Statistical power: implications for planning MEG studies

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Highlights:

- Adequate sample size (number of subjects and trials) is key to robust neuroscience
- We simulated evoked MEG experiments and examined sensor-level detectability
- Statistical power varied by distance, orientation & between-subject variability
- Source detectability at sensor-level should be considered in network-based studies
- Select sample size for MEG source in the network that has lowest statistical power

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Abstract

Statistical power is key for robust, replicable science. Here, we systematically explored how numbers of trials and subjects affect statistical power in MEG sensor-level data. More specifically, we simulated "experiments" using the MEG resting-state dataset of the Human Connectome Project (HCP). We divided the data in two conditions, injected a dipolar source at a known anatomical location in the "signal condition", but not in the "noise condition", and detected significant differences at sensor level with classical paired t-tests across subjects.

The group-level detectability of these simulated effects varied drastically with their anatomical origin. We thus examined in detail which spatial properties of the sources affected detectability, looking specifically at the distance and orientation of the source with respect to the closest sensor, *and* at the variability of these parameters across subjects. In line with previous single-subject studies, we found that the most detectable effects originate from source positions that are closest to the sensors and oriented tangentially with respect to the sensor array. In addition, cross-subject variability in orientation also affected group-level detectability, boosting detection in regions where this variability was small and hindering detection in regions where it was large. Incidentally, we observed a considerable covariation of source position, orientation, and their cross-subject variability in individual brain anatomical space, making it difficult to assess the impact of each of these variables independently of one another. Therefore, we performed an additional set of simulations with the same simulation strategy as before, but this time, controlling spatial properties independently of individual anatomy. This allowed us to disentangle the effects of position, orientation, and cross-subject variability of these properties on detectability, confirming the strong impact of distance and orientation. Moreover, it showed that orientation variability across subjects affects detectability, whereas position variability does not.

Importantly, our study indicates that strict unequivocal recommendations as to the ideal number of trials and subjects for any experiment cannot be realistically provided for neurophysiological studies. Rather, it highlights the importance of considering *the spatial constraints* underlying expected sources of activity while planning experiments. This is particularly relevant for performing network level analyses in cognitive neuroscience studies, where nodes in a network cannot be considered equally detectable. Finally, we note that this variation in detectability may also partly account for the over-representation of some fields of study over others in the MEG domain (e.g. sensorimotor and auditory studies).

Keywords:
statistical power, MEG, source modelling, simulation, distance, orientation

Introduction

Adequate statistical power is a requisite of robust, replicable science. An important variable affecting statistical power is sample size, and it has been shown that previous studies have been undermined by too small sample sizes (Button et al., 2013; Szucs & Ioannidis, 2017). Additionally, suboptimal scientific practices such as experimental designs and analysis approaches inappropriate to answering the posed scientific question have accentuated this problem (Gelman & Loken, 2013; Kerr, 1998; Kriegeskorte, Simmons, Bellgowan, & Baker, 2009; Luck & Gaspelin, 2017). Overall, data gathering and analytical procedures that once were widely used are now identified as being flawed, while reporting procedures are codified (Keil et al., 2014) and their endorsement is critically assessed (Clayson, Carbine, Baldwin, & Larson, 2019; Larson & Carbine, 2017). Emphasis is being put on the one hand on improving the robustness of statistical inference (Groppe, 2017; Kappenman & Keil, 2017; Kilner, 2013; Luck & Gaspelin, 2017; Simonsohn, Nelson, & Simmons, 2013), as well as, on the second hand, on the design, preparation, and documentation of carefully planned experiments (Chambers, Dienes, McIntosh, Rotshtein, & Willmes, 2015; Foster & Deardorff, 2017; Luck, 2005). The current study is at the crossroad of these two trends, aiming to aid researchers in cognitive, affective, and social neuroscience make principled decisions on how many trials and subjects to include in their experiments in order to achieve adequate levels of statistical power. More precisely, we highlight some important variables that one should pay attention to when considering how many repetitions of experimental conditions and how many subjects one should test to achieve robust statistical inference in an MEG experiment.

The question of knowing how many trials and subjects to include in an MEG or EEG experiment has to date been mostly a matter of “know-how” or “rules of thumb” (Gross et al., 2013; Luck, 2012). Indeed, this topic has been discussed for decades without reaching any

definitive conclusion (Duncan et al., 2009; Hari et al., 2018; Kane et al., 2017; Keil et al., 2014; Picton et al., 2000; Pivik et al., 1993). However, the above-mentioned concerns about power and reproducibility call for a systematic evaluation of variations in statistical power. Which variables critically affect statistical power? Understanding these, how can the necessary (and sufficient) number of trials be planned in advance? Recently, Boudewyn and colleagues (2018) took a first step at answering this question in a principled manner. They used EEG recordings from 40 participants to examine how the number of observations included in their analysis affected the probability of finding a significant effect in Event-Related Potential (ERP) measures. As expected, large effects (e.g., an error-related negativity, producing a 5-15 μV difference wave in the EEG) were detected at sensor level with fewer trials and participants than smaller effects were (e.g., a finer ~ 1 μV amplitude modulation in the lateralized readiness potential). We believe Boudewyn et al. (2018) to be the first EEG study that directly related the number of observations (trials and subjects) to statistical power. In another recent study, Baker and colleagues (2019) took this approach one step further, by acknowledging the difference between within-sample (i.e. inter-trial) and between-sample (i.e. between subjects) variability in a number of assessment modalities, including MEG and EEG. They introduced so-called "power contours"—plots that depict statistical power as a joint function of the number of trials and the number of subjects. These power contours reveal the level of statistical power reached for given trial and subject numbers. We illustrate the findings of the present study using this power contour plotting approach.

Here, we used MEG data from a large sample of subjects to reliably examine the behavior of statistical power measures. We simulated “experiments” with the open resting-state MEG dataset of the Human Connectome Project (HCP, Larson-Prior et al., 2013), varying the numbers of trials and subjects to assess the effects of within- and between-subject sample sizes, respectively. This allowed us to simulate power in the context of real physiological

background activity. We simulated effects by injecting dipolar sources of fixed amplitude at known anatomical locations in one half the data and not in the other half (Figure 1). We then detected the effects observed at sensor level with classical paired t-tests across subjects. In an initial exploration, we used detailed individual source models and observed that detectability varies drastically according to the anatomical location of the sources. Accordingly, we explored the spatial properties of the sources across the brain, namely their distance and orientation with respect to the closest sensor, and the cross-subject variability of these parameters. For this, we undertook two types of simulations. First, we examined changes of detectability across different anatomical areas, simulating sources in the real anatomy of individual subjects in our HCP dataset. In doing so, we observed that the measured properties were impossible to properly disentangle due to anatomical constraints. Therefore, we ran a second set of simulations, where we imposed spatial properties of the sources independently of individual brain anatomy while using the same simulation strategy as before. In this way, we could assess separately the effect of each spatial property of the sources (position, orientation, and their cross-subject variability) on detectability.

Material & Methods

Software

All data were processed using MATLAB (The Mathworks, Natick, MA, USA; Version 2018a) and the FieldTrip toolbox (Oostenveld, Fries, Maris, & Schoffelen, 2011, github release of October 25 2019). All analysis and visualization scripts and toolboxes, including the FieldTrip toolbox, and megconnectome toolbox used, are available online at https://gitlab.com/icm-institute/meg/solid_meg.

Data

Input data

We used MEG data from 89 subjects from the Human Connectome Project (HCP) open data repository, described in detail in Larson-Prior et al. (2013). Of the 95 original subjects in the latest release (V3.0) of the MEG HCP data, 6 subjects with missing data in one or another of the components described below were discarded, resulting in the final number of 89 subjects for our analyses.

A whole series of preprocessing steps are already performed on the HCP data. For complete details, see the descriptions for the pre-processing pipelines for MEG (Larson-Prior et al., 2013), structural MRI (Glasser et al., 2013), as well as the online resources available at https://www.humanconnectome.org/software. We henceforth refer to these two preprocessing pipelines as the *HCP-MEG pipeline* and the *HCP-structural pipeline*. Note that these pipelines are part of different releases of the HCP data that were downloaded separately from https://db.humanconnectome.org. Subject identifier codes being unique across these releases, we used them to merge the data from both pipelines.

Briefly, the MEG data were acquired at 2043.5101 Hz (bandpass filtering: DC-400 Hz) with a whole-head 248 magnetometer device (MAGNES 3600, 4D Neuroimaging, San Diego, CA). The HCP includes data from several activation tasks and three 6-minute resting state periods (eyes open, fixating). We used the latter dataset in this study. MEG fiducials were obtained with a Polhemus system for coregistration to structural MRI data. Each subject was scanned with a high-resolution structural MRI on a 3T Siemens Skyra MRI scanner (Van Essen et al., 2012). Within the HCP-MEG pipeline, the structural MRI was used to create a single-shell volume conduction model using FieldTrip. Furthermore, to define anatomical labels in our study, we

used the output of the HCP-structural pipeline, where a high-definition segmentation and anatomical labelling (Destrieux Atlas) of the cortical mantle was performed using Freesurfer (Version 5.2, Fischl, 2012).

All of the material described above is available from the HCP database (individual MEG sensor space time courses, magnetometer definition, individual source space and individual head model) and formed the input data for our study.

Head model, source space, and forward model

We used the head models as provided for each subject by the HCP-MEG pipeline. We used the 4D Neuroimaging/BTi coordinate system to identify source positions within these models throughout the paper[1] (with mm units for clarity). We created a forward leadfield with FieldTrip using the magnetometer description and the provided head models, and generated the sensor signals by projecting sources (see below) with a 10 nA.m amplitude through the leadfield to the sensors.

We used two different types of source models, where the source dipoles were either constrained to be orthogonal to the individual cortical mantle, or were free from any anatomical constraint, as explained below.

Source models constrained by anatomy

We performed a first set of simulations constrained by individual subjects' anatomy, where the signal to be added to the resting-state data was generated from one source (i.e. one vertex) of the cortical mesh provided by the HCP. This source was oriented orthogonally to the cortical mesh. Across participants, we identified sources of similar areal origin by means of their index in the cortical mesh provided by the HCP. It is worth mentioning that we do not

[1] Described here: http://www.fieldtriptoolbox.org/faq/how_are_the_different_head_and_mri_coordinate_systems_defined/#details-of-the-4dbti-coordinate-system

imply that these indices map strictly homologous portions of cortex across subjects. The mesh topology is, however, identical in all 89 subjects (all 64,984 vertices are connected in the same way in every subject), and it is used in practice to make cross-subject source-level comparisons and averaging (Fischl, 2012). No within- or between-subject variability other than the one already present in the HCP data was added. Between-subject variability thus occurred because of variations in position and orientation of the source vertices across subjects.

For high-resolution rendering of the cortical surface in figures, we used the high-definition segmentation with 163,842 vertices per hemisphere found in the HCP-structural pipeline (Glasser et al., 2013). These segmentations are linearly coregistered to MNI coordinates (same origin, scale, and orientation), but not "warped" (i.e. non-linearly transformed to closely map to the MNI template brain) as is often performed for mapping individual source reconstructions across participants in MEG. This non-linear mapping would have been inappropriate here since we were interested in between-subject anatomical variability.

All our brain anatomy-constrained simulations used an equivalent current dipole of 10 nA.m as the source of difference between the two conditions. Thus, we did not vary signal amplitude in this study. In a recent study, Murakami and Okada (2015) showed that current density due to local neuronal currents, expressed as equivalent current dipole per unit of surface (in nA.m/mm^2), is remarkably constant throughout a range of vertebrate species. They argued that a value of current dipole moment density of 0.16-0.77 nA.m/mm^2 may serve as an effective physiological constraint for human neocortical MEG/EEG sources. As an indication, according to these estimates, the current dipole value of 10 nA.m that we chose in this study could corresponds to an activation surface ranging between 13 and 62 mm^2. However, complexities arise due to the spatial extent of the sources (e.g. changing orientations around convexities of the cortical mantle). Accounting for these complexities is

beyond the scope of this paper, but it has been tackled in previous studies (Ahlfors, Han, Lin, et al., 2010; Fuchs, Kastner, Tech, Wagner, & Gasca, 2017).

Source models unconstrained by anatomy

As will become clear, measures of source properties that affect detectability tend to covary in ways that are not readily predictable. We thus also used a simulation approach where dipoles were placed independently of cortical anatomy. For these simulations, we took an initial source located around the precentral region (x=-12, y=33, z=70; source position illustrated in Figure 8), oriented it normally to the mean cortical surface at that location, and systematically varied the spatial source properties starting from that location (see "Source properties description", below).

