stimuli. However, a careful review of the literature reveals that the discrepancy stems largely from the types of stimuli used in various experiments.

In their seminal PET study, Fox and Raichle [1] first demonstrated stimulus rate dependency of rCBF changes in human visual cortex (around V1) using patterned-flash stimuli. Specifically, they showed that rCBF change was a linear function of stimulus repetition rate between 0 and 8 Hz, peaked at \(\sim 8\) Hz, and then declined as the rate further increased. Similar observations were also made subsequently in a number of studies using both PET and fMRI [2–10]. The stimuli used in all these studies, including the original one by Fox and Raichle, were...
flashing strobe lamps, checkerboards or light-emitting diodes (LEDs) with a flash duration ranging from 50 µs to 5 ms. The stimulation rate (or temporal frequency), therefore, was the number of flashes per second, and the luminance averaged over the entire stimulation period was a function of stimulus rate. In this sense, the observed rCBF change or the BOLD response following a contrast reversing checkerboard need to be further investigated.

A contrast reversing checkerboard (contrast reversals between black and white checkers at equal intervals), however, is very different from patterned-flash stimuli (flashing checkers or LEDs) because the average luminance during the stimulation period is constant across different reversal frequencies. That rCBF changes as a function of stimulus rate were nearly identical for both patterned-flash stimuli and contrast reversing checkerboard was claimed in a follow-up study by Fox and Raichle [11]. However, the data point for the contrast reversing checkerboard at 0 Hz in that study was from a scan without visual stimulation, instead of a scan using a stationary checkerboard that would have maintained the same stimulus luminance as with other frequencies. Naturally, this 0 Hz condition defined for the contrast reversing checkerboard, which physically was identical to the 0 Hz for the patterned-flash, also produced an rCBF change near the baseline. Thus, the apparent similarity of rCBF changes between 0 and 7.8 Hz for the two types of stimuli was most likely due to this inappropriate use of stimulus for the 0 Hz contrast reversing checkerboard. Indeed, inspecting their figure reveals that the rCBF change using the contrast reversing checkerboard at 1 Hz (the lowest frequency actually tested) was ~65% of the maximal response observed at 7.8 Hz, whereas the change using the patterned-flash at 1 Hz was only ~17% of the maximum.

Despite the fact that a linear relationship between rCBF or BOLD response and the reversal frequency of contrast reversing checkerboard has never been established, such a frequency dependency is often implicitly assumed in various neuroimaging studies when contrast reversing checkerboards are used. The results from several other studies, however, have provided very little support for such a relationship. In a parametric study using $^{13}$CO$_2$ PET, Law and colleagues [24] measured rCBF changes following a contrast reversing checkerboard stimulation with 10 reversal frequencies from 0.03 to 30 Hz, and found that the maximal response, observed between 6 and 15 Hz across subjects, displayed only a small increase (8%) from that observed at 0.03 Hz. Similarly, in fMRI experiments, Kastner and colleagues [22] reported that the BOLD response to 0.5 Hz was ~84% of the maximal response elicited by 20 Hz; Muthukumaraswamy and colleagues [33] showed that BOLD response to 0 Hz was ~65% of the maximal response elicited by 16 Hz under two different spatial frequencies (3 cphd and 0.5 cphd). Moradi and colleagues also made the observation that BOLD responses to reversal frequencies of 0.71 and 3 Hz were qualitatively similar [25, see their Figures 8B and 8C]. As different protocols were used in previous studies and there are essential difference between rCBF and BOLD signal, how rCBF or BOLD response in human V1 changes with the reversal frequency of contrast reversing checkerboard need to be further investigated.

