Novel magnetomechanical MR compatible vibrational device for producing kinesthetic illusion during fMRI

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Purpose: Functional MRI (fMRI) can provide insights into the functioning of the sensorimotor system, which is of particular interest in studying people with movement disorders or chronic pain conditions. This creates a demand for manipulanda that can fit and operate within the environment of a MRI scanner. Here, the authors present a magnetomechanical device that delivers a vibrotactile sensation to the skin with a force of approximately 9 N.

Methods: MRI compatibility of the device was tested in a 3 T scanner using a phantom to simulate the head. Preliminary investigation into the effectiveness of the device at producing cortical and sub-cortical activity was also conducted with a group of seven healthy subjects. The vibration was applied to the right extensor carpi ulnaris tendon to induce a kinesthetic illusion of flexion and extension of the wrist.

Results: The MRI compatibility tests showed the device did not produce image artifacts and the generated electromagnetic field did not disrupt the static magnetic field of the scanner or its operation. The subject group results showed activity in the contralateral putamen, premotor cortex, and dorsal lateral prefrontal cortex. Ipsilaterally, there was increased activity in the superior and inferior parietal lobules. Areas that activated bilaterally included the thalamus, anterior cingulate, secondary somatosensory areas (S2), temporal lobes, and visual association areas.

Conclusions: This device offers an effective tool with precise control over the vibratory stimulus, delivering higher forces than some other types of devices (e.g., piezoelectric actuators). It can be useful for investigating sensory systems and sensorimotor integration. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4824695>]

Key words: echo planar imaging, MRI, sensorimotor, brain, kinesthetic illusion

1. INTRODUCTION

Functional MRI (fMRI) is becoming increasingly popular as a means to investigate brain functionality in both research and clinical settings. This creates a demand for MR compatible technology that can be used during imaging to produce particular responses in the brain. Many devices have been created to deliver a range of stimuli (e.g., visual, sensory, cognitive) and also to record responses during imaging to later correlate with the fMRI data “offline” (e.g., eye movements, decision choices). Translating sensorimotor tasks from the lab to the MR environment is particularly difficult as it typically involves large manipulanda (often constructed with MRI incompatible elements) linked to robots or electrical sensing devices, e.g., Refs. 1–4. A number of MRI compatible sensorimotor devices have been constructed and include hydraulic and pneumatic systems,^{5–8} electric actuators (e.g., electrical stimulation, piezoelectric bender elements),^{9–11} and magnetomechanical devices.^{12,13} These are often very complex designs and can be costly to reproduce.

The main design considerations for such devices include:

1. the device must fit within the confined space of the MR scanner,
2. the device must work without the magnetic field causing deleterious effects to the device’s operation,
3. the operation of the device should not influence the magnetic field or degrade image quality,
4. safety should not be compromised, e.g.,
 - a. ferromagnetic materials becoming projectiles in the magnetic field,
 - b. the radiofrequency pulse inducing eddy currents in conducting materials producing an electric current and heating effects.

Here, we present a simple magnetomechanical device based on the Lorentz force law of electromagnetic theory which can produce a robust and adjustable vibrotactile sensation to a person’s right wrist in the MR environment. The purpose of designing the device was to separate motor proprioception from executive motor command function signals by using a kinesthetic illusion.^{14–16} Kinesthesia is the perception

of position and movement of a body part which arises from afferent nerve signals generated by muscle spindles.^{17–20} By stimulating a tendon at a particular frequency it is possible to generate nerve signals that are interpreted by the brain as the sensation of movement but without physically stretching the muscle. The amount of illusory movement can be quantified by measuring the angle that a subject “feels” the limb has moved through by a physical demonstration with the same or opposite limb. The frequency range that produces a kinesthetic illusion is approximately 40–120 Hz. The greatest illusion of movement, or the greatest angle of illusory motion felt, occurs around 70–80 Hz.^{14,16–18} Therefore, by applying a stimulus of this frequency range to the extensor carpi ulnaris tendon an illusion of the wrist flexing is induced and extension is felt when the vibration is discontinued.

