A novel MR-compatible device for providing forces to the human finger during functional neuroimaging studies

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Abstract

Many motor learning experiments involve subjects performing a task while experiencing external force perturbations. However, it is difficult to transfer these tasks to functional magnetic resonance imaging (fMRI) studies, and much of the technology that currently exists to facilitate this is expensive to produce and difficult to use. Here, we report on the design and construction of a novel device (the 'force coil') that is simple and inexpensive, and that uses the static magnetic field inside the scanner to provide forces to the human finger. The coil incorporates a potentiometer in the base to allow the recording of angular position. To test whether the magnetic field generated by the current flowing through the coil would interfere with the functional images collected, we compared images from a phantom during the use of the coil at arm's length in a 7T magnet. There was no noticeable interference from the coil at the levels of current used in this experiment, which produced about 10 N of force in a 7T scanner. In conclusion, the force coil is a cheap, easy to operate device which provides forces to the finger inside the scanner without affecting image quality. Designs based on this principle are likely to prove useful in studies of motor learning using fMRI.

Keywords

fMRI, motor control, human, forces, finger

Introduction

Functional magnetic resonance imaging (fMRI) is fast becoming the method of choice for neuroscientists and psychologists seeking to understand more about the neural correlates of human behaviour. There are limitations to the technique however, primarily in the poor temporal resolution of the imaging data collected, but also in the types of behavioural experiments that can easily be performed given the constraints imposed by the large magnetic field present in all MR scanners. Being unable to use ferrous materials for the construction of equipment for use in or in close proximity to the scanner also means that costs tend to spiral. Progress in this area has been made primarily in visual neuroscience and a range of eye-tracking devices (e.g. Kimmig et al., 1999) have been developed that use optic fibres to transmit information about eye position to a computer outside the scanner. For example, Miall and Jenkinson (2005) tracked subjects' eye movements in the scanner while they performed a hand-eye coordination task. However, in the field of motor learning there have been many studies published in which subjects encounter a perturbing force which they have to correct for. Studies that use robotic manipulanda (Kurtzer et al., 2005; Shadmehr and Mussa-Ivaldi, 1994) are particularly difficult to transfer to the scanner although this has been done with PET studies (Krebs et al., 1998; Nezafat et al., 2001). Multi-joint robotic arms manufactured specifically for use in such fMRI studies have been developed (Gassert et al., 2006) and show some promise but the major downside, as ever, is the cost and complexity of such designs.

Here we introduce a novel piece of equipment that uses basic physical principles to produce forces on the right index finger. Based on electromagnetic theory, a rectangular coil of wire pivoted at one end produces a force when a current is passed through it inside the magnetic field of the scanner. This 'force coil' and other similar designs using the same principle (e.g. Riener et al., 2005) can be constructed quickly and cheaply and do not require expensive materials or complex control systems. The coil can be attached firmly to the scanner bed and adjusted for comfort, depending on the length of the subject's arm. The displacement of the coil when a current is passed through it is recorded via a potentiometer in the base, and the whole system can be controlled through an integrated computational setup consisting of data acquisition cards and a central MATLAB interface.

We describe the design and construction of our force coil as well as procedures for use and calibration of the instrument. Even though a current passing through the coil creates a magnetic field, we show theoretically and experimentally that this field is not large enough to affect the functional images obtained at the head when the coil is at arm's length. We demonstrate the use of the coil *in vivo* by outlining an experiment in which it is used to provide a sequence of forces to subjects' right index fingers and report the functional images obtained, which show activation in the primary motor, primary somatosensory and posterior parietal cortical areas when force is applied to the finger in comparison with rest.

Materials and methods

We based the idea behind the force coil on the Lorentz force law (eqn.1), which states that the force on a current-carrying wire perpendicular to a static magnetic field is proportional to the length l of the wire, the size B of the field and the magnitude I of the current. A coil with n turns of wire will produce n times the amount of current when pivoted at one end (eqn. 1, and see Fig. 1).

$$F = nBIl \qquad (1)$$

A current of 0.5 A in 10 turns of wire of length 0.2 m will therefore produce 3 N of force in a 3 Tesla field (7 N in a 7 T magnet). The power dissipated in the coil (eqn. 2) depends on the resistivity ρ of the material used, the current I through the coil and the length and cross-sectional area of the wire.