Finally, it should be pointed out that the observed hemodynamic activity in the current study is a combination of the impulse

| Frequency (Hz) | Max value (%) | FWHM (s) | Area (% × s) | TTM (s) |
|---------------|---------------|----------|--------------|---------|
| 0.17          | 2.26±0.19     | 5.00±0.30| 25.54±3.22   | 5.66±0.16|
| 0.75          | 2.39±0.21     | 5.44±0.52| 30.08±3.79   | 5.38±0.23|
| 2             | 2.31±0.32     | 5.42±0.27| 29.24±4.54   | 5.75±0.23|
| 4             | 2.57±0.22     | 4.83±0.28| 27.82±3.38   | 5.71±0.33|
| 8             | 2.45±0.23     | 5.15±0.27| 26.72±3.29   | 5.99±0.27|
| 15            | 2.33±0.23     | 5.76±0.24| 27.73±2.11   | 5.59±0.28|

Data were obtained from eight subjects (five right and three left V1). All values are given in mean ± SEM. Frequency: contrast reversal frequency (Hz); Max value: maximum BOLD response in percent (%); FWHM: full-width at half-maximum of BOLD response curve in second (s); Area: integrated area under the BOLD response curve in percent multiplied by second (% × s); TTM: time to reach maximum BOLD response in second (s).

doi:10.1371/journal.pone.0099547.t002

Table 2. Parameters extracted from estimated hemodynamic responses to six temporal frequencies in an event-related experiment from another four subjects (three right and one left V1).

| Frequency (Hz) | Max value (%) | FWHM (s) | Area (% × s) | TTM (s) |
|---------------|---------------|----------|--------------|---------|
| 0.17          | 2.87±0.19     | 5.60±0.42| 34.21±3.81   | 5.56±0.40|
| 0.75          | 3.08±0.43     | 6.16±0.26| 40.99±6.41   | 4.99±0.52|
| 2             | 4.08±0.54     | 5.99±0.16| 56.93±11.23  | 5.30±0.42|
| 4             | 3.64±0.52     | 6.02±0.42| 45.46±6.22   | 5.54±0.53|
| 8             | 4.23±0.62     | 6.19±0.38| 55.15±8.80   | 5.72±0.32|
| 15            | 3.31±0.35     | 5.72±0.45| 40.31±2.30   | 5.85±0.43|

See conventions in Table 1.

doi:10.1371/journal.pone.0099547.t002
response to the sudden presentation of visual stimulation that included many visual features (e.g., spatial frequency, orientation, contrast, etc.) and the neural response to certain temporal frequency during the brief stimulation period (i.e., 3 seconds). Given that the difference of BOLD responses between different patterns and flash frequencies could be detected using a brief stimulation (i.e., 1 second) in a former study [5], we may claim that such difference is not existed or at least the tendency is not evident enough with the current experimental settings. Furthermore, even we could extend the duration of visual stimulation, another confound factor, that is the visual adaptation, which typically results in reduction in neuronal response over prolonged exposure to a visual stimulus, will be introduced and this extension will lead to a block-design experiment eventually. Taken the results using the block-design in our former study [28] and the results using slow event-related design in the present study, we conclude that there is no clear stimulate rate dependency in the transient characteristics of BOLD signal on contrast reversing frequencies.

Author Contributions
Conceived and designed the experiments: PS JG SF. Performed the experiments: PS JG SG JH. Analyzed the data: PS SG. Wrote the paper: PS.