Prior to scanning subjects the device was assessed for MRI compatibility to ensure conditions 1–4 were met. Then a functional MRI study was conducted with healthy subjects to test the effectiveness of the device at inducing the kinesthetic illusion and producing motor-related activity in the cortex and basal ganglia.

2. MATERIALS AND METHODS

2.A. The device

The device was designed to be inexpensive and easy to set up in the MR environment. It was based on the Lorentz force law of electromagnetic theory which describes how electromagnetic forces interact. When a straight current-carrying wire is placed in a magnetic field at any angle not collinear with the direction of the magnetic field, it will experience a force perpendicular to the direction of the magnetic field and the current. If the straight wire is exchanged for a coil of wire then the direction of the forces acting on it will vary as the direction of the current changes around the loop. These forces produce a torque, τ , on the coil, described by

$$\tau = nIA \times \mathbf{B}_0,$$

where n is the number of turns in the wire, I is the current, A is the cross-sectional area of the coil, and $\mathbf{B}_0$ is the static magnetic field. The magnitude of the torque is dependent on the angle ϕ between the planes of $\mathbf{A}$ and $\mathbf{B}_0$.

It is possible to use the torque to produce a mechanical motion by securing the coil at one end, creating a point about which the coil will pivot when the forces act on it. When an alternating signal, for example, a sine wave, is passed through the coil an oscillatory mechanical motion is produced as the direction of the forces will reverse as the direction of the signal changes. This can be transformed into a vibrating stimulus by adding an interface to contact with the skin. The force delivered to the skin is given by

$$F = \tau/r \sin \theta,$$

where r is the distance from the pivot to the point that contacts with the skin and θ is the angle between r and F .

The device coil was constructed of 100 turns of 43 AWG wire with a radius of 3.2 cm. Attached in the center of the coil was a nylon rod that protruded a short distance from the coil

FIG. 1. Magnetomechanical device. The coil (a) is attached to a frame and is positioned above the wrist. The mechanical oscillatory motion interfaces with the skin via the nylon rod (b). The hand is strapped into a molded splint and the position of the wrist and coil are adjustable.

to form the “tapping” point that contacted with the skin (see Fig. 1, marker B). This was located 5.5 cm from the pivot. A current of 0.5 A was used and the device was orientated so that θ and ϕ were both 90° . The force generated at the taper was calculated to be approximately 9 N in a 3 T magnetic field. The amplitude of the vibration was measured to be 0.6 mm peak to peak at 12.5 Hz and 0.2 mm at 83 Hz unloaded (i.e., not in contact with the skin).

The coil itself was attached to a plastic frame that allowed the position of the contactor to be adjusted from person to person and the tendon accurately targeted. The frame was capable of 3 translational degrees of freedom and 1 rotational (pitch). Also attached to the frame was a molded splint to immobilize the wrist joint at a flexed angle of approximately 35° relative to the forearm. The splint was constructed in two sections, one for the hand and one for the forearm. Each section was wrapped around the limb and was firmly secured by Velcro straps. Adjustable struts joined the two sections and these were fixed in place once the splint was fitted to the hand.

The electric signal that was passed through the coil was derived from an audio signal. It consisted of a modulating sine wave with a period of 12 s, the frequency was linearly modulated between 12.5 and 83 Hz.

The audio signal was played back on a computer located in the MR scanner control room. The electrical output from the headphone port was connected to a LED converting the signal to a series of light impulses. These were routed along a fiber optic cable into the scanner room via a wave guide in the wall. The purpose of the fiber optic cable was to reduce electromagnetic interference being transferred from the control room to the device and scanner room, preventing degradation of the MR images and deleterious operation of the device. The fiber optic cable was connected to an optoisolator in the scanner room which picked up the light impulses and converted it back into an electrical signal which was then passed along tinned copper wire to the coil. The copper wire contained multiple common-mode traps to suppress the induction of high frequency currents created by the RF and gradient field pulses typical of MRI.