$$P = I^2 R = I^2 \frac{\rho l}{A} \qquad (2)$$

With 10 turns of copper wire ($\rho = 1.7 \times 10^{-8}$ Ωm at 20 °C), arranged in a rectangle 10 × 20 cm (Fig. 1a), the cross-sectional area is approximately $A \approx 0.785$ mm^2 and the resistance R is calculated to be about 1.3×10^{-1} Ω. Therefore, the power dissipated at 0.5 A is about 3.25×10^{-2} W. This is a small amount of power, and while the coil could warm up if a constant high current was to be passed through it this is unlikely to occur very often.

Basic design

The coil itself was made of ten turns of copper wire 1 mm in diameter and was arranged in a rectangle measuring 20 × 10 cm and embedded in a clear Perspex case measuring 205 × 105 × 6 mm. The case was mounted inside a plastic pillar at one end (hereafter called the back end) which was attached to a solid plastic base, allowing current to pass through the coil and the case to pivot freely along this axis (Fig. 1). For support, two vertical plastic struts reinforced the central pillar. The back end of the case was also attached to a potentiometer powered by a 5V DC supply, which allowed measurement of coil movement. In order to provide a restoring force and keep the coil central when no current was passing through it, two elastic bands were used to pull the case section back to the midline.

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Figure 1 about here

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The front end of the case incorporated a small piece of rubber tubing into which subjects could insert their finger to allow the generated forces to act on them. A foam wrist rest was attached to the front of the base section so that subjects could comfortably rest their arm while their index finger was exposed to the forces. The base of the coil could be attached to the scanner bed through the use of two plastic 'feet' screwed on to runners on the underside of the coil. We designed the coil to be used on Philips 3T scanners and the experimental Philips 7T scanner at the Sir Peter Mansfield Magnetic Resonance Centre at the University of Nottingham. Results reported in this paper were obtained at 7T.

Implementation and calibration

Using a DAQ-6063E data acquisition card (National Instruments; Austin, TX, USA) along with the MATLAB Data Acquisition Toolbox (MathWorks; Natick, MA, USA) and a breakout box (Fig. 2), all data collection and experimental control could be performed directly within the MATLAB environment. The potentiometer in the base of the coil was powered by an interface on the breakout box and the voltage corresponding to the position of the coil was recorded using one of the DAQ card's analogue inputs. The coil was loaded with a 1 Ω resistor and the voltage across the analogue outputs from the DAQ card was stepped up using an amplifier so that the small voltage (a hardware maximum of 10 V) across the outputs of the card was translated to a larger voltage across the coil. The amplifier was an Amcron DC-300A Series II (Crown Amplifiers). It was used as a voltage amplifier with a peak 70V output, although any voltage-controlled voltage source or current source could be used. The DAQ card's sampling rate was set at 1000 Hz for both data acquisition and current generation.

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Figure 2 about here

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The potentiometer in the base of the coil was calibrated by using a MATLAB program to record the voltage across the potentiometer when the coil was held at five predefined positions. A linear regression across the resulting points produced a set of coefficients, which could then be used to relate voltage to coil angle. Preliminary tests revealed no additional noise caused by the MRI signal in the recorded voltage from the potentiometer.

The coil was designed so that subjects would be as comfortable as possible while using it. Since people have arms of differing lengths, the coil could be attached to the bed at any point along the bore of the scanner. However, the field was not uniform all the way down the bore so the same current passing through the coil would not produce a uniform force for all positions. Before each subject was studied, a short calibration sequence was performed that used the moment of inertia of the coil and the reading from the potentiometer in the base to produce an estimate of the acceleration and therefore the force on the coil when a predefined current was passed through it. It would have been easier to use an MRI-compatible force sensor to measure the force on the coil, but in the absence of this equipment the preceding method gave relatively accurate calculations of the force. See appendix A for calculations.

