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Convertible pneumatic actuator for magnetic resonance elastography of the brain

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Abstract

Here we present a novel pneumatic actuator design for brain magnetic resonance elastography (MRE). Magnetic resonance elastography is a phase contrast technique capable of tracing strain wave propagation and utilizing this information for the calculation of mechanical properties of materials and living tissues. In MRE experiments, the acoustic waves are generated in a synchronized way with respect to image acquisition, using various types of mechanical actuators. The unique feature of the design is its simplicity and flexibility, which allows reconfiguration of the actuator for different applications ranging from in vivo brain MRE to experiments with phantoms. Phantom and in vivo data are presented to demonstrate actuator performance.

Keywords: MR elastography; Brain; Pneumatic actuator

1. Introduction

Magnetic resonance elastography (MRE) is a relatively new technique which allows the measurement of mechanical properties such as shear moduli of biological tissues and materials [1], [2] and [3]. Most importantly, MRE can be used as a diagnostic tool, based on the contrast between mechanical properties of healthy and pathological tissue. MRE can be used to examine organs such as liver, breast, prostate, kidney, and new applications are still emerging [4]. The unique feature of MRE is that it can be used even for regions like the brain, where traditional manual palpation is not applicable [5] and [6].

In dynamic MRE, the estimation of elastic properties is based on the measurement of the propagation of mechanical waves in materials and tissues. The information about wavelength and thus velocity of mechanical waves is captured by the phase-sensitive

MRI method. The mechanical waves are generated synchronously with chronological image recording. MRE data can further be used for calculations of elastic material properties. The mechanical waves do not need to be generated continuously. Usually they are transmitted in a burst of certain duration prior to each MR signal excitation. However, timing of the MR sequence has to be setup in such a way that the waves are able to cover the whole region of interest. The actuator vibration frequency depends on the applications but is usually kept within the low acoustic range of 40–150 Hz [7].

Three common types of motion sources used in the actuator design are electromagnetic, piezoelectric and pneumatic. The most frequently used is the electromagnetic principle as it generally allows a simple and relatively low-cost design. However, various problems with electromagnetic interferences can occur during MR signal acquisition. The piezoelectric actuators provide high force and an excellent control of motion frequency and magnitude. However, the drawback of this design is its higher complexity, price as well as problems with distortions such as susceptibility and eddy current artifacts which have to be addressed as well [8]. The pneumatic-type actuator virtually eliminates these previously mentioned interference problems; however, the operating frequency range is usually limited to lower frequencies [8].

In the case of brain elastography, direct contact of the actuator with the examined tissue is not possible. Therefore, vibration of the whole head is typically used to induce shear waves into the brain and overcome the mechanical shielding of the surrounding skull [9]. Electromagnetically driven bite bars or cradle-type actuators have been used in recent brain elastography studies for this purpose [10] and [11]. Potential problems with MR compatibility can be avoided using an external vibration source, located away from the magnet and the RF coil. An example of such an MRE setup has been used in Ref. [12], where the subject's head is placed on a head cradle coupled to an external loudspeaker, acting as a source of vibrations, with a flexible carbon fiber rod of variable length. Alternatively, a pneumatic concept where the subject's head is resting on two flexible drum membranes which are positioned in a “V” shape underneath the head and are periodically acoustically excited with a phase difference of 180° has been reported in Ref. [13].

In this article, a simple actuator for brain elastography is introduced. The primary focal points of the design were subject comfort, simple usage and convenience for clinical use. However, as MRE typically involves extensive experimenting with various phantoms, the option to use the driver for such purposes as well was kept in mind. The novel and unique feature of this three-in-one design is its versatility, which allows the use of the same actuator system for human studies as well as for different phantom experiments. The simple actuator design allows easy integration in a wide range of standard RF head coils. The different configurations of the proposed convertible pneumatic actuator are presented followed by experimental results obtained on phantoms and on a healthy human volunteer.