All other parameters of the simulations were kept identical to those of the anatomically constrained simulations described above, except that the amplitude of the injected sources was reduced to 5 nA.m to avoid strong saturation of the power contour plots due to the very weak between-subject variability in simulations where individual parameters were held constant (see details below).

Simulations

In all of our simulations, we used the same procedure based on a Monte Carlo resampling strategy. The Monte Carlo procedure uses repeated random selection from a data sample to approximate a property of the general population, that is, here, the probability of finding a statistically significant effect in the general population (aka. statistical power).

Procedure: Monte Carlo simulations

As illustrated in Figure 1, for each simulation with a given number of subjects and trials, we first randomly chose the required number of subjects from the HCP database, sampling

without replacement. For each subject, we then randomly selected our "trials". Each trial consisted of 25-ms (50 samples) time segments of the continuous resting state data, at least 2 s apart from each other. We then split these trials randomly in two sets of equal size, added signal (according to the procedure explained above) in one set and averaged the data across time points and trials separately for each set. We then ran a paired t-test of the difference between the two sets across subjects at each sensor and noted significance ($p<0.05$, uncorrected). So as to summarize the whole MEG helmet with one value, we used the state-of-the-art spatial cluster-mass based correction for multiple comparisons (Maris & Oostenveld, 2007, 1000 permutations, cluster and significance thresholds both at $p<0.05$) and considered a given comparison significant only when the peak sensor (i.e. the sensor at which the absolute value of the projection of the source signal peaked) was included in a significant cluster. We repeated this procedure 500 times for each trial-by-subject number pair and noted the number of times where the comparison was significant among these 500 simulations. This so-called Monte Carlo statistical power estimate approximates the probability of finding a significant effect if we were to run an experiment with the given parameters (number of subjects, trials and signal properties) in the general population.

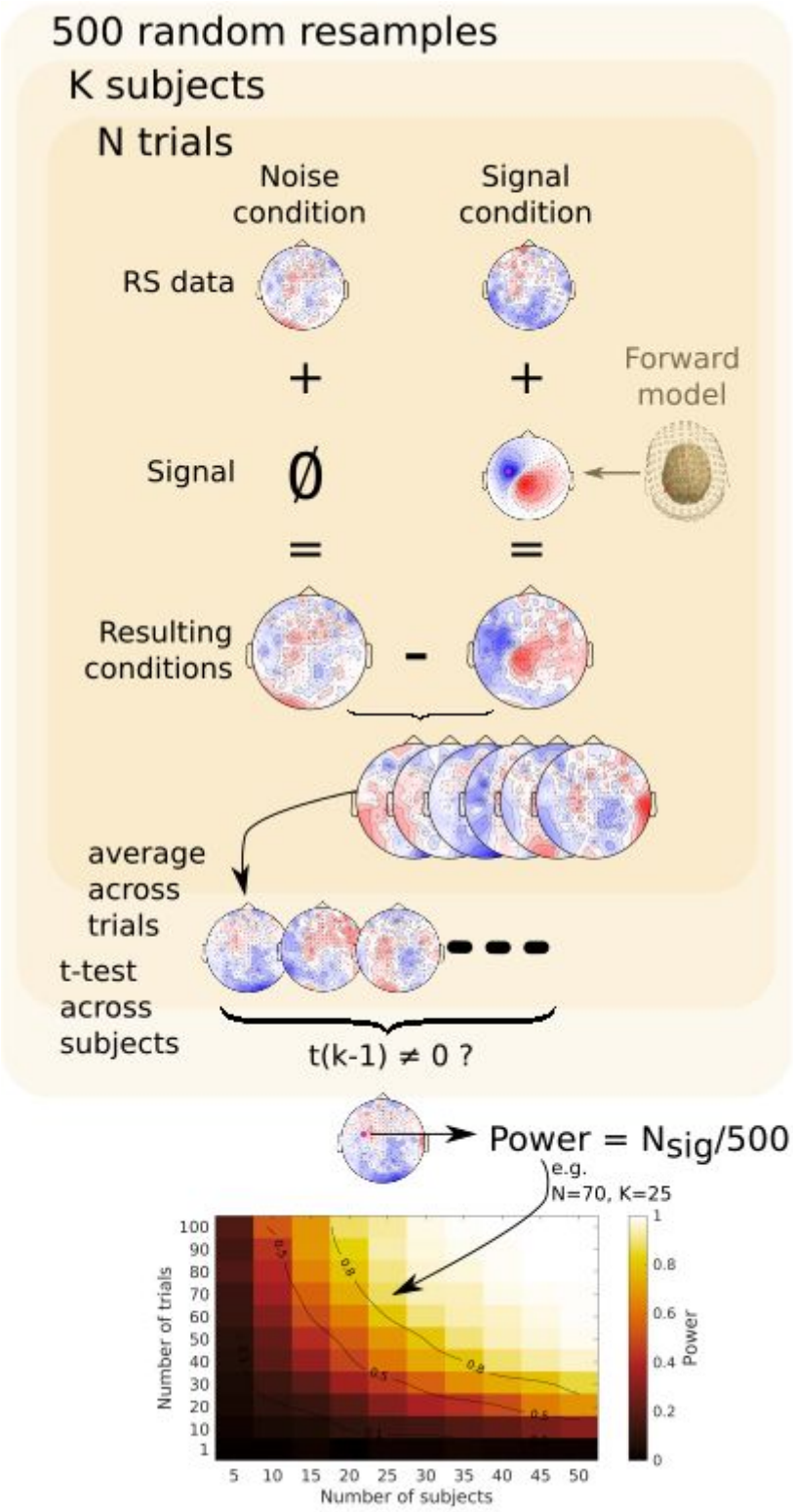

Figure 1. Schematic of simulation method. *For a given "experiment" simulation with K subjects and N trials, K subjects were first selected at random. In each subject, N trials (that is, 50 time-point segments averaged along time) were randomly chosen and assigned to the two "conditions" from the Resting State (RS) data of the HCP. A signal created with the individual's*

head model was added to the trials of the signal condition. The data were averaged across the trials in each condition and the procedure was repeated for the K subjects. Significance was then tested across subjects by means of paired t-tests between conditions, corrected for multiple comparisons. The whole computation was repeated 500 times with different random samples of N subjects and K trials to yield an estimate of statistical power for this [K,N] pair, which was the number of significant permutation tests at the peak electrode (marked with a pink dot on the signal topography and on the statistics topography) divided by 500. This value was then color coded in the bottom power contour plot which represents power for every tested [K,N] pair. Contour lines in this and other power contour plots indicate power values (spline-interpolated for visualization purposes). These plots are used throughout the paper to illustrate the detectability of a given effect.

In an initial simulation, we assessed overall variations in statistical power for simulated sources throughout the brain with the "rule of thumb" number of subjects and trials (25 subjects, 50 trials each). This is reported in the initial results section, "*Overall variations of statistical power throughout the brain*". In subsequent simulations, we ran Monte Carlo simulations for all combinations of trial and subject numbers, from a single trial to 100 trials (in steps of 10) in each condition and with 5 to 50 subjects (in steps of 5), in order to assess the impact of different source properties on statistical power, as described below.

Source properties description

We used the simulation approach described above to evaluate how changes in the spatial properties of the source dipole affect signal detection at sensor level. We examined what we call first- and second-order properties of the sources. First-order properties refer to the position and orientation parameters of the sources. For better interpretability, we defined these first-order properties relative to the sensor array. We defined position as the distance

between the source and the closest sensor, and orientation as the orientation of the source with respect to the closest sensor. Second-order properties refer to the *variability* in first-order properties across subjects. We examined separately the cross-subject variability in position, and the cross-subject variability in orientation, in two sets of simulations. Each property was first examined in simulations using individual brain anatomy, then manipulated in simulations unconstrained by anatomy.

First-order properties: position and orientation of the source with respect to the sensor array

The following two measures allowed us to explore the extent to which source detectability depends on their distance and orientation with respect to the sensor array.

Distance to the closest sensor

In simulations constrained by individual brain anatomy, the distance to the closest sensor for a given source was measured on the individual anatomy in each subject . In simulations not constrained by anatomy, we imposed this distance by shifting the position of the initial source (described in the "*Models unconstrained by anatomy*" section above) by a target distance inwards or outwards, along the radius passing through the sensor closest to the initial source (labeled A44) and the origin of the coordinate system, so that the distance to the closest sensor ranged from 20 to 100 mm.

Orientation with respect to the closest sensor

The orientation with respect to the closest sensor was computed as the arccosine of the product of the orientation vector of the source with the orientation vector of the sensor. The source orientation was normal to the cortex in simulations using individual brain anatomy. It was imposed in simulations not constrained by anatomy, by rotating source dipole around the

y-axis of the coordinate system (that is, the axis passing through both ears). Note that this measure is close to a measure of the extent to which the source may be considered radial versus tangential to the sphere encompassing the sensor helmet for the initial source considered. This measure ranged between 0 degrees for sources pointing directly at the sensor, up to 90 degrees for sources pointing orthogonal to the sensor (towards the front of the helmet).

Second-order properties: variability of the sources across subjects

Variability in source position and orientation are thought to decrease detectability at the group level, and were thus examined here.

Position variability

Position variability was defined as the standard deviation of source position across subjects. In simulations not constrained by individual anatomy, we sampled dipole locations at random from an uncorrelated trivariate (x, y, z) normal distribution with a standard deviation varying from 0 to 10 mm.

Orientation variability

Orientation variability was measured for each source location as the length of the average orientation vector across subjects. In simulations not constrained by anatomy, orientation variability was imposed by adding an uncorrelated random bivariate (azimuth and elevation) normally distributed angle to the individual sources with a standard deviation varying from 0 to 180 degrees.

Results

Overall variations of statistical power throughout the brain

Before getting into the description of the explored spatial properties, we describe here the overall variations of statistical power for detecting sources throughout the brain. For this, we calculated the average statistical power for detecting a source at each possible cortical location for a highly detailed source model following individual brain anatomy, in a simulated dataset of 25 human subjects and 50 trials per condition. Such subject and trial sample size is quite classical in neuroscience studies.

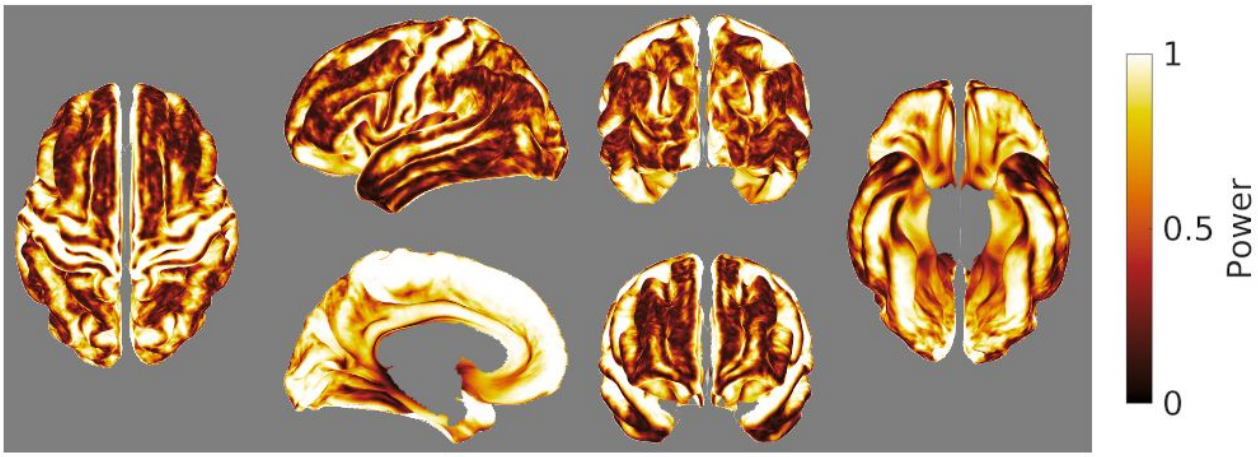

Figure 2. Estimated statistical power across the cortical surface. *Estimated statistical power for detecting (at MEG sensor level) the activity generated by a single dipole placed in turn at every possible neocortical position in simulated experiments with 25 subjects with 50 trials each. Images of superior, left lateral, left medial, anterior, posterior, and inferior brain views are depicted. Note the high power values within major sulci, as well as on the medial wall.*

Figure 2 presents the results of this initial analysis. Note that the absolute statistical power values *per se* are somewhat arbitrary, because we simulated the effect of dipoles with an arbitrary, constant amplitude. Yet, what is important is the striking heterogeneity of statistical power across the brain that this figure reveals. Some of this heterogeneity may appear trivial with regard to what is known on MEG signal sensitivity to source orientation and depth. For instance, sources at gyral crests tend to be detected with very low statistical power, probably because they are radially oriented. In contrast, sources on the walls of the central sulcus, and on the medial surface of the frontal lobe—which are tangentially oriented—are detected with

high power. However, some other observations may appear to be far less intuitive. For instance, the ventral surface of the temporal lobes shows relatively high power in comparison to the lateral surfaces, even though the it is further away from the sensors. This observed heterogeneity of statistical power across the brain calls for better understanding of the influence of source properties on detectable responses at the sensor level. For this, we examined the first- and second-order spatial properties of the sources, i.e. the spatial parameters defining source position and orientation, and their variability across subjects.

