The Interaction between motor fatigue and cognitive task performance

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MR compatible strain gauge based force transducer

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(in revision)
Abstract
In order to evaluate brain activation during motor tasks accurately one must also measure output parameters such as muscle force or muscle activity. Especially in clinical situations where the force output can be compromised by changes at different levels of the motor system, it is essential to standardize the task or force level. We have therefore developed a magnetic resonance compatible force transducer that is capable of recording index finger abduction force and to display the produced force in real-time. This transducer is based on stain-gauges techniques and designed to measure both small and large forces accurately (range 0.7-60N) as well as fast force fluctuations. Experiments showed that the MR environment did not affect the force measurements or vice versa. Although, this transducer is developed for measuring index finger forces, detailed schematic diagrams are provided such that the transducer can easily be adapted for measuring forces of other muscle groups.

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Intrinsic hand muscles (including the first dorsal interosseus muscle, FDI) are often used for neurophysiological studies regarding motor function. Intrinsic hand muscles have a large cortical representation (see for review Porter and Lemon 1993), and they are relatively easy to activate by either transcranial magnetic stimulation or peripheral nerve stimulation. A special advantage of the FDI is the well defined and almost unique mechanical function (it is by far the most important muscle for the index finger abduction force; An et al., 1983); moreover, its muscle belly is very superficial and thus readily available to measure muscle activity using electromyography (EMG). Furthermore hand muscles are relatively easily to activate without head movements. All these properties make hand muscles, extremely suitable for studies involving measurements of cortical activity, such as functional magnetic resonance imaging (fMRI).

For an adequate interpretation of the cortical activity it is essential to record simultaneously muscle activity. However, it is challenging to record force or EMG from active muscles during fMRI studies (see chapter 5). A strong main magnetic field is present in the scanner room and furthermore, radio frequency (RF) and magnetic gradient fields (gradients) are induced during scanning. All may influence the force and EMG recordings.

A few labs have developed MR compatible force transducers (Cramer et al., 2002; Dettmers et al., 1996a; Ehrsson et al., 2000; Hidler et al., 2006; Kuhtz-Buschbeck et al., 2001; Liu et al., 2000, 2002a; Thickbroom et al., 1998). Some transducers use hydraulic or air pressure systems to measure force, and they are made of non-magnetic synthetic materials. A disadvantage of these materials is their high compliance, often making it impossible to measure small muscle forces or fast changes in muscle forces accurately. Studies that do use force measuring systems without hydraulic pressure systems (Cramer et al., 2002; Ehrsson et al., 2000; Hidler et al., 2006) do not describe their setup in much detail and it is therefore not possible to reproduce their setup. It was the aim of this study to design an fMRI compatible force transducer that was capable of measuring small and fast force fluctuations accurately. In addition we want to describe this system in such detail that the system could easily be adapted by other investigators.

Strain gauges are often used for force transducers that are used in a non MR surrounding. Force transducers based on strain gauges are very precise in measuring small and fast forces and they are mechanically simple to use (e.g. Boonstra et al., 2005). However, an environment with a strong magnetic field, radio frequency and changing gradient fields – such as a magnetic resonance imaging room – may influence
the strain gauge measurements. Moreover, transducers with strain gauges are often made of ferromagnetic metals which are incompatible with MRI. Nevertheless, we succeeded in designing an MR compatible force transducer with strain gauges. The transducer is capable of measuring small forces (range: 0.7-60N) and fast force fluctuations accurately. Although this transducer has especially been designed for the index finger, the system can easily be adapted to measure forces of other muscle (groups) as well.

**Material and methods**

*Force transducer*

This force transducer was designed to measure index finger abduction forces accurately for both small and large hands in an MR environment. The transducer was used in combination with an instrumentation amplifier (to convert the voltage signal into an optical signal), a power supply, an optical cable, a receiver, and a PC (with data acquisition interface) to store the data. All components will be described in more detail below.

The force transducer consists of several materials (see Fig. 1, close-up of the force transducer; the numbers in the text refer to the numbers in the figure). Strain gauges that are compensated for electromagnetic effects (Fig. 1 A; TML® MFLA-5.350-1L; Tokyo Sokki Kenkyujo Co., Japan) were placed in a full bridge (Wheatstone) configuration on an epoxy glass laminated bar (Fig. 1 B; Tufnol® 10G/40 20mm diameter; RS components number: 771-314; RS Components Ltd, United Kingdom). This material is strong and not easy to distort. Perpendicular to the bar, there is a PVC tube (Fig. 1 C), which is attached to the bar with a PVC clamp (Fig. 1 D). To hold the tube comfortably the tube is equipped with a bicycle hand grip (Fig. 1 E).