2.B. Experimental design

2.B.1. Device testing

Prior to scanning subjects, the device was tested in the MR imager. The device was positioned approximately 0.4 m from the base of a head coil to simulate the position of a subject's hand during imaging. A 32 channel transmit/receive volume head coil was used with a MRI phantom to simulate the head (coil manufactured by Quality Electrodynamics, Mayfield, OH, USA).

A series of T_2^* echo planar images with BOLD contrast were acquired using a Siemens whole-body Magnetom Verio 3 T magnetic resonance scanner with a 70 cm wide bore. A 128×128 matrix was used with a $256 \times 256 \text{ mm}^2$ FOV producing a $2 \times 2 \text{ mm}^2$ in-plane resolution. Four mm thick slices were acquired in an interleaved fashion with 30 slices per volume and an interslice gap of 0.5 mm, TR = 3000 ms, TE = 29 ms, and flip angle = 80° . The fMRI was carried out four times—(1) without the device in the room, (2) with the device in the room but not connected, (3) with the device connected but not running, and (4) with the device running. For each condition a total of 40 volumes were collected to evaluate the temporal SNR. In addition, an anatomical T_1 weighted spoiled gradient echo image with inversion recovery was also acquired for each condition consisting of 176 slices with a voxel resolution of 1 mm^3 , TR = 1900 ms, and TE = 1.83 ms.

The playback of the audio wave was synchronized with the acquisition of the MR images using E-prime 2.0 software (Psychology Software Tools Inc., PA, USA). The scanner sequence was programmed to output a trigger pulse at the beginning of every scan volume and this was picked up by a serial response box and relayed to the software. The audio signal was constructed to begin at the minimum (12.5 Hz) and increase to 83 Hz before returning back to the starting frequency 12 s later. When used with subjects this produced the effect that as the frequency increased and decreased the volunteers felt the illusion of their wrist slowly flexing and extending.

2.B.2. Subjects

Seven volunteers, five males and two females, with no history of neurological disorders were recruited, with mean age 32 yr and range 26–38 yr. Ethics approval was obtained from the Institutional Review Board at University Hospitals, Cleveland, OH, USA prior to consenting participants.

Subjects were positioned head first supine on the MR scanner table. Foam blocks were packed around each subject's head inside the volume coil to minimize movement during data acquisition. The device was attached securely to the hand and forearm and the contactor was positioned over the extensor carpi ulnaris tendon of the right wrist. The arm was then positioned comfortably across the subject's abdomen with padding underneath for support and to reduce vibrations traveling to other parts of the body. Any stray vibrations that transferred from the frame were very weak and were not directly over a tendon so they did not contribute to the move-

ment related activations. In the fMRI signal they would most likely exist as a continuous sensory activation and the signal contributions would then be filtered out in the statistical modeling. The positioning of the contactor over the wrist tendon was checked prior to collecting the fMRI data by allowing subjects to experience a few minutes of the device working once they were positioned in the scanner. Subjects were instructed to attend to the stimulus and to keep their eyes open during imaging but not to look at their wrist. The mirror on the head coil was angled away from the body so that they viewed the wall behind their head. The sine wave was repeatedly presented in a contiguous manner for 8 min (160 volumes) during image acquisition allowing the subject to experience 40 illusions of flexion and extension. An anatomical image was also acquired, with MRI acquisition parameters as described in Sec. 2.B.1.

Immediately at the end of the fMRI scans the subjects were asked to demonstrate the maximum angle they felt their wrist had moved to during the illusion. This was measured using a protractor and their comments about how the illusion felt were also noted.

2.C. fMRI data processing

Functional MRI data was preprocessed using BrainVoyager QX software version 2.4 (Brain Innovation, Maastricht, Netherlands). Cubic spline slice time correction was applied to each volume in addition to trilinear/sinc interpolation for 3D motion correction. The data were spatially smoothed using an 8 mm full-width-half-maximum Gaussian kernel. High pass temporal filtering (288 s) was also performed. The images were then coregistered to the subject's own anatomical image. The coordinate system was set such that the anterior commissure was marked as the origin and the anterior/posterior planes were aligned to the anterior/posterior commissure plane. The subjects own space was then warped to Talairach space.