In the following sections, for each examined property, the results unfold as follows. We first describe how the property varies across the entire brain. We then use two example locations in the brain to illustrate realistic variations in this property. These realistic variations are always concomitant with changes in other properties. We thus go on with simulations unconstrained by anatomy, exploring an imposed range of values to evaluate the specific impact of this property on statistical power.

Distance to sensors

Figure 3a shows the average distance to the closest sensor for every location on the cortical surface, plotted on the average brain of the 89 subjects in the HCP data. The histogram of this distance across all vertices is plotted in Figure 3b. The distance varies from about 4 mm in cortical regions close to the sensor array (central sulcus, occipital cortex) to about 10 mm for the deepest structures.

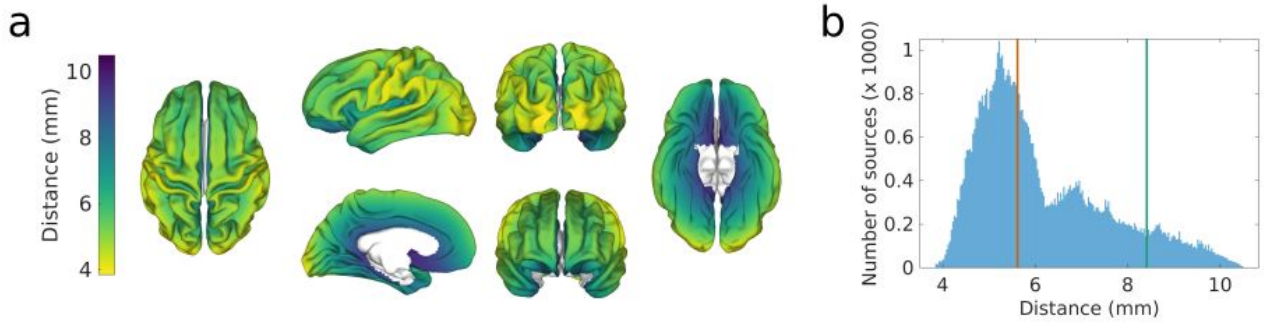

Figure 3. Mean distance to the closest sensor across subjects. *Data calculated from 89 subjects in the HCP MEG dataset and presented on the average brain.* ***a. Mean distance to the closest sensor,*** *mapped on average brain views (from left to right and top to bottom: superior, left lateral, left medial, posterior, anterior, and inferior views).* ***b. Histogram of the distribution of the mean distance to closest sensor across all cortical vertices.*** *The red and green vertical lines indicate the distance to closest sensor of the two dipoles that we selected to illustrate the effect of this parameter in Figures 4 and 5, respectively.*

Based on physics, the distance to the sensors is expected to have a large effect on signal amplitude, and thus on detectability of brain responses at sensor level (Hämäläinen, Hari, Ilmoniemi, Knuutila, & Lounasmaa, 1993; Hillebrand & Barnes, 2002; J. Malmivuo, Suihko, & Eskola, 1997). We now illustrate how group-level statistical power changes with the numbers of subjects and trials for a 10 nA.m dipole placed in two locations - one in the middle frontal gyrus and the other in the orbitofrontal gyrus.

The first location in the middle frontal gyrus (x=71, y=21, z=41 mm) is at a distance of 5.6 cm to the closest sensor (on average across subjects). Figure 4 presents simulation results for this dipole. An identical figure structure will be repeated for illustration of other source locations. Figure 4a displays the chosen dipoles across the 89 subjects of the HCP MEG dataset as well as the average dipole across all subjects. The position variability across subjects can be well seen in these top panels, as well as in Figure 4b, which shows the histogram of dipole positions relative to the average position across subjects, indicating that dipoles spread within +/- 10 mm from the average position in most subjects. Similarly, Figure 4c shows the distribution of

dipole orientations with respect to the closest sensor across subjects. This plot indicates that the orientation for the chosen middle frontal gyrus dipole is on average 69° away from the radial orientation (i.e. away from the orientation that points exactly in the same direction as the closest sensor). Figure 4d shows the average projection of this dipole to the sensor array, revealing a high signal amplitude at frontal sensors. This amplitude, when added to the resting state data could be detected with a standard paired t-test across subjects (corrected for multiple comparisons across sensors) against resting state data alone, with the estimated statistical power level shown in Figure 4e for varying numbers of subjects and trials. In this particular case, achieving 80% power to detect the dipole signal requires at least 20 subjects with ~50 trials per subject, or up to 50 subjects with only 10 trials per subject.

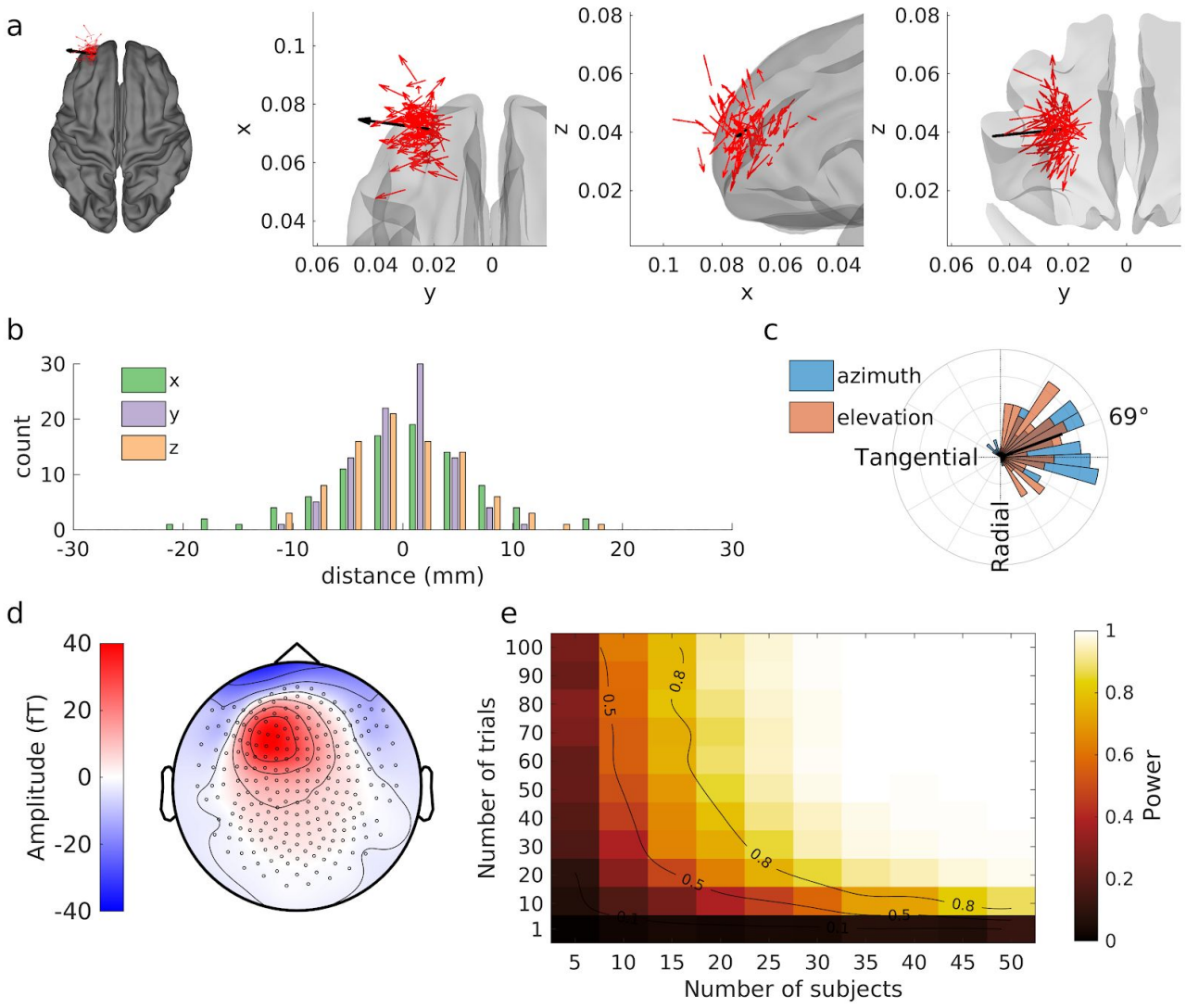

Figure 4. Detecting a source close to the sensors in the middle frontal gyrus. a. *positions of individual dipoles in 89 subjects (red arrows) are represented on the average brain in a superior overall view, and in zoomed superior, lateral, and posterior partial brain views. The bold black arrow in each panel shows the resultant vector or average dipole across 89 subjects.* ***b. Histogram of the distance of each dipole to the mean position of all dipoles in the three cartesian dimensions. c. Polar histogram of the orientations of each dipole relative to the orientation of the closest sensor.*** *Azimuth and elevation are referenced to the sensor orientation vector, so an orientation orthogonal to that of the closest sensor is shown as tangential (90°), and the parallel orientation is shown as radial (0°). The black thick line coming outwards from the center of the plot represents the orientation and length of the resultant vector scaled so that a resultant vector of length 1 (if all dipoles were strictly collinear and pointing in the same direction) would span the whole radius of the plot.* ***d. Average projection of the dipoles in sensor space.*** *Black circles identify sensor positions on the topographical view; the nose is at the top of the view, and left hemisphere appears on the left. The color bar indicates the strength of magnetic field exiting (red) and entering (blue) the head in femtoTeslas (fT).* ***e. Monte Carlo simulations at this location.*** *Color represents estimated statistical power, i.e. the number of significant tests divided by the number of simulations (500) for all tested combinations of trial and subject numbers. Black isocontour lines on the plots highlight spline-interpolated power estimates of 0.5 and 0.8.*

After examining the effects on statistical power of this relatively close-to-sensors dipole, we now turn to the detection of a deeper dipole, situated in the orbitofrontal gyrus (Figure 5, x=44, y=13, z=7), at a distance of 8.4 cm from the closest sensor (on average across subjects). Compared to the previous location just described, in spite of its slightly more consistent location (Figure 5b) and orientation (Figure 5c) across subjects, and an almost tangential orientation (85° away from the radial orientation), the signal from this dipole projects with a much weaker amplitude to the sensor array (Figure 5d), and is accordingly less well detected than the previous dipole (Figure 5e). At least 25 subjects with 90 trials each, or 50 subjects with at least 40 trials each are necessary to detect this dipole with 80% statistical power.