The force transducer can be adjusted to hand size in three ways. First, the grip can slide up and down the PVC tube to move the laminated bar to a height horizontally parallel to the index finger. If the grip is in the correct position the PVC tube is tightened by a copper screw in a titanium ring (Fig. 1 F). Second, the bar can be turned in relation to the PVC tube to position the bar parallel to the index finger. In the correct position this bar can be tightened by a brass screw (Fig. 1 G). And third, a delrin C-shaped connector (Fig. 1 H) can slide along the epoxy glass laminated bar to a position above the proximal interphalangeal joint of the index finger. After adjusting the force transducer to the subjects’ hand, the index finger is taped to the C-shaped connector (Fig. 1 I), the middle
and ring fingers are taped to the subject’s hand (Fig. 1 J), and the thumb taped to the fingers (Fig. 1 K), in order to fixate the position of the hand relative to the force transducer.

Figure 1. Photo of a hand holding the force transducer for measuring index finger abduction force. The force transducer consists of several components (for details see text): A. strain gauges; B. laminate bar; C. PVC tube; D. connector between tube and bar; E. hand grip; F. titanium ring plus screw to adjust height of hand grip; G. screw to adjust angle of the bar; H. C-shaped connector. The subject’s hand is taped to transducer in several ways: I. the index finger is taped to the connector; J. the middle and ring fingers are taped to the subject’s hand; and K. the thumb is taped to the hand as well. To transmit the signal, the following parts are also necessary: L. transmitter; M. (cable for) power supply; N. optical cable for signal transduction. To measure muscle activity simultaneously with force, one electromyography (EMG) surface electrode is placed on the belly of the first dorsal interosseus muscle (O) and one electrode is placed on the metacarpophalangeal joint (P).
The force applied to the bar will cause an unbalance in the strain gauge bridge. The output signal is amplified by an instrumentation amplifier (300x; Analog Devices AD8230; Analog Devices, USA). The amplifier is placed in a copper box (Fig. 1 L) to shield it from the RF field. The copper box is placed on the PVC clamp to keep the distance to the strain gauges as short as possible (to minimize scanner artifacts). The copper box also contains the transmitter system. This system consists of a voltage-to-frequency converter, a transmitter for the optical signal, and a voltage stabilizer (9V). Figure 2A shows a simplified scheme of the transmitter (see appendix I for detailed version).

**Figure 2.** Simplified schematical representation of the transmitter (A) and the receiver (B). The power of the transmitter is supplied by a lead acid dry battery (see Fig. 1'M'). The signal derived from the Wheatstone strain gauge bridge is amplified (AD8230, Analog Devices, USA), and converted to a frequency-modulated optical signal (AD654, Analog Devices, USA). This optical signal is transmitted (HFBR1522, Agilent Technologies, USA) via an optical cable (see Fig. 1'N') to the MR operator room. The optical signal enters the receiver via the optical signal detector (HFBR2522, Agilent Technologies, USA); the signal is then re-converted to voltage signal, linearly corresponding to the applied force (LM331N, National Semiconductor, USA), filtered (OP-07, Texas Instruments, USA), and amplified (OP-07, Texas Instruments, USA). The analogue signal is logged to a PC via Spike 2 (CED, Cambridge, UK).
The power of the bridge and transmitter is supplied by a lead-acid rechargeable battery (12V, 1.9Ah; Velleman ® Components N.V., Belgium). This battery, which is not attracted by the magnet, is placed in a separate aluminum box and connected to the copper box with a wire (Fig. 1 M). All electrical connections to and from the copper box are decoupled with feed through capacitors to protect the amplifier against RF effects.

A 10m long glass fiber cable (Fig. 1 N) is plugged into the optical transmitter of the box. This cable leaves the scanner room and enters the operator room through an RF waveguide in the RF cabin. In the operator room, the optical cable is connected to the receiver.