There is a latency between 2 and 5 s between the illusionary frequencies being applied to the tendon and the feeling of movement being perceived by the subject.^{14,18,21} The duration of the latency varies from person to person and without additional testing and feedback from the subjects it was not possible to know the exact point in time which the illusion was felt; therefore, the data were analyzed using a phase analysis method. This method is typically used in retinotopic studies to map activations in the visual cortex due to a changing stimulus.^{22,23} The stimulus was approximated to a box car function of 6 s of flexion illusion (12.5 → 83 Hz) and 6 s of extension illusion (83 → 12.5 Hz). The resulting waveform was convolved with the hemodynamic response function (HRF) to produce a model of the expected response. A phase analysis was then performed producing phase maps of the activations. The phase maps showed at which time point in the repeated 4 TR stimulus sequence each subject experienced the kinesthetic illusion. The Pearson *r*-values were converted to *t*-scores before a random effects one sample *t*-test was performed.

3. RESULTS

3.A. Device testing

The equipment tests showed the device operated safely in the MR environment, without degradation to the temporal SNR and images were free of artifacts. The small electromagnetic field created by the current flowing through the coil was not sufficient to interfere with the magnetic field of the scanner or its operation. The images were also visually inspected for artifacts but none were found. It was not possible to obtain the K-space data at the time of data collection and so the images inspected were the reconstructed images exported from the scanner before any processing was carried out. The mean temporal SNR of the data was calculated from the mean of the voxel time series divided by the standard deviation of the voxel time series and it showed no difference between the conditions. The signal was determined from a slice close to the device as this would be most likely affected by noise. The temporal SNR for each condition was calculated as (1) without device in the room: 4.47 ± 0.006 , (2) with device disconnected: 4.45 ± 0.04 , (3) with device connected: 4.44 ± 0.03 , and (4) with device running: 4.43 ± 0.03 .

A phase analysis was performed on the fMRI data from the four equipment tests. Some statistically significant clusters of activity were detected in each condition at a threshold of $P \leq 0.001$ (uncorrected). The active voxels were randomly distributed and none survived a higher Bonferroni (family wise error (FWE)) threshold of $P \leq 0.05$.

3.B. Subjects

During the initial positioning of the contactor over the tendon 6 of the 7 subjects reported feeling the illusion. At the end of the scan, the same subjects reported feeling the illusion for the duration of data collection. Although the duration of the session was long for this type of stimulus no subjects reported that the illusion was reduced by the end of the scan. Four of the 7 subjects demonstrated the angle through which they felt their wrist was bending and the measurements ranged from 30° to 40° . The fMRI data for the one subject that did not feel the illusion was inspected and activations in the motor network were still demonstrated; therefore, the subject's data was included in the group analysis.

Individually, the results of the phase analysis showed that all 7 subjects experienced a latency in the range of 3.0–5.9 s between the presentation of the illusion inducing frequencies and the resulting movement related brain responses. The stimulus was presented in a regularly repeating fashion over 4 TR and constructed so that the high frequencies occurred during the second and third time periods; however, the activations were present during the third and fourth phases.

A group random effects t -test performed on the data of all seven subjects revealed sensorimotor related activity in the basal ganglia and cortex. These results are shown in Fig. 2 overlaid on axial and sagittal slices of a template brain,²⁴ aligned to the AC–PC plane and warped to Talairach space, threshold $P \leq 0.05$ (FWE). The foci of the activations

FIG. 2. Activations from the seven subject group random effects t -test, threshold $P \leq 0.05$ (FWE), overlaid on a template brain (Ref. 24). Axial slices through the brain shown at the levels indicated, and a sagittal slice at $X = 3$. The areas indicated by white circles from the top left are: (A) bilateral superior temporal cortex (BA 38), (B) contralateral putamen and thalamus and medial prefrontal cortex (BA 10), (C) bilateral S2 (BA 37, 39, and 40) and contralateral dorsal lateral prefrontal cortex (BA 9, 10, and 46), (D) inferior parietal lobule (BA 7), and (E) thalamus and anterior cingulate cortex (BA 24).

are also listed in Table I with their respective t -scores and cluster sizes.