Testing for interference

One particular issue that we were concerned about was that running a current through a coil near to the head might interfere with the functional images collected. The change in the z-component of the magnetic field produced at the head when a current is passed through the coil can be approximated using the Biot-Savart law to equation 3.

$$\Delta B_z \approx \frac{3\mu_0 InlxzW}{4\pi\left(x^2+z^2\right)^{5/2}} \qquad (3)$$

Here, l is the length of the vertical wire of the force coil nearest to the head (0.2 m); W is the distance between this wire and the pivot (0.1 m); x is the perpendicular distance from the force coil to the axis of the bore (about 0.3 m); z is the distance along the bore between the force coil and the head coil (about 1 m); and all the other symbols have their usual meanings. At a current of 0.5 A, the resulting magnetic field change is about 7.25×10^{-9} T, corresponding to a frequency shift of less than 1 Hz, much less than the line width of the water response. This implies that the simple act of passing current through the coil should not affect the functional images obtained.

To make sure of this however we ran a series of experiments in which functional images were collected while the coil was in various modes of operation. Functional imaging of a phantom was performed using multislice gradient-echo echo planar imaging (TR = 2200, TE = 25 ms) on a Philips 7T MR scanner with a SENSE head coil. Acquisition parameters included a field of view of 19.2 cm and an acquisition matrix of 64 × 64 (15 slices, 3 mm thickness, 0.5 mm gap). We collected 10 volumes of images while the coil had a constant current of 780 mA passing through it to compare against 10 volumes when the current was switched off. We also collected 24 volumes using the paradigm described below for comparison to the *in vivo* images.

In vivo example

5 male and 5 female healthy adult subjects aged 22-32 years with no known neurological or motor defects were recruited to this study after completing a standard fMRI safety form from the Sir Peter Mansfield Magnetic Resonance Centre and giving their informed consent. The fMRI paradigm was designed to identify brain regions responsible for force prediction and reaction in a task where subjects were instructed to resist the forces that the coil applied to their right index finger. Subjects experienced 96 blocks (8 volumes per

block) each lasting 17.6 seconds, of which the first 4.4 seconds consisted of a sequence of forces up to 3 N applied to the finger and cued by a tone, while in the remaining 13.2 seconds they received no force. The 24 volumes (3 blocks) of phantom data were acquired in the same way – the coil was activated for the first 4.4 seconds in each block and deactivated thereafter.

The volunteer data were collected as for the phantom data described above, but with the addition of a single volume whole-brain 'anatomical' T2* image (TR = 26 ms, TE = 10 ms, FOV = 25.6 cm, acquisition matrix 256 × 256, 100 slices, 3 mm thickness, 0.5 mm gap). Functional imaging data sets for both the *in vivo* data and the phantom images were pre-processed using the SPM5 software package (Friston et al., 1995). The data sets were realigned and unwarped using the first image as a reference with 4 mm separation between points sampled in the reference image and 2nd-degree B-spline interpolation. Following this, the functional images were coregistered to the single-volume image. The functional images were then normalised to MNI space and smoothed using a Gaussian kernel with a 4 mm FWHM (full-width at half-maximum).

The volunteer and phantom data were analysed using SPM5's general linear model. For the phantom data, the following comparisons were made: a) in the coil on/coil off test, the 10 'on' volumes to the 10 'off' volumes; and b) in the test mimicking the *in vivo* study, the 6 'on' volumes to the 18 'off' volumes. For the volunteer data, we compared the volumes during which subjects experienced a force sequence to those where they experienced no force. These statistical contrasts were used to create an SPM{t}, which was transformed into an SPM{Z} and then thresholded at $p < 0.001$. We used a canonical HRF plus time derivative for both phantom and functional data to make sure that any activation found was closely modelled on what we would actually find in neuronal activity. The design matrix was estimated for individual subjects and a group random-effects analysis was performed for the various contrasts. Significant foci of activation were extracted in MNI coordinates to determine the approximate brain region in which the activation took place.

Results

In the coil on/off test for the phantom data, we found a few 'activated' voxels when the realignment parameters were used as regressors ($p < 0.001$ uncorrected). These vanished when either the family-wise error (Bonferroni) or false discovery rate (Genovese et al., 2002) correction were applied at $\alpha = 0.05$. Similar numbers of voxels (between 15 and 16) appeared to be activated in both the 'on' vs. 'off' comparison and the 'off' vs. 'on' comparison. In the same way, for the test mimicking the *in vivo* study, there were similar numbers of 'activated' voxels (between 2 and 4) found in each comparison at $p < 0.001$ uncorrected. Furthermore, signal-to-noise ratio calculations showed no significant difference between 'on' and 'off' blocks. These results suggest that the activations were not caused by the coil but were instead more likely to be systematic errors due to the sensitivity of the phantom, and would most likely be subsumed in the background noise of a normal, active brain. This was confirmed by data from significant activations in the ‘on’ condition alone and the ‘off’ condition alone. The majority of the activations were single voxels with occasional clusters of no more than two or three activated voxels.