The receiver has an internal power supply which is connected to the 230V mains. In the receiver, the frequency-variable signal is readapted to a varying voltage signal. Thereafter, the signal passes a low-pass filter (-3dB at 200Hz) and an amplifier with a maximum variable gain of 6. Figure 2B shows a simplified schematic illustration of the receiver (see appendix II for detailed diagram). The output signal of the receiver is logged in a PC with Spike2 version5 for Windows, via an A/D converter (sample frequency: 500 Hz; CED, Cambridge; UK). This PC was connected to a beamer, which projected on a screen at the back of the scanner, providing the subjects with visual feedback of the force.

The delay of the setup of transmitter and receiver was determined by measuring the electrical step response; an electrical input was given to the transmitter, and the electrical output was measured at the receiver (see Fig. 3A). The delay, both of the positive and the negative step response, was 2 ms (Fig. 3B).

This total system can be used to measure forces of other muscle groups as well. An appropriate force transducer needs to be designed, and then the detailed scheme of the transmitter and receiver, shown in appendix I and II, can be used for the system.
Tests

To determine the linearity of the force transducer, we reconstructed a response curve by plotting the measured voltage against the force applied to the transducer. We held the transducer upside down when hanging calibrated weights (1-60 N) at the bar. This induced a force in the same direction as the voluntary index finger abduction force. The calibration was performed in both ascending and descending directions. This procedure was performed both in and out of the scanner room. The repeatability of the force measuring system was analyzed by measuring two weights 10 times successively.

Scanning was performed on a 3Tesla Philips MRI scanner (Best, the Netherlands) using the standard eight channel SENSE head coil as receiver and the body coil as RF transmitter. We used the following pulse sequence parameters: fast field echo (FFE) single shot echo planar images (EPI); 39 slices; slice thickness 3.5 mm; no gap; field of view 224 mm; scanning matrix 64x64; transverse slice orientation; repetition time (TR) = 2 s; echo time (TE) = 30 ms; minimal temporal slice timing (1957 ms); flip angle 90°. To test whether the force transducer system would influence the echo planar images, we scanned both a subject and a phantom with and without the presence of the force measuring system (three runs per scanning condition, only four runs of the subject without force transducer).

Furthermore, 9 right-handed subjects (mean age: 41.1 ± 10.7 years; 7 females) were asked to produce maximal voluntary index finger abductions, after signing an informed consent. The experiment was approved by the medical ethical committee of the University Medical Center Groningen, conform the standards set out in the Declaration of Helsinki (2000). To measure the maximal voluntary contraction (MVC) force, the force transducer was adjusted to the subjects hand size as described above. When the force transducer was adjusted, we taped the proximal interphalangeal joint of the index finger to the connector at the laminated bar to maintain contact between the finger and the transducer during relaxation (Fig. 1 I). Furthermore, the thumb, ring and middle fingers (Fig. 1 J and K) were also taped to the subjects own hand to prevent the subject from repositioning his hand in relation to the force transducer. The contractions lasted 8s (4 scans), followed by 52s rest, and were repeated three times.

To evaluate the MR compatibility of the force transducer it was checked whether the transducer emitted any RF radiation in the frequency range of 127.78 ± 0.44 MHz using a protocol provided by the scanner manufacturer (Philips, Best, Netherlands). The measurement to detect ‘spurious’ frequencies was repeated twice: once with an empty scanner bore (‘baseline’), once with a fully operational transducer on the scanner bed at
a location close to the usual hand position (‘force transducer’). In all cases, the body coil was used as receiver coil.

Electrically, the rise time of the transmitter and receiver was fast (only a delay of 2 ms); however, we also wanted to know whether the complete setup, including the transducer itself, could measure rapid force fluctuations. This was tested outside the MR room by hitting the laminated bar of the transducer with a hammer. The dampening of the recorded oscillations gives an indication of the frequency response of the force transducer (see Fig. 6).

**Analyses**

Offline, we used Spike 2 version 5 for Windows to analyze the force data. For the calibration, we determined the mean output per weight, and we measured the standard deviation of the force during baseline with and without scanning. Furthermore, we determined the maximal voluntary contraction force of nine subjects. The highest peak of the three contractions was considered the MVC force. Also, the dampening effects of the hammer stroke were analyzed, both the time until the oscillation amplitude was below 5% of the maximal amplitude and the oscillation frequency.