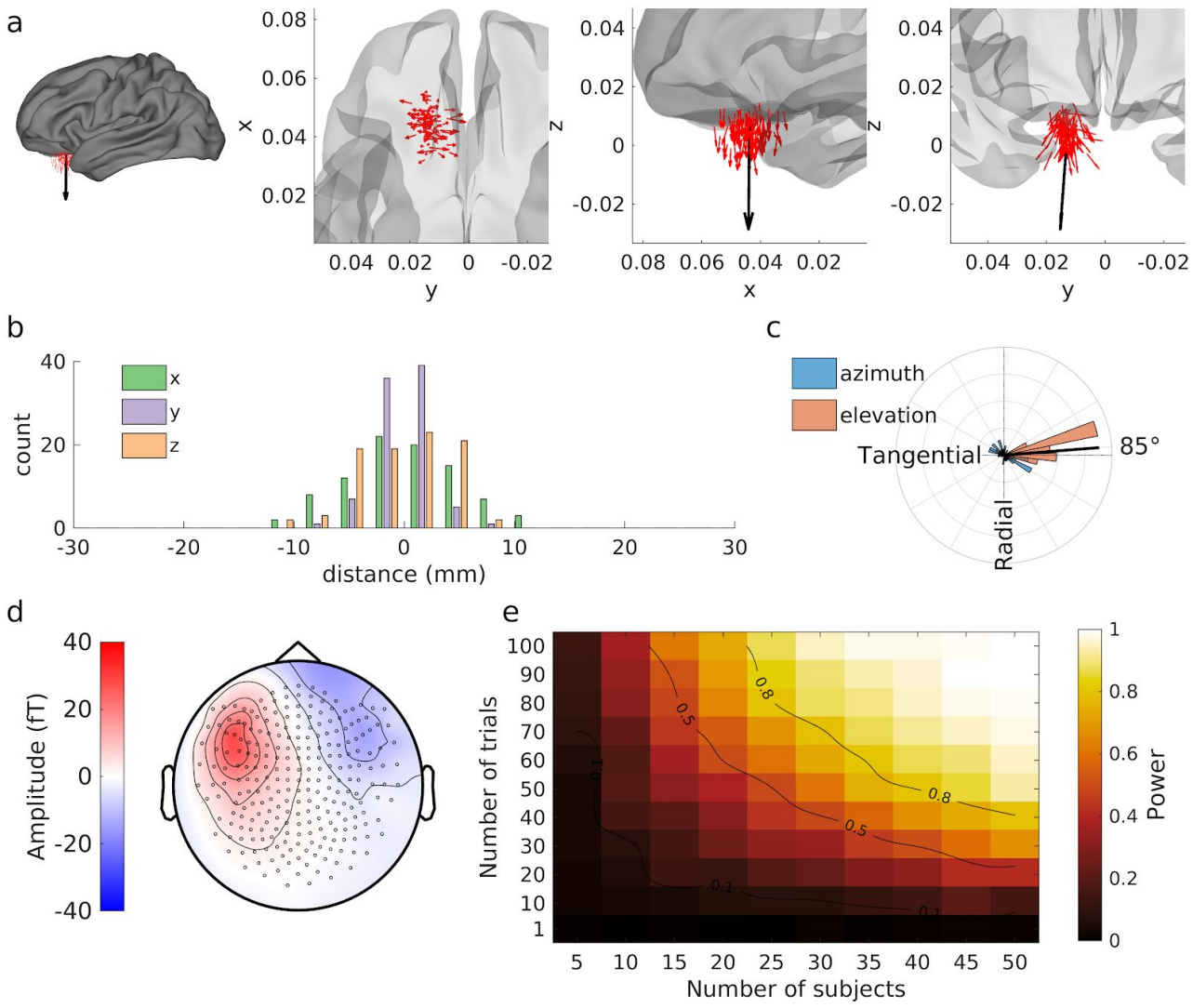

Figure 5: Detecting a deep source in orbitofrontal cortex. *The same conventions as for Figure 4 apply. See main text for further explanation.*

The two anterior frontal dipole locations presented above illustrate qualitatively the effect of distance to the sensors, but also highlight a clear limitation of this first approach of describing the effect of source properties on statistical power. Given how dipole properties covary in realistic source models across brain regions, it is nearly impossible to disentangle the respective contributions of distance, orientation, and their variability across subjects with these realistic source models. Therefore, we now turn to a more selective manipulation of the

distance to sensors by eliminating individual variation in other source properties—due to constraints of individual brain anatomies—in our simulations.

We selectively placed sources at set distances from sensors, along a radius running from an initial source position in the precentral sulcus (x=-12, y=33, z=70; a region close to sensors, detected with high power, as we later discuss; see Figure 8) towards the center of the head, and ran simulations in the same way as before. Figure 6 shows the effect of this manipulation on statistical power for distances to closest sensor ranging from 20 to 100 mm. For all distances explored, dipoles were arranged in strictly the same orientation and relative positions (spatial variability in this case came only from the position variability of the initial source, Figure 6a). The projected signal at sensors is illustrated in Figure 6b, showing the sole effect of distance to sensors on signal amplitude. Finally, Figure 6c shows how estimated statistical power varies across the distance parameter in these simulations. Note that our goal here is not to make a specific claim about statistical power at any specific distance, but rather to show how power varies with distance. In fact, we used dipoles with an amplitude of 5 nA.m in those simulations, i.e. half the amplitude used in the simulations constrained by anatomy, so as not to completely saturate the power plots. Under these conditions, our simulations show that distance to sensors greatly affects detectability: Sources that could be detected with 100% power for any number of trials above 10 in as few as 10 subjects when placed at 20 mm were almost undetectable when placed at 100 mm away from the closest sensor.

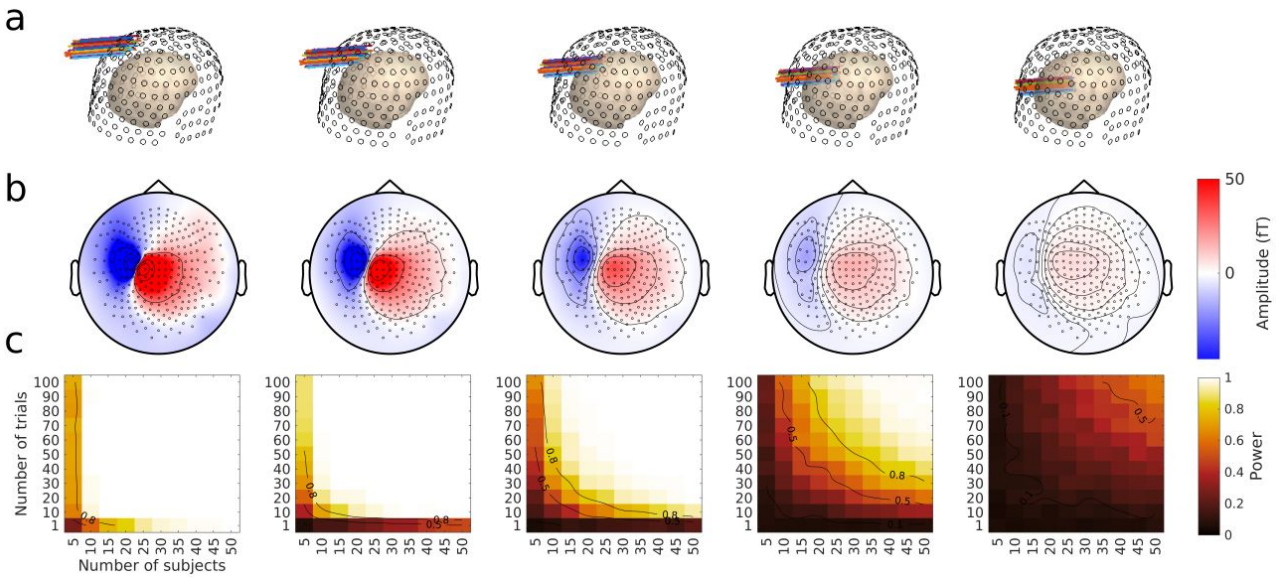

Figure 6. Detecting sources at varying distances from the sensors. a. Locations of individual dipoles. *Source dipoles in 89 subjects (colored arrows) are represented on an individual head model (pinkish inner surface) relative to sensor positions (black open circles on outer surface). The source dipoles were placed at five equally spaced distances ranging from 20 to 100 mm from the sensor closest to an initial source in the precentral sulcus (see main text), towards the center of the head (origin of the coordinate system).* ***b. Average projection of the dipoles in sensor space.*** *The same conventions as in Figure 4d apply.* ***c. Results of the Monte Carlo simulations at each location.*** *The same conventions as in Figure 4e apply.*

In summary, distance to the sensors affects detectability drastically. A source with a given amplitude and spatial properties that is readily detected with very few trials and subjects when placed at a neocortical location close to the sensors is virtually impossible to detect if it occurs in the center of the head. However, other parameters, such as source strength, may compensate for a large distance to sensors, and all relevant parameters should thus be considered, when trying to predict statistical power for given sources in the real brain.

Orientation

Figure 7a shows the average source orientation relative the closest sensor for every location on the cortical surface, plotted on the average brain of the 89 subjects in the HCP data. Most sulcal walls and a large part of the inferior temporal cortex are close to tangential orientation

(0°), whereas the crest of gyri (e.g. superior frontal gyrus) and the depths of sulci (e.g. the insular cortex) are closer to radial orientation (90°). The histogram of source orientations across all vertices is plotted in Figure 7b. It is evident that the distribution is skewed towards values below 90°, reflecting the overall convex topology of the cortical mantle.

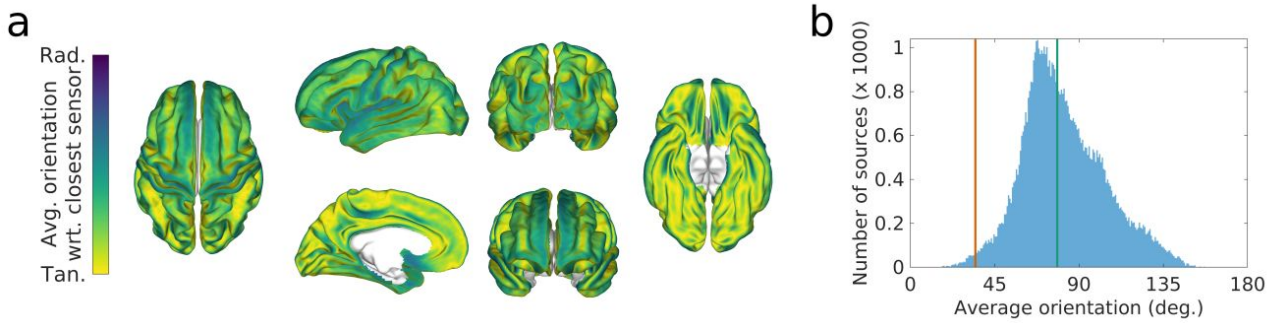

Figure 7. Average dipole orientation with respect to the closest sensor across subjects. *Data from 89 subjects in the HCP data plotted on the average brain.* ***a. Average orientation relative to the closest sensor.*** *The figure shows the mean orientation mapped on superior, left lateral, left medial, posterior, anterior, and inferior views of the brain. For illustration clarity, here, we mirrored the computed angle around the tangential axis (90°). In other words, we discarded information on whether the dipoles pointed towards or away from the sensors and only considered orientation (computationally, the values displayed are the inverse cosine of the absolute value of the product of interest). Legend on color bar: Rad. = radial, Tan.= tangential, Avg. = average, wrt. = with respect to.* ***b. Histogram of the distribution of mean orientation across all cortical vertices.*** *Contrary to panel a, values spread the full half circle. The red and green vertical lines indicate the orientation of the two dipoles that we selected to illustrate the effect of this parameter in Figures 8 and 9, respectively. deg. = degrees.*

Based on physics, we expected the orientation of the sources relative to that of the sensor coils to have a large effect on signal amplitude (Hämäläinen et al., 1993; Hillebrand & Barnes, 2002; J. A. Malmivuo & Suihko, 2004), and thus on detectability. In the following, we use the same logic and display format as in the previous section on distance to sensors.

We first illustrate the effect of source orientation relative to sensors on group-level statistical power by examining two 10 nA.m dipoles at closeby locations in the precentral region, one in the posterior bank of the precentral sulcus, and the other on the gyral crest of the precentral gyrus . Each of these locations have very similar spatial properties, except for their orientation

relative to the closest sensor. The sulcal source (x=-12, y=33, z=70) is close to a tangential orientation, while the gyral source (x=-4, y=37, z=78) is closer to a radial orientation. Our simulations show how the detectability of MEG signals at sensor level varies across numbers of subjects and trials for these two dipoles, as shown on the contour power plots of Figures 8 and 9. The effect of orientation is dramatic in this situation, with 80% power for as little as 15 subjects and 40 trials, or 10 trials in 40 or more subjects for the first (more tangential) dipole, and no sufficient number of subjects and trials in our simulations to reach 80% power for the second (more radial) dipole. This illustrates the well-known effect of source orientation on MEG signal (Cohen & Hosaka, 1976).