References
1. Fox PT, Raichle ME (1984) Stimulus rate dependence of regional cerebral blood flow in human striate cortex, demonstrated by positron emission tomography. J Neurophysiol 51: 1199–1120.
2. Hagenbeck RE, Romhouts SA, van Dijk BW, Barkhof F (2002) Determination of individual stimulus—response curves in the visual cortex. Hum Brain Mapp 17: 244–256.
3. Kwong KK, Belliveau JW, Chesler DA, Goldberg IE, Weisskoff RM, et al. (1992) Dynamic magnetic resonance imaging of human brain activity during primary sensory stimulation. Proc Natl Acad Sci U S A 89: 5673–5679.
4. Menis MJ, Alexander GE, Grady CL, Horvitz B, Krausnius J, et al. (1997) Frequency- and pattern-specific neuronal responses to flicker stimulation in human visual cortex. J Neurophysiol 78: 2405–2427.
5. Ozus B, Liu HL, Chen I, Iyer MB, Fox PT, et al. (2001) Rate dependence of human visual cortical response due to brief stimulation: an event-related fMRI study. Magn Reson Imaging 19: 21–25.
6. Pastor MA, Aridiya J, Arbizu J, Valencia M, Masdeu JG (2003) Human cerebral activation during steady-state visual-evoked responses. J Neurosci 23: 11621–11627.
7. Singh M, Kim S, Kim TS (2005) Correlation between BOLD-fMRI and EEG signal changes in response to visual stimulus frequency in humans. Mag Reson Med 49: 108–114.
8. Thomas CG, Menon RS (1998) Amplitude response and stimulus presentation frequency response of human primary visual cortex using BOLD EPI at 4 T. Magn Reson Med 40: 203–209.
9. Zhu RH, Kim SG, Andersen P, Ogawa S, Ugurbil K, et al. (1998) Simultaneous oxygenation and perfusion imaging study of functional activity in primary visual cortex at different visual stimulation frequency: quantitative correlation between BOLD and CBF changes. Magn Reson Med 40: 703–711.
10. Parks LM, Fries P, Kersken CM, Norris DG (2004) Reduced BOLD response to periodic visual stimulation. Neuroimage 21: 236–243.
11. Fox PT, Raichle ME (1985) Stimulus rate determines regional brain blood flow in striate cortex. Ann Neurol 17: 303–305.
12. Ikeda H, Wright MJ (1975) Spatial and temporal properties of ‘sustained’ and ‘transient’ neurones in area 17 of the cat’s visual cortex. Exp Brain Res 22: 363–383.
13. Movshon JA, Thompson ID, Tolhurst DJ (1978) Spatial and temporal contrast sensitivity of neurones in areas 17 and 18 of the cat’s visual cortex. J Physiol 263: 101–120.
14. Reul RG, Victor JD, Shapley RM (1992) Broadband temporal stimuli decrease the integration time of neurons in cat striate cortex. Vis Neurosci 9: 39–45.
15. Saul AB, Humphrey AL (1992) Temporal-frequency tuning of direction selectivity in cat visual cortex. Vis Neurosci 5: 365–372.
16. Alitto HJ, Usrey WM (2004) Influence of contrast on orientation and temporal tuning in ferret primary visual cortex. J Neurophysiol 91: 2797–2808.
17. Moore Iv BD, Alitto HJ, Usrey WM (2005) Orientation tuning, but not direction selectivity, is invariant to temporal frequency in primary visual cortex. J Neurophysiol.
18. Foster KH, Gaska JP, Nagler M, Pollen DA (1985) Spatial and temporal frequency selectivity of neurones in visual cortical areas V1 and V2 of the macaque monkey. J Physiol 365: 331–363.
19. Hawken MJ, Shapley RM, Goss D (1996) Temporal-frequency selectivity in primary visual cortex. Vis Neurosci 13: 477–492.
20. Isa NP, Rosenberg A, Hasson TR (2000) Models and measurements of functional maps in V1. J Neurophysiol 99: 2745–2754.
21. Shoham D, Hubener M, Schulte S, Grünwald A, Bonhoeffer T (1997) Spatio-temporal frequency domains and their relation to cytochrome oxidase staining in cat visual cortex. Nature 385: 529–533.
22. Kastner S, O’Connor DH, Fukui MM, Fehd HM, Herwig U, et al. (2004) Functional imaging of the human lateral geniculate nucleus and pulvinar. J Neurophysiol 91: 380–400.
23. Kaufmann C, Ehbel GK, Goos C, Patz B, Auer DP (2001) Frequency dependent and gender effects in visual cortical regions involved in temporal frequency dependent pattern processing. Hum Brain Mapp 14: 28–38.
24. Law I, Jensen M, Holm S, Nickles RJ, Paulson OB (2001) Using (10)CO2 for single subject characterization of the stimulus frequency dependence in visual cortex: a novel positron emission tomography tracer for human brain mapping. J Cereb Blood Flow Metab 21: 1003–1012.