We used SPM2 (http://www.fil.ion.ucl.ac.uk/spm, Wellcome Department of Imaging Neuroscience; Friston et al. 1995) to evaluate the effect of the force transducer on the EPI images. Image series with and without the presence of the force transducer were realigned to the first image of all series to reduce motion artifacts. Thereafter, we calculated the difference between the EPI slices with and without the presence of the force transducer equipment. Furthermore, we applied a regression technique to the realigned data using the movement parameters as model vectors to remove residual motion effects. The pooled standard deviations (i.e., the averaged standard deviation of the voxel time series) were calculated on the residual data. The pooled standard deviations were also determined for the phantom measurements.

Brain activation during the maximal voluntary contractions was not analyzed (for functional data, see chapter 6, 7, and 8).

As mentioned above, the measurements to detect spurious frequencies were repeated twice (‘baseline’ and ‘force transducer’). A mean noise spectrum (in time) was determined for each frequency in the range of $127.78 \pm 0.44$ MHz. The noise spectrum of the ‘force transducer’ was divided by the noise spectrum of the baseline. When the MR does not detect spurious frequencies in the frequency range of interest, the quotient should be around 1.
Results

Force

Figure 4 shows the calibration of the force transducer. Linear regression analysis showed a good fit between the used weights and the voltage changes of the transducer ($R^2=0.9994$), and the intercept of the fitted line was close to zero (0.017 N).

At rest, both during scanning and non-scanning the standard deviations of the force baselines were small (0.065 and 0.062 N, respectively; $F_{(1,44)}=-0.920$, n.s.). This demonstrates that the changing magnetic fields inside the scanner room do not influence the force recording. The repeatability of the force transducer was estimated by measuring two weights (4.9N and 27.2N) ten times. All ten measurements varied less than 5% from the average value (range: -3.5 to 4.1%).

![Figure 4. Calibration of the force transducer, the actual recordings were made inside the scanner-room during scanning. The input weights (range: 1 - 60 N) versus the output (Volts).](image)

Figure 5 shows an example of a maximal voluntary contraction of the index finger in abduction direction. The highest peak was used to determine the MVC. The mean maximal voluntary contraction force (the mean of the peak values) of all subjects was 25.8 ± 8.6 N.

![Figure 5. An example of an MVC of the right index finger abductor recorded during MR-scanning.](image)

Figure 6 shows the result of hitting the force transducer with a hammer. The amplitude of the signal was 3.68 V, which equals 37.38 N (this falls within the range of the MVCs of the nine pilot subjects). The frequency response of the system was approximately 250 Hz, which is high compared to the relatively slow force signals. During the hammer
stroke the maximal amplitude was reached within 5 ms and within 15 ms the evoked oscillations remained below 5% of the maximal amplitude, which shows that the transducer can measure fast force changes accurately and that the signal quickly returns to baseline.

**Figure 6.** Response as a result of a hammer stroke on the force transducer. The horizontal line at 0 V represents the baseline. The horizontal line at 3.68V represents the maximal amplitude of the hammer stroke. The lines at 0.18 and -0.18 represent +5% and -5% of maximal amplitude, respectively. The vertical lines represent the start of the response (at 3.570s) and the end at which the oscillations pass the 5% of maximal amplitude for the last time (3.579s).

**Brain activity**

Figure 7 shows the mean of the time series of EPI slice 22 of one subject, obtained without (left), with (middle), and without (right) the force transducer system. The data were realigned to the very first scanned volume to remove movement artifacts. The signal intensity of the slices was scaled with their mean signal, so that they were comparable. The lower grey panel below shows the difference between the mean EPIs with and without the presence of the force transducer (left and middle), and also with and with the force transducer (right). If the transducer had had an effect somewhere on the EPIs, it would have been visible in this image. Only small differences are visible at
the borders of the brain; however, these differences are more likely to be movement artifacts, even though the images were realigned. The pooled standard deviations of the subject with and without the presence of the force transducer were not significantly different (13.15 and 11.27, respectively; ANOVA: F(1,5)=3.32, p=0.13). This was also the case for the pooled standard deviations of the phantom (6.68 ± 0.56 and 6.35 ± 0.35, respectively; ANOVA: F(1,4)=0.474, p=0.53). These results indicate that the force measuring equipment did not affect the EPIs.