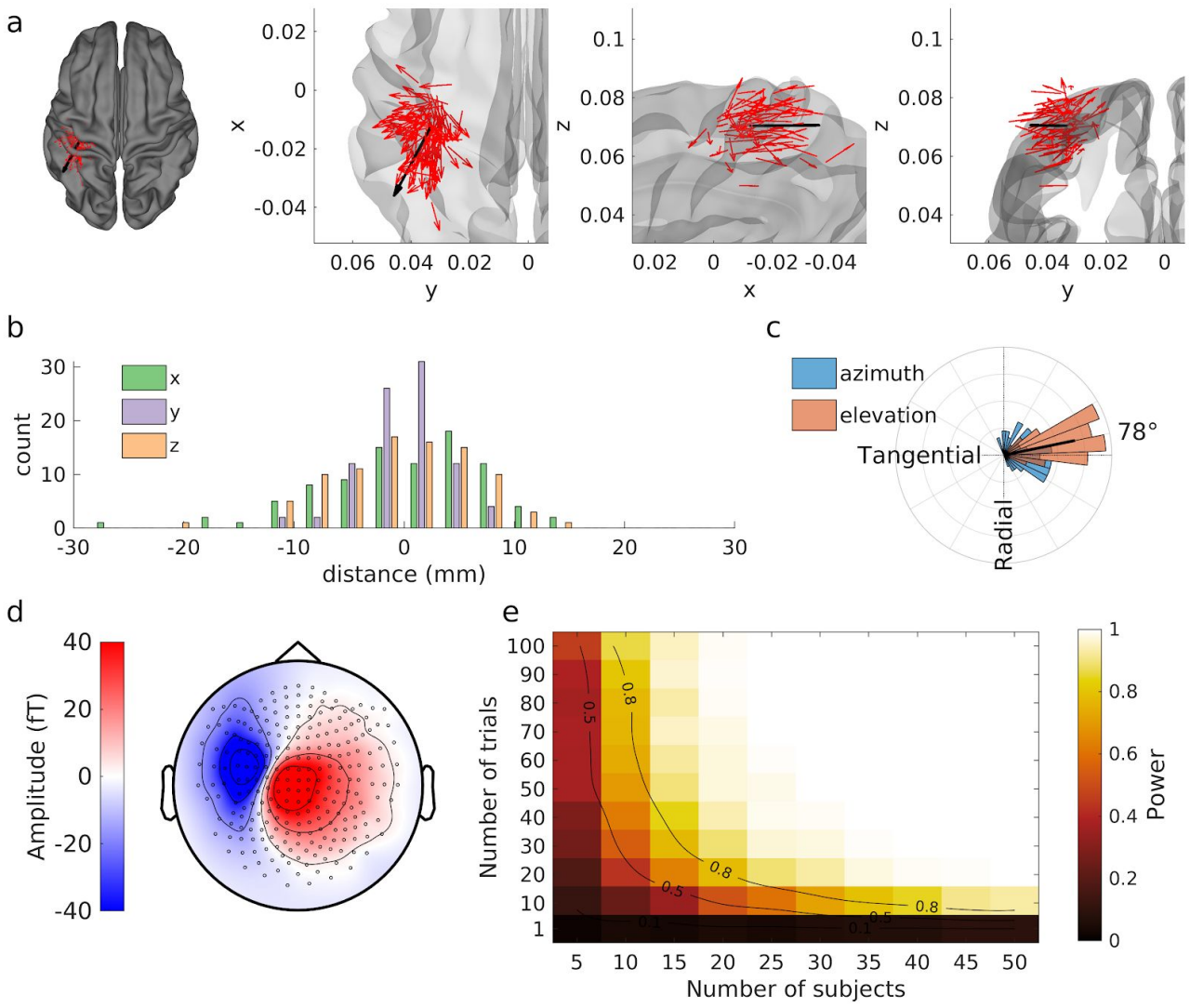

Figure 8. Detecting a tangential source in the precentral sulcus. *The same conventions as for Figure 4 apply. See main text for further explanation.*

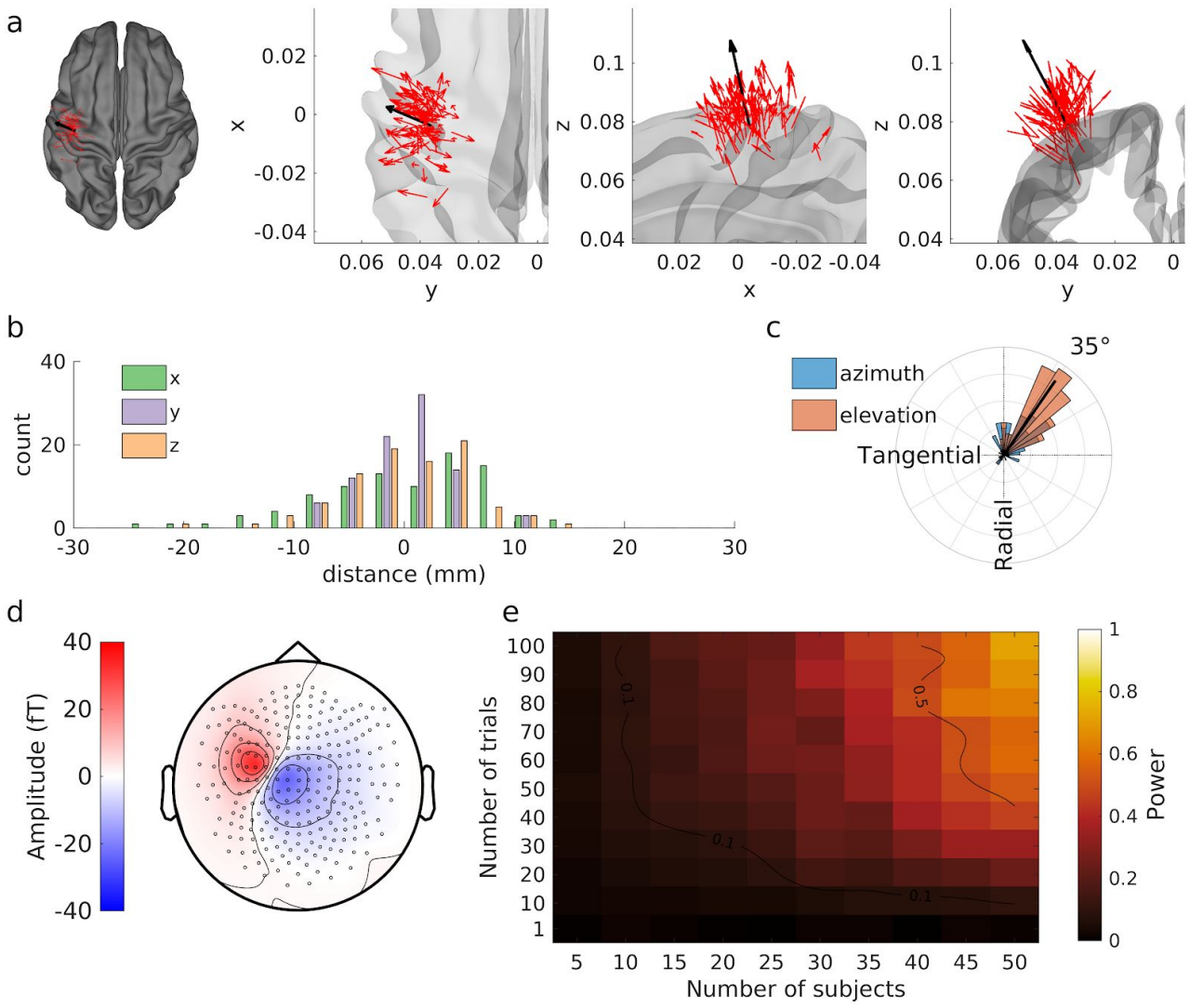

Figure 9. Detecting a radial source in the precentral gyrus. *The same conventions as for Figure 4 apply. See main text for further explanation.*

Similar to the previous section, we now turn to a more selective manipulation of orientation by eliminating individual anatomical variation. Here, we chose a source at a set position and oriented it at set angles with respect to the closest sensor. We then ran simulations in the same way as previously. Figure 10 shows the effect of this manipulation on statistical power for five equally spaced orientations ranging from 0° to 90°, starting from an original orientation strictly parallel to the closest sensor, and progressively tipping around the y axis (passing through both ears), up to an orthogonal orientation (approximately tangential to the

sensor helmet; Figure 10a). The projected signal at sensor level is illustrated in Figure 10b, showing the effect of source orientation on signal amplitude. The most radial sources (left) produce a simulated signal that is about 10 times smaller than the most tangential sources (right), although with similar topography. Finally, Figure 10c shows how estimated statistical power varied in these simulations from a situation where signal could never be detected above 40% with the number of subjects and trials explored for radial orientations, to a situation where any number of trials above 10 to 20 in 15 or more subjects yielded 100% estimated power. Noteworthy, the decrease in detectability is narrowly focused at the radial orientation. A mere 22.5° shift away from this orientation (second orientation from the left in Figure 10) produces a major improvement in detectability, with 80% power reached with for instance 50 trials in 20 subjects.

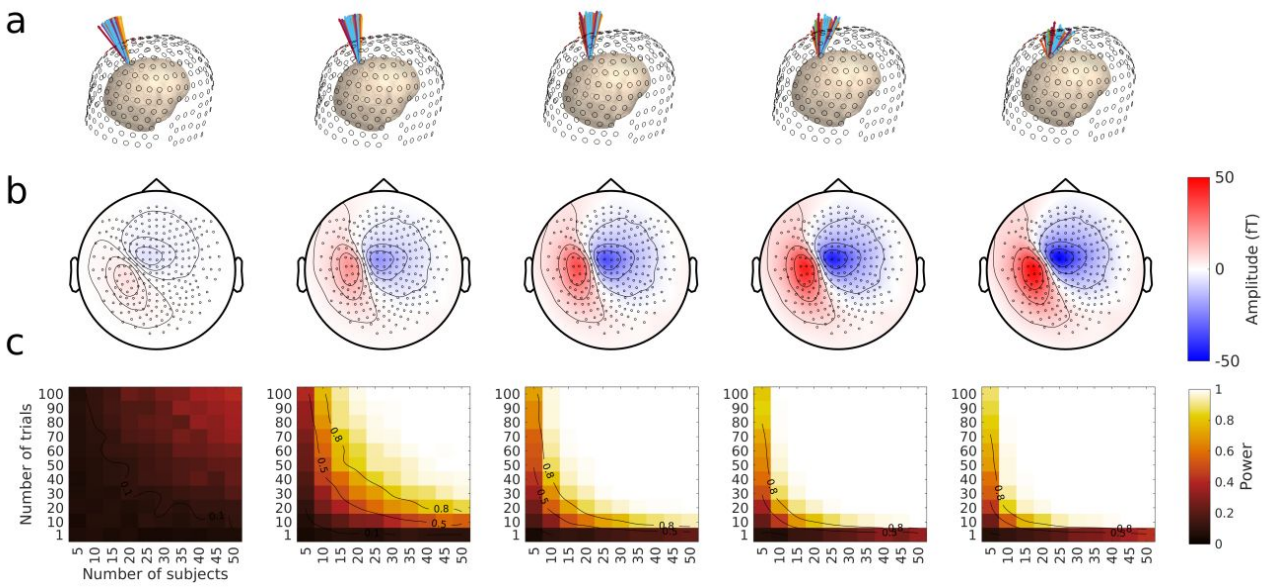

Figure 10. Detecting sources at varying orientations with respect to sensors. *The same conventions as in Figure 6 apply. Orientations vary from 0 degrees (pointing toward the closest sensor), i.e. radial orientation, on the left, to 90 degrees (pointing toward the front of the helmet), i.e. tangential orientation, in 5 equally spaced angles (0, 22.5, 45, 67.5, 90 degrees). See main text for further explanations.*

In summary, source orientation affects detectability drastically, especially when sources are pointing exactly in the same direction as the closest sensor (i.e. radially). However, this loss of power is recovered rapidly as soon as the orientation of the sources shifts away from the radial orientation.

Taken together, the previous two sections allowed us to explore the effect of the position and orientation of dipolar sources—our so-called "first-order properties". Quite predictably, we have seen that these first-order properties have a strong impact on signal detectability. In the following two sections, we describe our explorations of second-order spatial properties derived from the cross-subject variability in source position and orientation, and their effect on group-level detectability.

Position variability across subjects

Figure 11a shows the average cross-subject standard deviation (across the three cartesian dimensions) in position for every location on the cortical surface. The variability histogram across all vertices is plotted in Figure 11b. Overall, values range from 3.4 to 8.1 mm, with maximal values occurring in areas where cortical folding is more variable across subjects, such as in occipital cortex, and minimal values occurring in anterior medio-ventral regions, as well as in insular cortex. Dipoles placed in the most variable regions could be up to 10 mm away from each other, whereas dipoles placed, for example in insular cortex, are on average within 5 mm from each other.