Increased activity was detected contralaterally in the putamen (involved in movement selection), premotor cortex

TABLE I. Coordinates of the center of gravity of the locations of the activity detected in the seven subject group random effects one sample t -test, threshold $P \leq 0.05$ (FWE). All positions are given in Talairach coordinates.

Anatomical region	X	Y	Z	Cluster size, mm ³	Peak T score
Ipsilateral					
<i>Middle Temporal Gyrus</i>					
BA 19	38	-62	18	125	756
BA 22	61	-38	8	1054	343
BA 37	51	-59	6	2527	275
BA 21	63	-23	-9	160	195
BA 39	50	-65	18	113	132
<i>Transverse Temporal Gyrus</i>					
BA 41	54	-19	15	892	526
<i>Superior Temporal Gyrus</i>					
BA 22	62	-17	3	117	291
BA 38	36	7	-17	550	275
<i>Supramarginal Gyrus</i>					
BA 40	53	-50	21	114	202
BA 40	59	-47	21	103	182
<i>Middle Occipital Gyrus</i>					
BA 37	49	-65	-10	291	215
<i>Middle Frontal Gyrus</i>					
BA 46	47	40	3	96	150
<i>Inferior Parietal Lobule</i>					
BA 7	36	-58	42	189	179

TABLE I. (Continued.)

Anatomical region	X	Y	Z	Cluster size, mm ³	Peak T score
<i>Superior Parietal Lobule</i>					
BA 7	34	-67	44	631	361
<i>Precuneus</i>					
BA 7	14	-80	48	107	159
<i>Cuneus</i>					
BA 18	5	-98	4	193	319
<i>Anterior Cingulate</i>					
BA 24	2	32	-6	176	190
<i>Basal Ganglia</i>					
Thalamus	3	-17	9	425	190
Contralateral					
<i>Middle Temporal Gyrus</i>					
BA 39	-58	-57	8	2437	323
BA 39	-43	-65	12	117	289
BA 21	-52	-25	-10	227	184
<i>Middle Occipital Gyrus</i>					
BA 19	-42	-86	9	298	167
<i>Precentral Gyrus</i>					
BA 13	-50	-10	13	1865	296
BA 6	-56	3	20	390	280
<i>Insula</i>					
BA 13	-41	-2	7	16	22
<i>Inferior Frontal Gyrus</i>					
BA 47	-32	26	1	586	238
<i>Superior Temporal Gyrus</i>					
BA 41	-46	-28	14	7264	394
BA 38	-33	2	-20	63	24
<i>Middle Frontal Gyrus</i>					
BA 10	-36	43	17	1118	313
BA 10	-25	52	3	94	147
BA 8	-19	20	39	17	25
BA 9	-30	23	32	10	22
<i>Medial Frontal Gyrus</i>					
BA 10	-16	54	3	12	23
<i>Anterior Cingulate</i>					
BA 32	-1	48	-2	393	310
<i>Basal Ganglia</i>					
Putamen	-21	8	0	1290	576
Thalamus	-13	-14	0	116	174
Thalamus	-10	-23	9	76	97

(Brodmann area (BA) 6-movement planning), and dorsal lateral prefrontal cortex (BA 9, 10, 46-movement planning) and ipsilaterally in the superior and inferior parietal lobules (BA7-sensorimotor integration). Bilateral activations included the thalamus (relays sensory and motor signals to the cortex), S2 (BA 37, 39, 40-sensory perception), anterior cingulate cortex (BA 32, 24-BA 24 projects to premotor cortex), medial prefrontal cortex (BA 10-sensorimotor cognition), the temporal pole (BA 38-spatial awareness), and the visual association

area (BA 19). Most of these areas are typically associated with the cognition and planning of movements. Additionally, at a slightly lower threshold of $P \leq 0.1$ (FWE) a contralateral primary sensory cortex (S1) activation in BA 3 was also detected.