Figure 7. The upper panel shows the mean time series of EPI slice 22, in the absence of the force transducer (1), in the presence of the force transducer (2), and in the absence of the transducer again (3); the lower panel shows the difference in the mean time series of EPI slice 22 between 1 and 2 (absence and presence of transducer), between 2 and 3 (presence and absence of transducer), and between 1 and 3 (both absence of transducer). The difference between both conditions in absence of the transducer was the largest, suggesting that the differences are more affected by changes and movements over time than by the presence of the transducers. The signal intensity of the slices was scaled to the mean signal intensity to allow a better comparison of the slices.

MR compatibility
The force measuring equipment did not affect the EPIs, as was confirmed by the test for spurious frequencies. Figure 8 shows the results of this test. The test revealed no differences between the empty scanner bore and the bore containing the force transducer at the usual hand location in the frequency range of 127.78 ± 0.44 MHz, as dividing ‘force transducer’ by ‘baseline’ resulted in values around 1.
Accurate measurements of force and force fluctuations are extremely important for understanding the relationship between muscle output and fMRI data. Especially, studies concerning the relation between the timing of muscle activation and the activity of various brain areas are of large interest both from a basic neurophysiological and pathological viewpoint.

We have demonstrated that it is possible to use a force transducer with strain gauges in an MR scanner environment. Despite the magnetic field (both static and changing), the transducer was able to measure a large range of forces accurately. In other words, the functioning of the transducer was not affected by MR imaging. Moreover, the transducer did not affect the MR images.

The voluntary contractions of the index finger abduction resulted in similar data as described earlier (Enoka et al., 1989; Fuglevand et al., 1993, 1995; Rutherford and Jones, 1988; Zijdewind et al., 1998; for brain activation data see chapter 6, 7, and 8). Generally, one of the advantages of strain gauges is that they can measure fast force fluctuations. Even with the MR adjustments, this advantage still holds, since the transducer was able to measure fast force fluctuations (caused by a hammer stroke) accurately. This suggests that the functional properties of our force transducer were compatible to the properties of other non MR compatible transducers.

The anatomical image of a single subject demonstrated no artifacts that could be caused by the force transducer. Furthermore, the EPI slices were not affected by the force transducer, although some differences were visible along the borders of the brain.

Figure 8. Noise spectrum of the fMRI scanner with the force transducer on the scanner bed divided by the noise spectrum of an empty scanner bore. Note the lack of significant peaks at all frequencies.
in the lower panel of Fig. 7. These were probably due to residual movement artifacts. Both the human and phantom measurements showed no effects of the force transducer on the pooled standard deviations of the EPI time series. Furthermore, no spurious frequencies were detected in the RF frequency range of 127.78 ± 0.44 MHz.

In summary, our data showed the possibility of measuring a broad range of index finger forces and fast force changes during brain imaging, which will be helpful in studies on human motor performance.
Appendix I. The detailed schematic of the transmitter (including the Wheatstone strain gauge bridge)

The power of the transmitter is supplied by a lead acid dry battery (J1). The signal derived from the Wheatstone strain gauge bridge (J5, J6) is amplified (U1; AD8230, Analog Devices, USA), and converted to a frequency-modulated optical signal (U2, AD654, Analog Devices, USA). This optical signal is transmitted (U6, HFBR1522, Agilent Technologies, USA) via an optical cable to the MR operator room. All resistors are made of SMD thick film 0805, except R6 and R7 which are made of 250mW TH metal film. All capacitors are made of SMD ceramic multilayer 0805, except C5 and C6 which are made of tantalum, and C1 which is made of metallized polyester TH.
Appendix II. The detailed schematic of the receiver

The optical signal enters the receiver via the optical signal detector (U1; HFBR2522, Agilent Technologies, USA); the signal is then re-converted to voltage signal, linearly corresponding to the applied force (U2; LM331N, National Semiconductor, USA), filtered (U3; OP07, Texas Instruments, USA), and amplified (U4; OP07, Texas Instruments, USA). The analogue signal is logged to a PC via Spike 2 (CED, Cambridge, UK).

All resistors are made of 250mW TH metal film. The capacitors C1, C3, C9, C10, and C11 are made of ceramic multilayer TH; C6, C7, and C8 are made of tantalum TH; C2, C4, and C5 are made of metallized polyester TH.