Figure 3 illustrates the results from the *in vivo* experiment. Sample behavioural data from one force sequence for a single subject is shown in figure 3a. The sample was collected over 1800 ms and transformed from volts to degrees using the procedure outlined above. Positive values are away from the body, to the right of the subject; negative values are towards the body.

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Figure 3 about here

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The statistical maps (Fig. 3b) were corrected for false discovery rate at $p < 0.01$ and for cluster extent at $p < 0.05$. Table 1 shows the main foci of activation for the force vs. rest comparison. Foci of activation were present in bilateral primary motor cortex (Brodmann area 4), right premotor cortex (Brodmann areas 6 and 9), supplementary motor area (Brodmann area 6), bilateral posterior parietal cortex (Brodmann areas 40 and 7) and anterior cingulate cortex (Brodmann area 32). This represents increased activations in these areas when subjects experienced a force on their finger compared with no force.

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Table 1 about here

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Discussion

In this paper we have introduced a novel piece of equipment for providing forces to the finger during functional magnetic resonance imaging experiments. The force coil has a simple design, is cheap and easy to assemble and has the potential for use in many different kinds of investigations. The coil can be controlled by any computer system capable of sending and receiving analogue voltages and is very easy to integrate into MATLAB or other data acquisition environments. Despite concerns about the field generated by the coil interfering with the functional images collected, we have shown that this is not an issue at least at the low levels of current required to provide the levels of force which the finger can resist. Apparent activations in the phantom were most likely systematic errors in the calculation of statistical maps, and these spurious activations disappeared when parameters were estimated in a paradigm mimicking that of the *in vivo* example. This system is similar to a device proposed by Riener and colleagues (Riener et al., 2005), who used a dual coil coupled with a force sensor to likewise provide a 1-DOF force output to the hand. In this technical note, however, we have also demonstrated one possible use of the coil in providing forces to the finger.

The *in vivo* results show large, significant activations in bilateral primary motor cortex, as well as supplementary motor area (SMA) and right premotor cortex. Anterior cingulate cortex (ACC) and bilateral posterior parietal cortex (PPC) were also activated. Since the right index finger was exposed to the force sequence, the contralateral M1 activation was expected as this area is known to be activated in the production of movement (Ashe et al., 2006; van Oostende et al., 1997). Similarly, the ipsilateral activation in this kind of task is well documented (Verstynen et al., 2004). The premotor and supplementary motor area activation is also expected as these are known to be active in the learning of motor sequences (Muller et al., 2002). Many motor sequence studies have found activations in the posterior parietal cortex (Jenkins et al., 1994; Sakai et al., 2002) and this is generally thought to be due to its role in spatial attention and the coding of actions (Fogassi and Luppino, 2005). The ACC is traditionally thought to be active in error detection and the monitoring of conflict (van Veen et al., 2001), which makes sense in this case as subjects have to monitor the incoming force on their finger and produce a countering force accordingly.

The current limitations of the coil include the fact that when the coil is not directly perpendicular to the static magnetic field of the magnet the force passing through it will vary depending on the angle to which the coil is held in the field. This was not an issue in the *in vivo* study discussed above due to the simple instructions to the subject to resist the applied force and keep the coil as central as possible, but it may have implications for future studies. However, this limitation can be addressed by dynamically recording the position of the coil at all times and adjusting the current on-line to provide the correct amount of force. Similarly, the fast switching gradient field produces small amounts of back EMF in the coil that could potentially interfere with the generated forces, but this effect is not very large and a dynamic programming system taking this into account would nullify it completely. Another issue is that the device cannot be used outside the scanner for testing or for training subjects, as the operation of the coil is contingent on it being inside a large magnetic field.