4. DISCUSSION

The aim here was to test the effectiveness of a novel MRI compatible magnetomechanical device. The device tests showed that it operated safely without disturbing the main magnetic field or degrading the image quality. The *in vivo* tests showed that it was capable of inducing a kinesthetic illusion in the extensor carpi ulnaris tendon of the right wrist and the resulting cortical and subcortical activity was measurable with fMRI and has predominance in the contralateral hemisphere.

The device itself is inexpensive, small, and can be easily transported and assembled. It is also possible to mount the coil onto another frame and splint formed to accommodate a different part of the arm, enabling different tendons or nerve receptors to be stimulated. The need for the device to sit within the bore restricts its applications to the upper body limbs. The range or pattern of frequencies presented can also be manipulated, via the audio file, to produce either a simple vibrotactile sensation to target various properties of sensory receptors in the skin and muscles or to further investigate kinesthetic illusions.

Custom made and commercially available MRI compatible devices are often very expensive. The cost of all the materials for the device was less than US\$ 5000, with the MRI compatible materials for the frame that supports the hand and coil incurring the greatest expenses. A simpler design of the support frame could have been used reducing the costs. The electronics and the coil cost less than 10% of the final amount.

The operation of a magnetomechanical device is sensitive to its orientation in the magnetic field. This limitation can be overcome by securing the device, e.g., to the scanner bed. Another limitation is the inability to use the coil outside of the scanner room to train or test subjects as it relies on the presence of the magnetic field. A different type of device would be necessary for use in the lab.

The issue of any heating effects induced by the presence and operation of the device was not addressed during the device testing phase. As a transmit/receive head coil was used and the device and all wires were placed distant to it, it was assumed that no heating would occur. Furthermore, the wires were physically isolated from the subject. If a body coil is to be used for transmission of the radiofrequency pulse in future applications then further safety testing would be needed.

The *in vivo* studies showed the stimulus was effective at activating the contralateral premotor cortex, thalamus, and putamen, consistent with previous kinesthetic illusion imaging studies.^{15,25,26} At a slightly lower threshold of $P \leq 0.1$ (FWE) revealed a contralateral cluster of activation in S1 further supporting the stimulation of the sensorimotor network by the kinesthetic illusion. Although our protocol of continuous stimulation was unusual the stimulus changed regularly between the high illusion inducing frequencies and

the low sensory frequencies so the muscle spindles endings were not constantly producing movement signals. The clear activation of the basal ganglia using a motor task without associated confounding task effects could have useful applications in elucidating disease mechanisms in movement disorders such as Parkinson's disease and dystonia. Recent investigations have revealed conflicting results in these conditions, e.g., Refs. 27–33, and this may be due to the type of task used, differences in effort, and proprioceptive feedback.

Brodmann area 19 was bilaterally activated during the task. It is located within the extrastriate visual cortex and is associated with the visual processing area V3. This area has been shown to activate in another kinesthetic illusion study²⁵ and also during mental imagery studies involving a tactile task or movement.^{34,35} In the study reported here, during stimulus presentation subjects were unable to see their wrist but they were instructed to give their attention to how it felt. Although none voluntarily reported using imagery it is possible that subjects were imagining the movement as they felt the sensation of motion. This is also supported by the increased activity in the prefrontal cortex. In addition, it is possible the visual activations may have been detected due to the phase analysis method used. As with a GLM analysis, a regularly changing stimulus is detected and anything with a different period is filtered out. Here, the visual activations in the group analysis must have an on/off duration similar to the stimulus to be detected. What the phase analysis method allows is the detection of activations with the same period which do not coincide directly with the stimulus. Any such activation would be lost in a GLM analysis. If the phase analysis method was applied to other kinesthetic illusion studies it may reveal visual activations are regularly associated with the stimulus.

5. CONCLUSION

In summary, the device fulfilled the four criteria set out in the Introduction. It is inexpensive and fits within the MR scanner, it utilizes the magnetic field to generate mechanical motion, yet is sufficiently weak as not to disrupt the operation of the scanner, safety issues were addressed through using copper wires and plastic in its construction and common mode traps were used to suppress RF coupling during imaging. We have also shown the magnetomechanical device is effective at producing sensorimotor activity in the cortex and basal ganglia.

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