2. Materials and methods

The concept of a pneumatic actuator design has been chosen because of its simplicity, good MR compatibility and maximization of subject safety. An important restriction that has to be considered prior to usage of such an actuator type is its limitation to low-frequency excitation [8]. Recently published brain elastography studies exploited frequencies in the range of 25–100 Hz [10], [12] and [14]. This is very similar to the frequency range achieved with a pneumatic approach, where a loud speaker is connected to a flexible hose itself connected to an actuator placed near the organ of interest [15], [16] and [17].

In order to increase the actuator performance we choose a dual-driver approach, which helps to compensate for shear wave attenuation and therefore provides a better coverage over the whole field of interest [18]. A pair of standard active subwoofers has been modified with airtight acrylic lids mounted over the speaker membrane and used as the source of air pressure. At the center of the lids a hose connector has been mounted to attach flexible tubing for the delivery of acoustic waves into the passive drivers (see Fig. 1A). One of the subwoofers is equipped with an electrical switch, which could be used to reverse the membrane motion and thus induce a mutual phase shift of 180° into the air pressure waves generated from both speakers.

Fig. 1. (A) Pair of modified active subwoofers used to deliver the air pressure waves into the passive drivers, the programmable waveform generator and oscilloscope are visible on the top shelf of the cart. Three different actuator configurations are shown: (B) head in vivo, (C) probe and (D) bed-type configuration.

The actuator concept is designed to operate with a standard transmit/receive head coil and for usage with three different configurations: head, probe and bed type. This design allowed using the actuator for both in vivo human brain MRE studies and experimenting with various phantoms.

2.1. Head actuator configuration

The driver unit, instead of a solid drum covered with flexible membrane [18] and [19], consists of standard plastic bottles commonly used as containers for various food and cosmetics products. We found that polyethylene bottles from liquid honey (no name brand) are well suited for this purpose. The containers have a flat rectangular shape with flexible walls and can be placed in between the patient's head and the bottom of the coil. A homemade fitting is used to connect the container to the hard-walled flexible tubes. In order to induce side-to-side, cradle-like head motion the drivers are inflated with air pressure waveforms with a 180° mutual phase difference. The gap, of approximately 60 mm, between the actuator bottles, is filled with semisoft foam, providing additional support for the head. The drivers are attached with Velcro fasteners to the plastic tray inserted into the bottom of the head coil. The tray is made from high-density polyethylene and was thermally molded with a hot gun to conform to the bottom of the coil. Fig. 1B shows the actuator setup inserted into the head coil. A small towel can be placed over the drivers for better subject comfort and for hygienic purposes and can be changed between exams.

2.2. Probe-type configuration

For the phantom experiments where a small contact area with the source of oscillations is desirable, the configuration can be changed to a so-called probe-type actuator. The passive drivers are placed in homemade brackets, which are attached to the coil instead of to the head restrainers. A T-bar with an actuator contact plate is positioned between the passive drivers and is secured with Velcro strips glued to the surfaces of both ends of the shaft and drivers. The height of the contact plate is adjustable using a screw, according to the size of the phantom. The contact plates are replaceable according to the desired waveform shape. Fig. 1C shows a typical example of an experimental setup with the probe-type actuator.

2.3. Bed-type configuration

For phantom studies where a large contact area with the source of oscillations is desirable, the configuration can be changed to a so-called bed-type actuator [20]. The passive drivers are placed in the brackets replacing the head restrainers in the same way as for the probe-type configuration. A table holder, equipped with four rollers, is positioned into the bottom part of the head coil. The rollers have glass ball bearings and their axes are positioned parallel to the main magnetic field, allowing motion in the direction perpendicular to the B_0 field. The small cart positioned between the drivers

(see Fig. 1D) serves as the sample holder. On the bottom of the cart, two notches were machined to keep it stable on the rollers during vibrations.