It is a relatively simple matter to change the dimensions of the coil and therefore make it fit more easily into the bore of the magnet. Since there is limited space, we did find on occasion that the coil caught on the edge of the bore as it curved over the top of the device. Halving the height of the coil and doubling the number of turns would solve this problem. Another possible improvement of the design would be the implementation of a track for the coil to move on, which would enable subjects to make reaching movements rather than just finger movements. There would again be issues with the changing B-field at different points along the bore but this could be dealt with using dynamic programming, as described above.

Conclusion

The force coil is a cheap, easy to assemble device which provides forces to the finger inside the MRI scanner without causing undue interference to the functional images collected. The coil has been used successfully in one study to produce a measurable force sequence. Designs based on this principle are likely to prove useful in studies of motor learning using fMRI.

Appendix: calculating the force on the coil

To calculate the force on the coil at a particular voltage, we measured the angular deflection of the coil using the potentiometer for periods of one second following application of a constant 780 mA current to the device. This was repeated five times and the maximum angular acceleration was calculated by taking the second derivative of angular position for each measurement, using a fourth-order Butterworth dual pass filter with a frequency cutoff of 20 Hz to smooth the data. The median of these values was taken to find the angular coil acceleration for the particular voltage sent to the coil.

The moment of inertia of an oblong block pivoted on the corner of one of its longest edges is defined as

$$I = \frac{1}{3} m \left(W^2 + \frac{1}{4} D^2 \right) \qquad \text{(A1)}$$

with mass m, width W and depth D. The mass of the coil was determined to be 200 g ± 5 g. With W = 105 mm and D = 6 mm the moment of inertia of the case section was calculated as $I \approx 7.4 \times 10^{-4}$ kg m^2. The torque on the end of the coil where the subject's finger is placed is given by

$$\tau = \mathbf{W} \times \mathbf{F} \qquad \text{(A2)}$$

where $\boldsymbol{\tau}$ is the torque, $W = |\mathbf{W}|$ is the width of the coil and $\mathbf{F}$ is the force applied. Since the initial force is perpendicular to the coil (as the coil is initially parallel to the bore of the magnet and thus perpendicular to the field), the magnitude of the torque is simply FW. The torque can also be defined as

$$\tau = I\alpha \qquad \text{(A3)}$$

where α is the angular acceleration and I is the moment of inertia of the coil. Thus substituting equations A1 and A3 into A2 we produce

$$F \approx \frac{I}{W}\alpha \qquad \text{(A4)}$$

and we can thus use the moment of inertia, angular acceleration and width of the coil to calculate the approximate force on the subject's finger at the initiation of the force pulse. Theoretically the force produced in a 7T magnet will be around 14 N for every amp of current, but this is limited by the heating effect of the current, the physical robustness of the coil and the amount of interference with the functional images that a higher current would cause. More testing is required to optimise the coil for dealing with these issues and to determine how precise and accurate the force production of the coil is.

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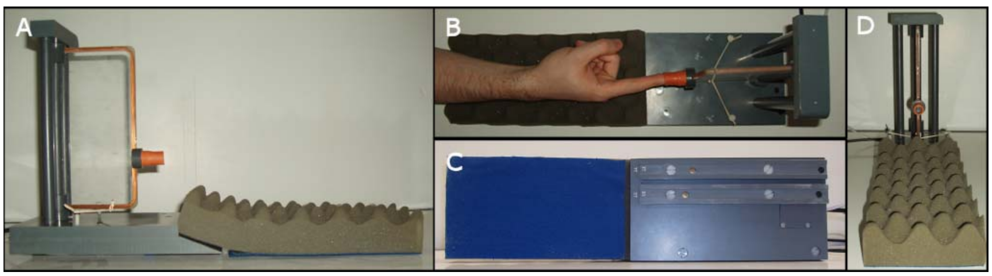


Fig. 1 Photographs of the force coil. **a** Side elevation showing coil, finger guard and wrist rest. **b** Top elevation showing position of wrist on rest, and with restoring bands and marked angles visible. **c** Bottom elevation showing tracks for 3T and 7T magnets. **d** Front elevation showing supports on either side and finger guard.