25. Moradi F, Liu LC, Cheng K, Waggoner RA, Tanaka K, et al. (2003) Consistent and precise localization of brain activity in human primary visual cortex by MEG and fMRI. Neuroimage 18: 595–609.
26. Singh KD, Smith AT, Greenlee MW (2000) Spatiotemporal frequency and direction sensitivities of human visual areas measured using fMRI. Neuroimage 12: 530–564.
27. Toronov V, Zhang X, Webb AG (2006) A spatial and temporal comparison of hemodynamic signals measured using optical and functional magnetic resonance imaging during activation in the human primary visual cortex. Neuroimage.
28. Sun P, Ueno K, Waggoner RA, Gardner HL, Tanaka K, et al. (2007) A temporal frequency-dependent functional architecture in human V1 revealed by high-resolution fMRI. Nat Neurosci 10: 1404–1408.
29. Mirzajan A, Riahi-Alam N, Oghabian MA, Saheri H, Firooznia K (2007) Spatial frequency modulates cortical visual response to temporal frequency variation of visual stimuli: an fMRI study. Physiol Meas 28: 547–554.
30. Sinivasan R, Fornari E, Iyengar MG, Meuli R, Maeder P (2007) fMRI responses in medial frontal cortex that depend on the temporal frequency of visual input. Exp Brain Res 180: 677–691.
31. Lin AL, Fox PT, Hartlits DJ, Duong TQ, Gao JH (2010) Nonlinear coupling between cerebral blood flow, oxygen consumption, and ATP production in human visual cortex. Proc Natl Acad Sci U S A 107: 8446–8451.
32. Fawcett IP, Barnes GR, Hillebrand A, Singh KD (2004) The temporal frequency tuning of human visual cortex investigated using synthetic aperture magnetoencephalography. Neuroimage 21: 1542–1553.
33. Muthukumaraswamy SD, Singh KD (2008) Spatiotemporal frequency tuning of BOLD and gamma band MEG responses compared in primary visual cortex. Neuroimage 40: 1532–1560.
34. Mullen KT, Thompson B, Hess RF (2010) Responses of the human visual cortex and LGN to achromatic and chromatic temporal modulations: an fMRI study. J Vis 10: 13.
35. D’Souza DV, Auer T, Strasburger H, Frahm J, Lee BB (2011) Temporal frequency and chromatic processing in humans: an fMRI study of the cortical visual areas. J Vis 11: 1.
36. Emin UE, BakrCattaro Z, Ozturk C, Ademoglu A, Demiralp T (2008) Changes in BOLD transients with visual stimuli across 1–44 Hz. Neurosci Lett 436: 185–188.
37. Yeni CC, Fukuda M, Kim SG (2011) BOLD responses to different temporal frequency stimuli in the lateral geniculate nucleus and visual cortex: insights into the neural basis of fMRI. Neuroimage 58: 82–90.
38. Norcia AM, McKee SP, Boneh Y, Petett MW (2005) Suppression of monocular visual direction under fused binocular stimulation: evoked potential measurements. J Vis 5: 34–44.
39. Torstel RB, Reppas JB, Kwong KK, Malach R, Born RT, et al. (1995) Functional analysis of human MT and related visual cortical areas using magnetic resonance imaging. J Neurosci 15: 5215–5230.
40. Engel SA, Glover GH, Wandell BA (1997) Reticular organization in human visual cortex and the spatial precision of functional MRI. Cereb Cortex 7: 181–192.
41. Bruder H, Fischer H, Reinfelder HE, Schmitt F (1992) Image reconstruction for echo planar imaging with noiseequidistant kspace sampling. Magn Reson Med 23: 311–323.
42. Kim SG, Hu X, Adriany G, Ugurbil K (1996) Fast interleaved echo-planar imaging with navigator: high resolution anatomic and functional images at 5 Tesla. Magn Reson Med 35: 895–902.
43. Hu X, Le TH, Parrish T, Erhard P (1995) Retrospective estimation and correction of physiological fluctuation in functional MRI. Magn Reson Med 34: 201–212.
44. Maas LC, Frederick BD, Renshaw PF (1997) Decoupled automated rotational and translational registration for functional MRI time series data: the DART registration algorithm. Magn Reson Med 37: 131–139.
45. To L, Woods RL, Goldstein RB, Peli E (2013) Psychophysical contrast calibration. Vision Res 90: 15–24.
46. Bouvier SE, Engel SA (2011) Delayed effects of attention in visual cortex as measured with fMRI. Neuroimage 57: 1177–1183.
47. Sirotin YB, Das A (2009) Anticipatory haemodynamic signals in sensory cortex not predicted by local neuronal activity. Nature 457: 475–479.