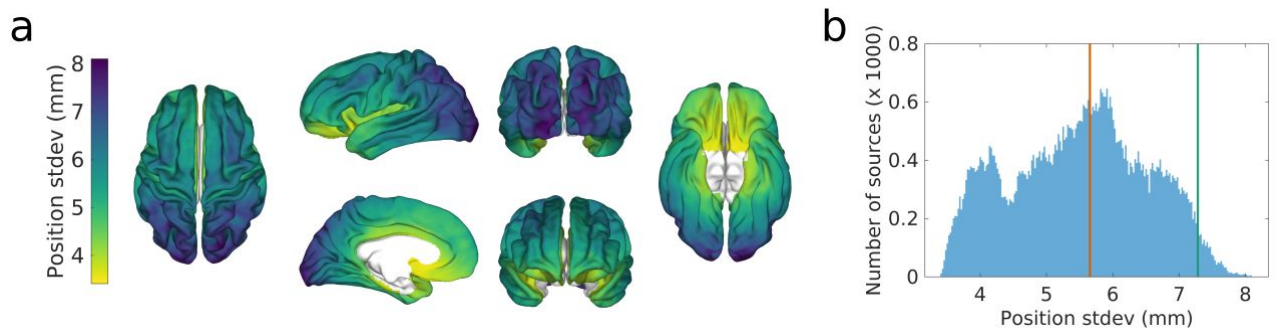

Figure 11. Average cross-subject variability in position *shown on the average brain of 89 subjects in the HCP dataset.* ***a. The cross-subject variability in position*** *is the standard deviation of position of the sources across subjects, shown on superior, left lateral, left medial, posterior, anterior, and inferior views of the brain.* ***b. Histogram of the distribution of average standard deviations across all cortical vertices.*** *The red and green vertical lines indicate the orientation of the two dipoles that we selected to illustrate the effect of this parameter in Figures 12 and 13, respectively.*

Position variability is expected to have an impact on detectability because dipoles at different locations, even if they share the same orientation, project a magnetic field that is topographically different. However, changes in position will rarely change the polarity of the magnetic field at a given sensor, and the similarity in the topographies created by two dipoles may override their differences. So, it is hard to predict the extent to which variability in position will affect detectability at sensor level. To address this question, we first selected two posterior locations in the individual brains of the HCP subjects, one with relatively low spatial variability in fusiform gyrus (x=-39, y=33, z=5), and the second one with large variability in superior occipital cortex (x=-69, y=12, z=48).

For the fusiform gyrus source, the dipole was comprised within a narrow +/- 10 mm region in the horizontal plane for all subjects (x-y plane in Figure 12a, left zoomed view, and Figure 12b), and within +/- 15 mm in the vertical plane (x-z plane in Figure 12a, middle zoomed view, and Figure 12b). The orientation of this dipole was relatively consistent across subjects (Figure 12c). This source resulted in a signal with a strong amplitude at sensor level (Figure 12d), despite being at a distance from sensors, and it was thus detected with a relatively high

statistical power across subjects, with 80% power observed with as little as 15 subjects with 60 trials each, or only 20 trials in 35 or more subjects (Figure 12e).

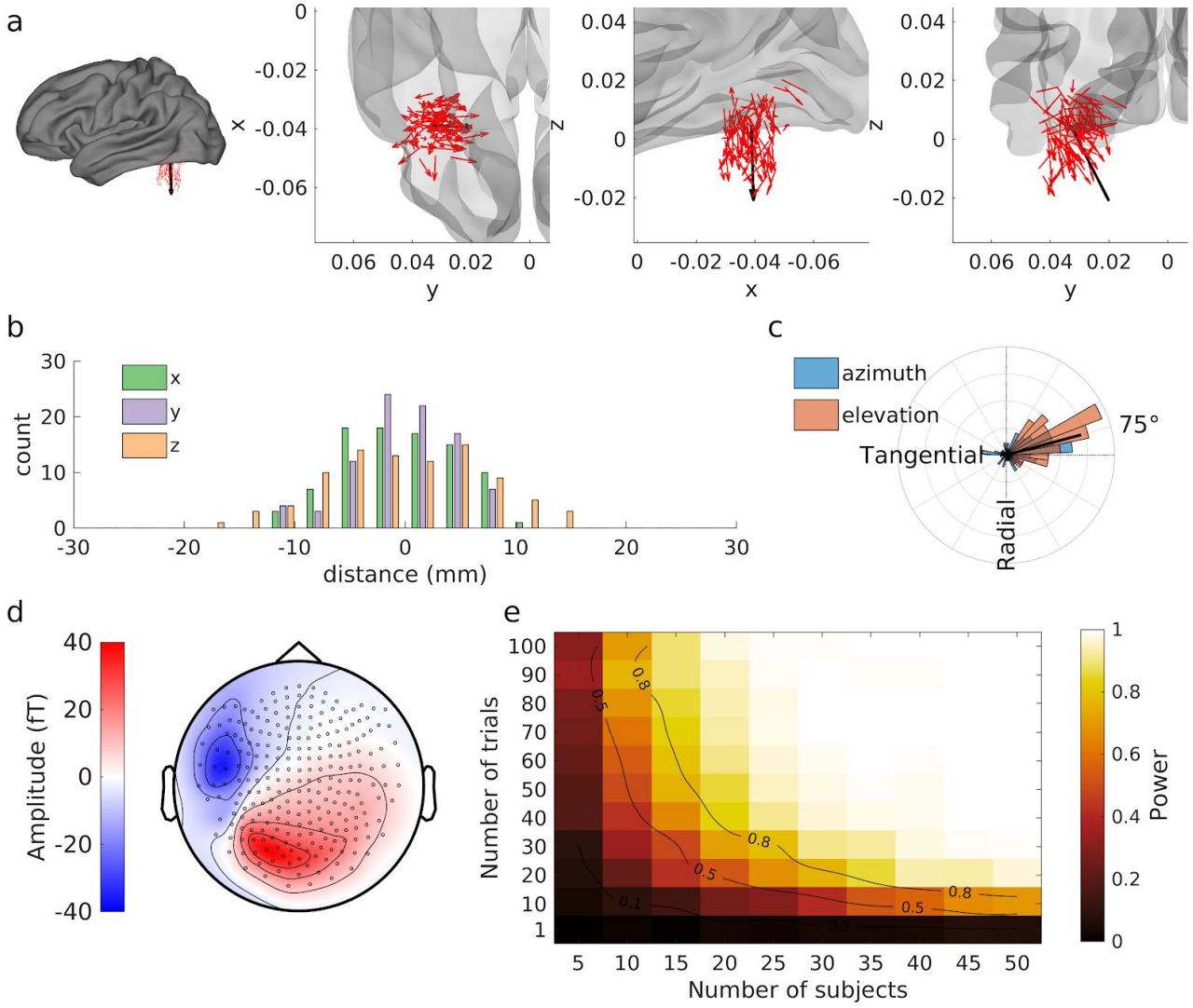

Figure 12. Detecting a source with relatively low cross-subject position variability in the fusiform gyrus. *The same conventions as for Figure 4 apply. See main text for further explanation.*

The source in the superior occipital gyrus, on the other hand, had a much larger variability in position across HCP subjects. This spatial variability is visible in Figures 13a (zoomed insets) and 13b. We note that, although relatively narrow compared to some other sources, orientation variability at this location was also larger than in the fusiform cortex (Figure 13c).

Together, these differences yielded a relatively lower amplitude of signal at sensor level, as compared to the previous dipole in the fusiform gyrus. Accordingly, the detectability of this source was poor, with no tested number of subjects and trials reaching 50% power.

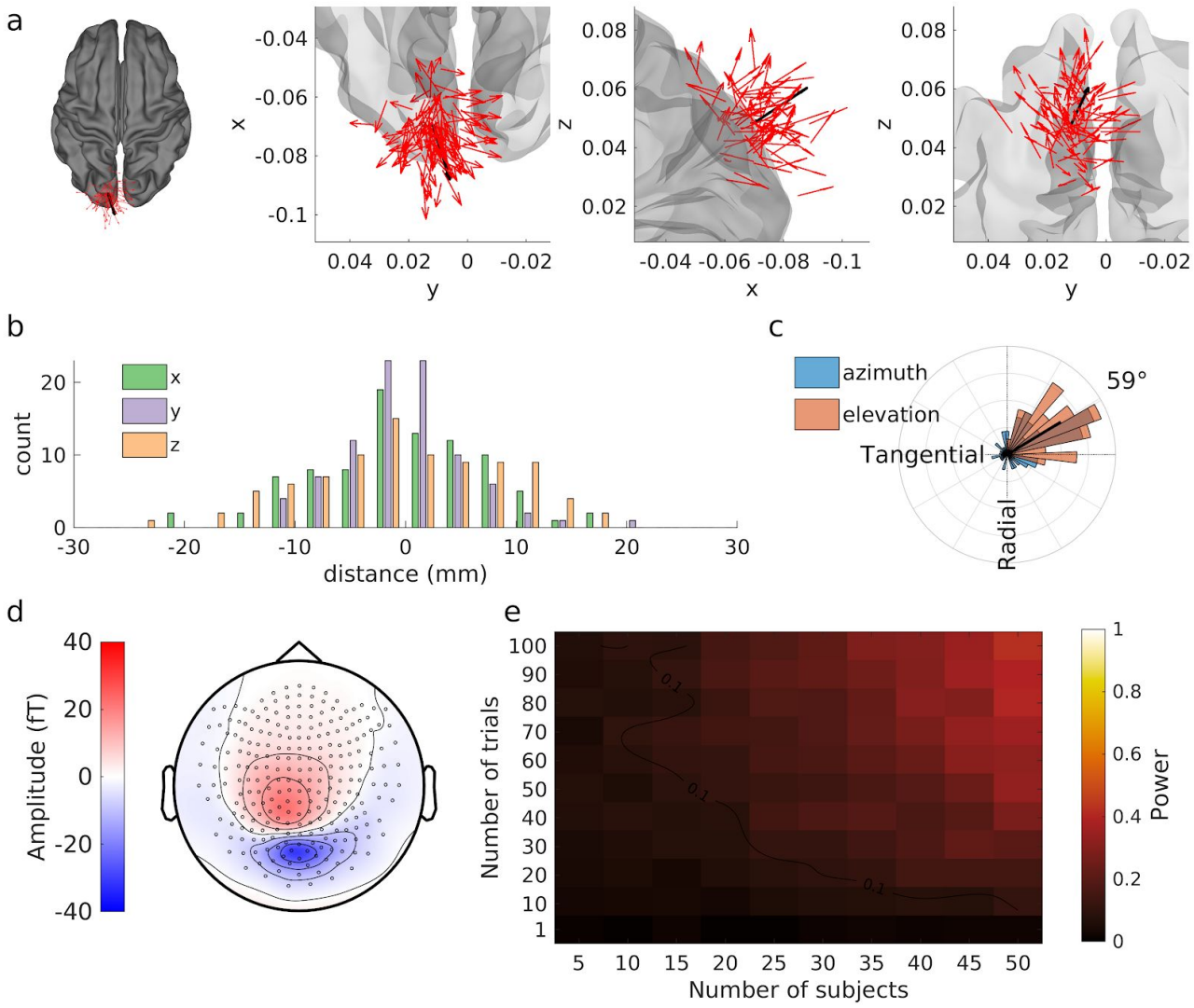

Figure 13. Detecting a source with high cross-subject position variability in the superior occipital gyrus. *The same conventions as for Figure 4 apply. See main text for further explanation.*

The two dipole locations presented above illustrate qualitatively the effect of position variability on detectability. However, as already mentioned, the co-variability of the different spatial properties in anatomically constrained simulations is a confound that undermines

robust conclusions specific to position variability. Therefore, we next move to a more specific manipulation of position variability, maintaining other variables constant, in simulations unconstrained by anatomy. Here, we selectively allowed dipoles to change position across subjects by sampling positions at random from a 3D normal distribution with a set standard deviation across subjects. Figure 14 shows the effect of this manipulation on statistical power for five equally spaced position variabilities ranging from 0 to 10 mm standard deviation. For all explored position variabilities, the average position of dipoles across participants tended toward the same position in the precentral gyrus (Figure 14a). The projected signal at sensor level is illustrated in Figure 14b, showing little effect on signal amplitude, and in turn little effect on estimated statistical power (Figure 14c).