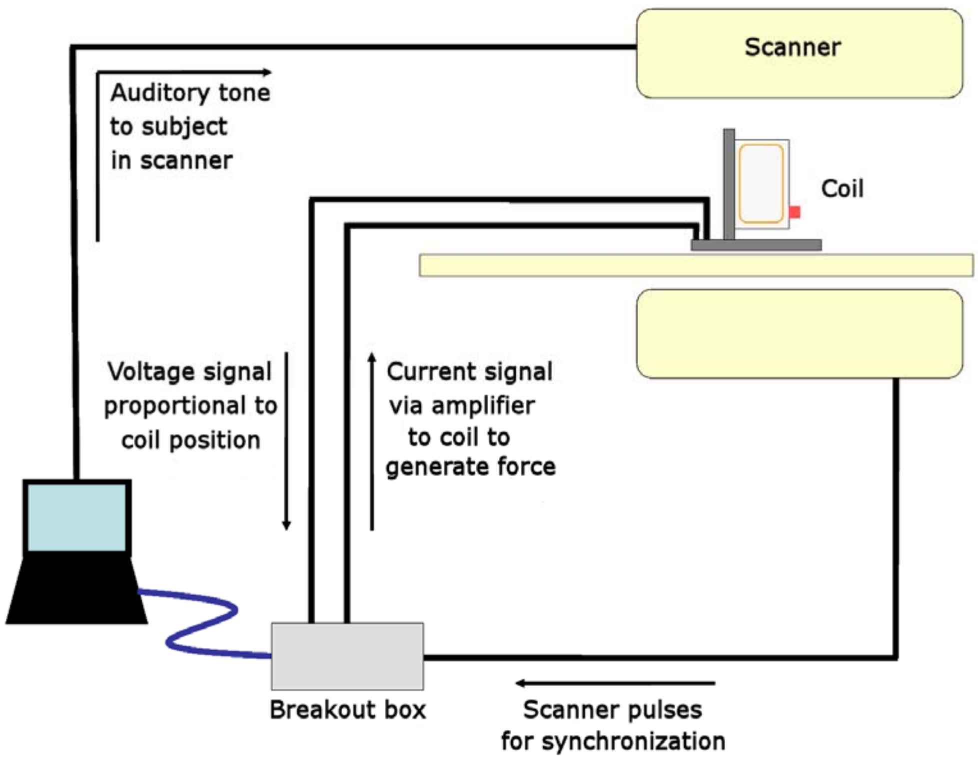
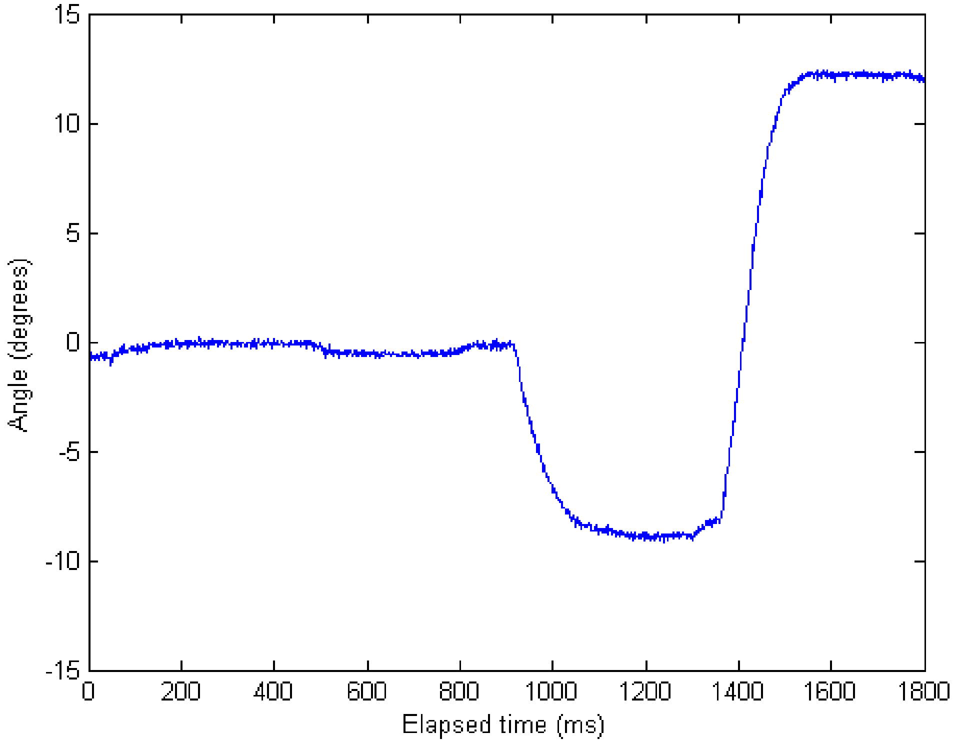


Fig. 2 Experimental setup and sample data. A laptop running MATLAB controls the session. To time the force sequences, pulses from the scanner are read by a digital input on the breakout box and recorded using a NI-DAQ card. The laptop sends a tone to the subject in the scanner at the start of each force sequence and analogue voltages are sent to the coil via a current amplifier to produce these forces. During the sequence the position of the potentiometer on the coil is recorded.

A



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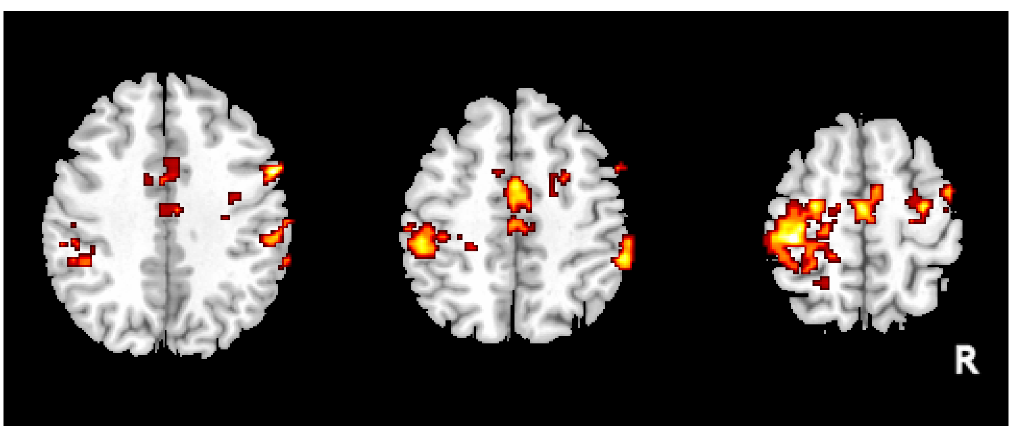

Fig. 3 Sample data from the *in vivo* experiment. **a** Behavioural data from a single force sequence for a single subject. Over 1800 ms the subject's finger position varies from -10°

to 15°. **b** Functional imaging data across all ten subjects in a random effects analysis. Two axial slices displayed in neurological convention, FDR corrected ($p < 0.01$) and cluster extent corrected ($p < 0.05$). Activations correspond to trials where subjects experienced a force sequence versus rest and show bilateral M1, SMA, ACC, and bilateral posterior parietal and premotor cortex.

Table 1: Regions showing significant activation for force > rest contrast

No. voxels	Anatomical region	BA	MNI coords			T-value	Z-score	p_{voxel}	$p_{cluster}$
			x	y	z				
86	Right premotor cortex	9	54	6	40	12.75	5.04	0.002	< 0.001
1097	Left primary motor cortex	4	-38	-30	58	12.35	4.99	0.002	< 0.001
547	Supplementary motor area	6	0	-14	62	10.72	4.75	0.003	< 0.001
195	Right premotor cortex	6	32	-12	62	10.07	4.65	0.003	< 0.001
226	Right inferior parietal lobule	40	56	-38	48	9.72	4.59	0.003	< 0.001
33	Right anterior cingulate cortex	32	2	4	38	6.94	3.99	0.005	0.006
40	Right premotor cortex	6	24	4	48	6.80	3.95	0.005	0.002
31	Left superior parietal lobule	7	-20	-54	64	6.70	3.92	0.005	0.008
35	Right primary motor cortex	4	28	-18	42	6.51	3.87	0.006	0.004
30	Left anterior cingulate cortex	32	-10	6	46	6.44	3.85	0.006	0.009

Note. Overview of regions showing significant activation for the force > rest contrast, FDR corrected at $p = 0.01$ and cluster extent corrected at $p < 0.05$. Shown are the number of voxels in each cluster; anatomical region; Brodmann area; x, y and z coordinates in MNI space; T-values; Z-scores; voxel-level p-values; and cluster-level p-values.

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