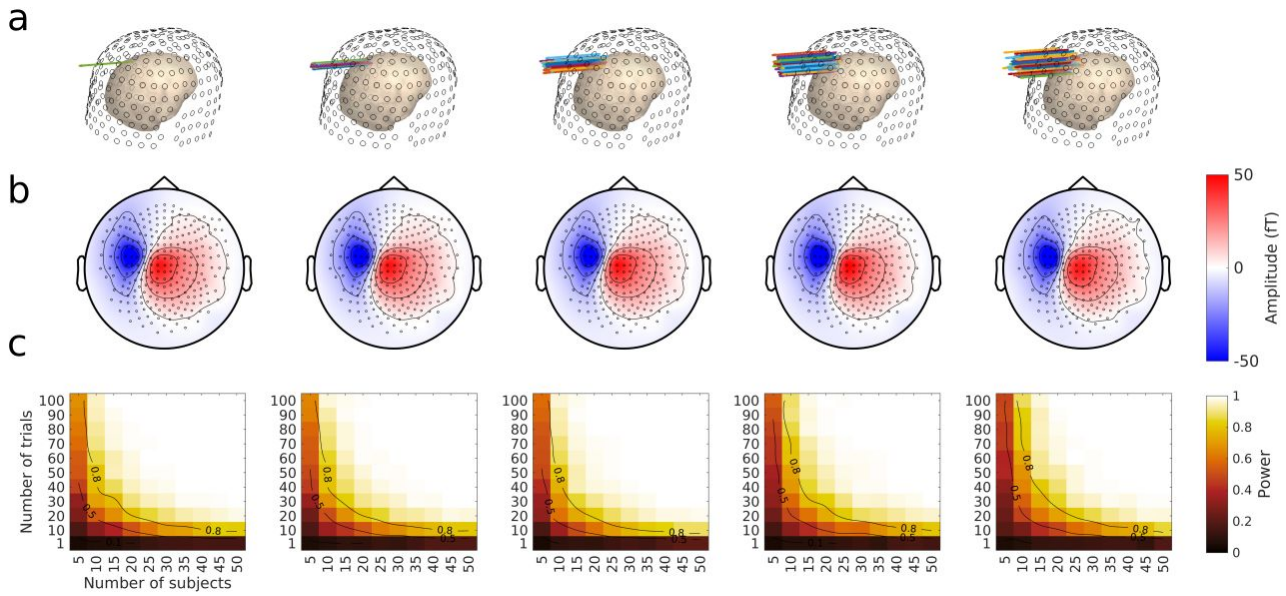

Figure 14. Detecting sources with varying cross-subject position variability. *The same conventions as in Figure 6 apply. See main text for further explanations.*

In summary, manipulating position variability had little effect on detectability when the manipulation was selective. In realistic settings, regions with high position variability across

subjects in the occipital cortex tend to be also more variable in terms of orientation. We explore further the effect of orientation variability across subjects in the next section.

Orientation variability across subjects

Figure 15a shows the resultant vector length (i.e. the length of the average dipole across subject) for every location on the cortical surface, plotted on the average brain of the 89 subjects in the HCP dataset. The histogram of these vector lengths across all vertices is plotted in Figure 15b, which highlights the heterogeneity of orientation variability across the brain. Overall, values ranged between 0.35 and 1, with minimal values in lateral occipital regions, and maximal values along the central sulcus, the ventral prefrontal cortex, dorsomedial prefrontal cortex and insular cortex. Dipoles in the most variable regions of the lateral occipital cortex could be oriented up to 180° apart (i.e. in opposite directions) across subjects, dramatically reducing the net contribution to the signal at sensor level across a group of subjects.

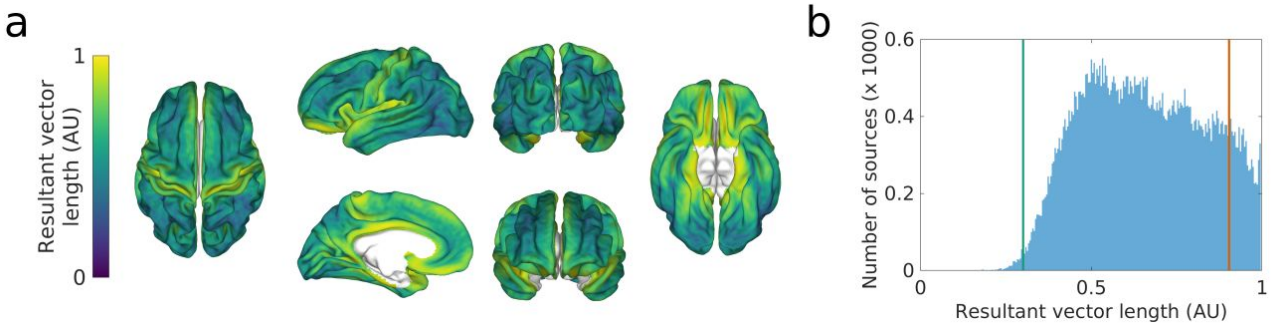

Figure 15. Average cross-subject variability in orientation *shown on the average brain of 89 subjects in the HCP data.* ***a. The cross-subject variability in orientation*** *is the average resultant vector length across individual sources, for each cortical location. It is shown on superior, left lateral, left medial, posterior, anterior, and inferior views of the brain.* ***b. Histogram of the distribution of resultant vector length across all cortical vertices.*** *The red and green vertical lines indicate the orientation of the two dipoles that we selected to illustrate the effect of this parameter in Figures 16 and 17, respectively.*

Orientation variability is expected to have an impact on detectability because dipoles with different orientations, even if they share the same position, project a magnetic field that has a different topography. Contrary to position variability, changes in orientation can change the polarity of the magnetic field at a given sensor even with minimal orientation change. Hence a major effect on statistical power may be expected. Here, we review two locations, one with relatively low orientation variability across subjects, in the insula (x=6, y=36, z=30), and the second one with large variability, in the posterior superior temporal sulcus (x=-47, y=42, z=30).

For the insula source, Figure 16a shows that most individual sources are parallel (see in particular the middle zoomed inset), resulting in a length of the resultant vector of 0.90, shown as a black line in Figure 16c, and an average projection to sensors with relatively high amplitude. The power to detect this signal is such that 20 subjects with 90 trials each, or 50

subjects with 30 trials each allow detecting the 10 nA.m source with 80% statistical power.

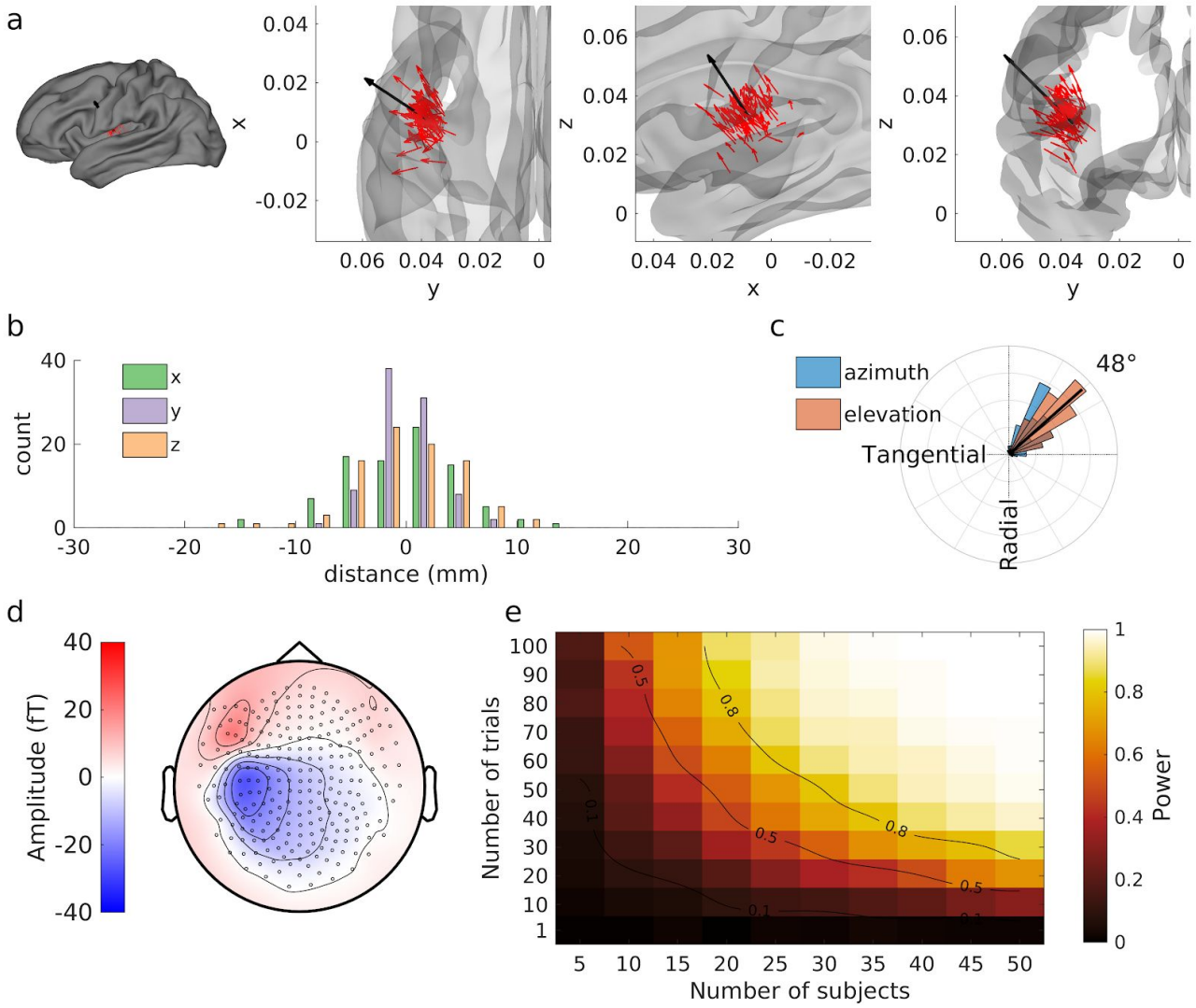

Figure 16. Detecting a source with low cross-subject orientation variability in the insula. *The same conventions as for Figure 4 apply. See main text for further explanation.*

The source in the posterior superior temporal sulcus on the other hand showed a highly variable orientation across subjects (Figure 17a and c), with a resultant vector length of 0.30, and a very weak average amplitude of the signal projected at sensor level (Figure 17d). Accordingly, this signal was virtually undetectable within the range of explored subjects and trial numbers (Figure 17e).

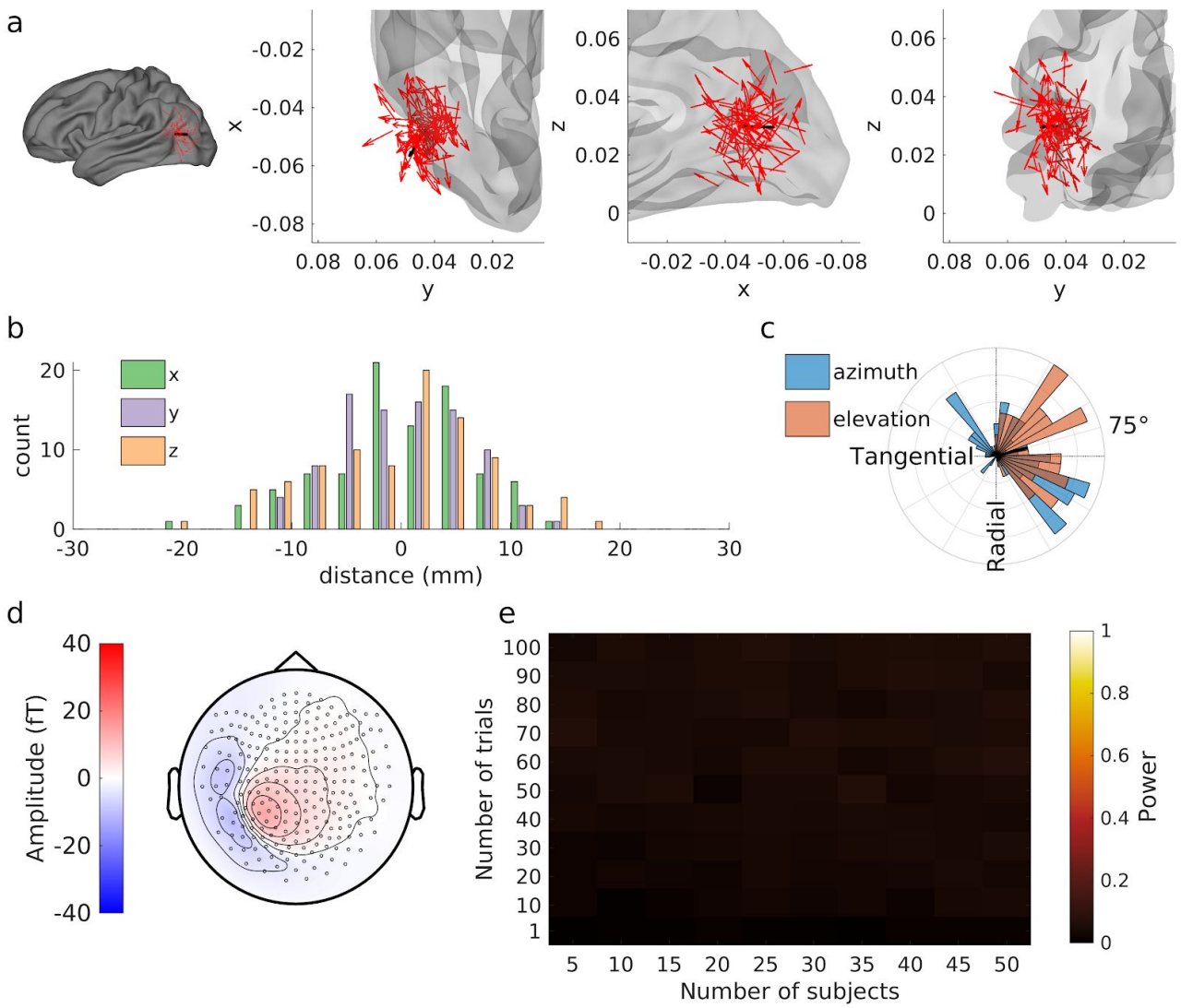

Figure 17. Detecting a source with high cross-subject orientation variability in the posterior superior temporal sulcus. *The same conventions as for Figure 4 apply. See main text for further explanation.*

The two dipole locations presented above illustrate qualitatively the effect of orientation variability on detectability. To examine this effect more rigorously, we next move to a specific manipulation of orientation variability, in simulations unconstrained by anatomy. This time, position was held constant and only orientation was varied across subjects. We sampled orientations at random, adding a normally distributed random azimuth and elevation to the original orientation (arbitrarily set to the tangential source in the precentral sulcus examined

in Figure 8), with a standard deviation set to five evenly spaced values between 0° (fixed orientation) and 180°. The effect of this manipulation on statistical power is illustrated in Figure 18. It is noteworthy that the effect of orientation variability is most visible for random orientations with a standard deviation above 90°. This is arguably due to the fact that dipoles cancelling each other out (i.e. oriented in opposite directions) can only occur when the distribution of orientations is sufficiently broad.

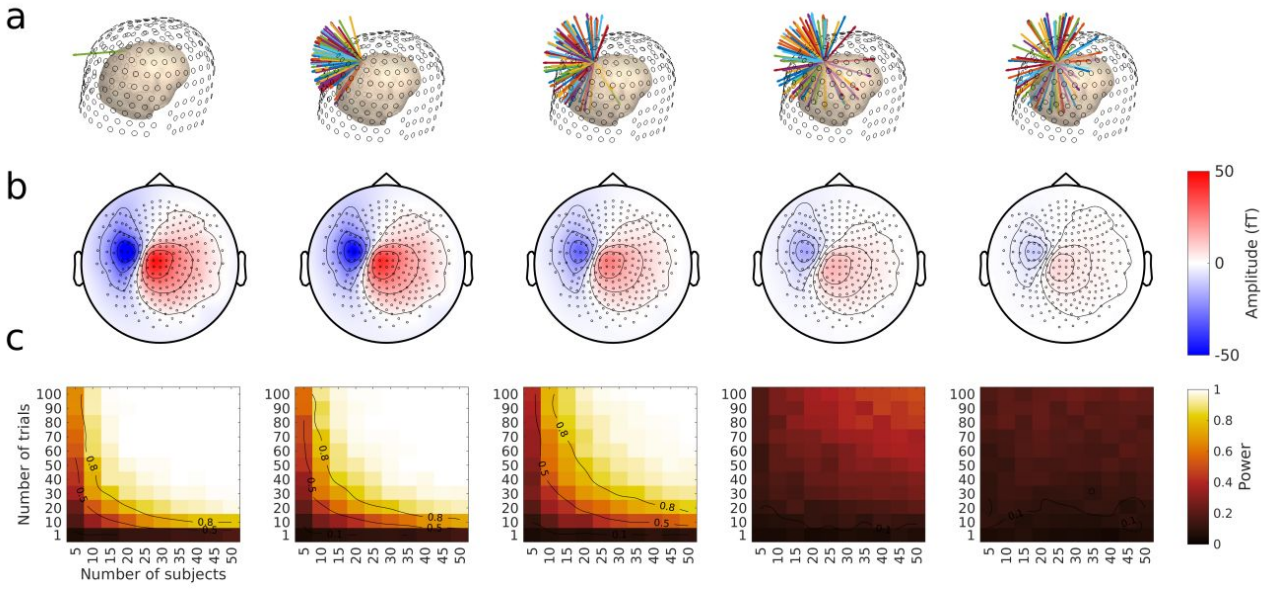

Figure 18. Detecting sources with varying cross-subject orientation variability. *The same conventions as in Figure 6 apply. See main text for further explanations.*

In sum, we showed that second-order properties of the sources have very different effects on detectability. Within the range of variabilities explored, orientation variability had a much larger effect than position variability. Together with the effect of the first-order properties examined above, these results highlight the importance of considering source properties when planning a well-powered MEG experiment. We discuss this and other issues in more detail below.

Discussion

Adequately powered MEG (and EEG) experiments require an appropriate sample size. In this study, we showed that the properties of the brain sources expected to contribute effects in a given experiment critically affect attainable statistical power in MEG. Specifically, we focused on the spatial properties of the sources and on their variability across subjects to examine how these factors affect group-level statistical power with classical tests at sensor level. We showed--quite expectedly--that the distance of the sources to the sensor array has a strong effect on detectability, with deeper sources being virtually undetectable with reasonable sample sizes for a single-session MEG experiment. We also observed another well-known effect, namely that sources oriented radially are almost undetectable in MEG data, as compared to other orientations. Looking next at the influence of cross-subject variability, we observed that position variability across subjects had, in fact, little effect on statistical power, whereas orientation variability (or the lack thereof) strongly affected statistical power. Signal detectability at sensor level thus depends not only on the source origin (position and orientation), but also on the less well predictable cross-subject variability of this origin. Therefore, one take home message for this study is that there is no simple solution for finding the optimal number of trials and subjects for all types of evoked MEG (or EEG) studies. Specifically, there will likely be variations in power across different brain structures within a single experiment that we believe make calculations of power challenging. Thus, we emphasize the importance of considering the expected brain sources of activity, their anatomical location, and their cross-subject variability while planning studies in cognitive, affective, and social neuroscience. Below we discuss some issues and considerations for planning MEEG studies, taking these variables into account.

Previous studies have explored the effects of spatial properties from a single-subject signal detection perspective, aiming to provide estimates of how much MEG signal can be detected given these properties and the amount of noise in an individual subject's data (Ahlfors, Han, Belliveau, & Hämäläinen, 2010; Goldenholz et al., 2009; Hillebrand & Barnes, 2002). It was demonstrated that source depth relative to sensors is the most important factor affecting source localizability, with additional effects due to source orientation (Hillebrand & Barnes, 2002). Additionally, others showed that MEG signal-to-noise varies considerably with source location (Goldenholz et al., 2009) and orientation (Ahlfors, Han, Belliveau, et al., 2010) in the brain. Pushing the analysis one step further, our group-level statistics approach takes a pragmatic stance on this issue, speaking directly to experimenters for whom changes in signal-to-noise ratio are particularly relevant for deciding on sample size, i.e. how many subjects or trials to include in an experiment.

We showed here that this question cannot be answered unequivocally. First, the large effects of spatial properties across brain structures on detectability that we observed (Figure 2) are a major challenge for predicting statistical power. Indeed, not only do the parameters that we examine greatly vary across the brain (as shown by Figures 3, 7, 11 and 15), but also these parameters are not independent of one another across the cortical mantle. For instance, sources with the shortest distances to sensors also tend to have a radial orientation, as gyral crests are generally found on the outer surface of the brain. Other regions with mostly tangential orientations, such as along the central sulcus, can be spatially highly consistent across subjects. In contrast, some brain regions have high degrees of inter-individual variability due to variations in cortical folding (e.g. the posterior STS, MT+/V5, inferior parietal cortex, Caspers et al., 2006), high degrees of curvature (e.g. occipital pole), or hemispheric asymmetries (Croxson, Forkel, Cerliani, & Thiebaut de Schotten, 2018; Ochiai et

al., 2004). Therefore, variations of power across the brain for a given sample size are difficult to predict in practice.

Furthermore, statistical inference is usually made at the group-level. Predicting the adequate number of subjects for a given study also implies considering between-subject variability in different cortical regions. Although first-order spatial properties and their effect on signal are straightforward to measure and model with Maxwell's equations, second order properties' effects on group-level statistics are much more challenging to model. Our approach was to directly manipulate anatomical variations in our simulations free from anatomical constraints. Our first observation is that strictly parallel dipoles tend to combine across subjects, even when their position is relatively spread out, to create a more readily detectable net magnetic field. Randomly oriented dipoles on the other hand tend to cancel each other out. Beyond a certain orientation variability, statistical power becomes critically low, giving no opportunity to detect a signal even with a large number of subjects. This observation has an interesting implication: when comparing our whole brain power analysis for a sample of 25 subjects (Figure 2) with the whole brain signal-to-noise mapping for a single subject in MEG obtained by Goldenholz et al. (Goldenholz et al., 2009, their Figure 2, first row), at least two differences are apparent: First, some regions with high SNR e.g. the occipito-parietal cortex, remain hard to detect at the group-level. Second, other regions with low SNR, in particular on the medial and ventral surfaces of the brain, are detectable at the group level (e.g. in anterior cingulate cortex, or ventromedial prefrontal cortex). We hypothesize that cross-subject variability explains these discrepancies. On the one hand, it is thanks to a particularly consistent source orientation across subjects that some regions with low SNR in Goldenholz' study still show high power in our study, and on the other hand other regions with high SNR are less detectable at the group level due to high variability in source orientations across subjects. Overall, although there is no unequivocal answer to the question

of an appropriate sample size in MEG (or EEG) experiments, our study highlights the importance of considering between-subject variability in addition to the long-known spatial properties of sources when it comes to deciding how many subjects and trials to include in an experiment.

An important point to note regarding the decreased detectability with higher between-subject variability is that it is standard procedure in MEG research to localize sources in individual subjects on their own anatomy before combining data for group analysis in source space (Hari & Puce, 2017; Jas et al., 2018). Going for source-level analysis at the individual subject level can avoid a large portion of the problem of between-subject variability, provided that data at the cortical level are properly aligned across subjects. Specifically, the cancellation effects we observe here across subjects when sources of different subjects point in opposing directions (Figure 17) should be largely avoided when sources are estimated in individual subjects first, then aligned across subjects on a template before being averaged (Hari & Puce, 2017). Further studies will have to examine the limits of this reasoning in detail. In particular, it will be important to examine group-level source detection in detail if we want to formulate more precise recommendations in the future.

A similar group-level approach has recently been used by Boudewyn et al. (2018) for EEG data in sensor space, exploring the dependency of statistical power on sample size (subjects and trials), while focusing on effect size at the electrode level for a set of ERP components. Our approach here is similar, but instead of starting from an expected ERP difference at sensor level, we started from the expected neural source. This complementary approach may allow principled investigations beyond known ERP effects, and could allow a better planning of studies targeted at activating specific brain regions. It also has potentially strong practical

implications as power analyses are usually mandatory in grant applications, or for preregistering studies.

Another important aspect to consider for future studies will be that of spatially extended sources. Indeed, although dipolar sources are easy to model and often used to appreciate detectability, more realistic extended patches of activity can have a difficult-to-predict net effect at sensors. For instance, such extended patches that occur in the homologous regions of the two hemispheres on the mesial wall, or are spread across opposite walls of a sulcus can show relatively lower detectability due to opposite sources cancelling each other out (Ahlfors, Han, Lin, et al., 2010; Fuchs, Kastner, Tech, Wagner, & Gasca, 2017; Hari & Puce, 2017). It should be noted, however, that functional area borders often tend to follow gyral and sulcal crests, making synchronous activation on both sides of a sulcus less likely (Destrieux, Fischl, Dale, & Halgren, 2010; Glasser et al., 2016). The detectability of extended sources at the group-level in such cortical regions needs to be explored in the future for additional practical recommendations.

High-temporal resolution is desirable in cognitive, clinical and social neuroscience studies, particularly where naturalistic and hyperscanning data are acquired. It will therefore be of importance to extend explorations to the time domain, and characterise changes in variability in the time domain, particularly before and immediately after stimulus onset and examine effects on group-level signal detectability at specific latencies. In the future, it will be important to consider investigations in sensor space, in addition to those in source space, as portable EEG and MEG studies are likely to use relatively low numbers of sensors, making it unrealistic to transform these data into source space.

Given the current emphasis on whole-brain data collection in cognitive neuroscience and the explosion of network science based data analyses (Bassett & Sporns, 2017) and the

modulation of activation as a function of perceptual and cognitive manipulations (Medaglia, Lynall, & Bassett, 2015), it is important to understand how statistical power can vary across the brain and the critical dimensions in which these variations occur. Critically, the statistical power for detecting a given network will *at most be only as good as its least detectable node*. It is thus important while planning an experiment entailing a network analysis to plan sample size accordingly.

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