2.4. Experimental setup

All experiments have been conducted on a 3-T scanner (TimTrio; Siemens Medical Solutions, Erlangen, Germany) using a standard transmit/receive head coil. In order to eliminate any RF interference the hoses from the remotely placed speakers enter the magnet room through waveguide filters installed on the shielding cage. An arbitrary function generator (Tektronix AFG3022B, Tektronix, Inc., USA), triggered from the MRE pulse sequence, is used to generate acoustic oscillations. Homemade software written in LabVIEW (National Instruments Corp., Austin, TX, USA) has been developed to setup waveform parameters. Synchronization and timing of the MRE experiments, i.e., trigger pulses, magnetic field gradients and the acoustic waveform, are monitored with a four-channel oscilloscope (BitScope BS-442N, BitScope Designs, St. Leonards, NSW, Australia). Both instruments are controlled through a local area network from a laptop computer placed beside the spectrometer console.

Two types of modified pulse sequences are used for data acquisitions: a gradient-echo (GE) and a single-shot spin-echo (SE) echo planar imaging (EPI) sequence, and both are extended with trigger timing and motion-encoding gradient (MEG) modules. Reconstruction of the wave images and calculations of elastograms are accomplished with MRE/Wave, a software package obtained from the Mayo Clinic in Rochester, MN, USA. This software is based on the local frequency estimation (LFE) algorithm [21].

3. Results

The frequency characteristics of all three actuator configurations have been examined in a bench test using a vibration meter (PCE-VT 2700, PCE Group, Meschede, Germany). The accelerometer probe was attached to the actuator in the presumed location of contact with the sample. The sinusoidal signal from the waveform generator was set for 0.3 V amplitude and a frequency range of 10 to 200 Hz, stepped with 10-Hz increments. The displacement plots for all three configurations are shown in Fig. 2. The plot reveals that the actuator is operating most efficiently for frequencies up to 80–90 Hz, where the motion amplitudes are above $\sim 100\text{ }\mu\text{m}$; however, experiments with phantoms demonstrate that a higher frequency range can be exploited as well.

Fig. 2. Actuator displacement as a function of frequency for different actuator configurations. The data were measured on the bench for unloaded actuators and reveal that for frequencies up to 80 Hz the displacement magnitude is larger than $\sim 100 \mu\text{m}$. The inset shows a magnified part of the displacement for the 80- to 150-Hz frequency range.

3.1. Phantom experiments

The performance of the probe-type configuration is demonstrated using a cylindrical gel phantom of $\sim 100 \text{ mm}$ diameter and $\sim 90 \text{ mm}$ length, which is made from $\sim 0.5\%$ agarose (weight %). A spherical inclusion of $\sim 30 \text{ mm}$ diameter made from 0.75% agarose is enclosed in the center.

The measurement was conducted for mechanical frequencies of 50, 100 and 150 Hz. Prior to each excitation of the MR signal, the mechanical vibration was applied for a time interval of 200 ms. The GE images were acquired with the following parameters: TR/TE=350/26, 15, 12 ms; flip angle 45° ; field of view (FOV) $150 \times 150 \text{ mm}$; resolution of 128×128 pixels; and slice thickness of 5 mm. The isochromats displacements due to wave propagation were encoded with one sinusoidal-shaped gradient cycle of 1, 3 and 18 mT/m magnitude in eight equally spaced time increments within one wave cycle. The increasing gradient magnitudes were used to compensate smaller displacements at higher mechanical frequencies.

Fig. 3 shows the experimental results obtained for sagittal slice orientations. The comparison of calculated displacement maps (Fig. 3H–J) reveals that the wave amplitude reduces with increasing acoustic frequency, i.e., while the maximum displacement at 50 Hz is more than $1000 \mu\text{m}$, it drops down to $100 \mu\text{m}$ at 150 Hz. However, even for 150 Hz, the wave amplitude in the whole slice is larger than the theoretically predicted minimum, estimated from the acquisition parameters and signal-to-noise ratio of the magnitude images (~ 90) which is about $0.88 \mu\text{m}$ (see Ref. [2]).

Fig. 3. MRE data obtained with the probe-type actuator and a cylindrical phantom made from 0.5% and 0.75% agarose for matrix and inclusion, respectively. (A) Density image; (B–D) shear wave images for 50, 100 and 150 Hz; (E–G) calculated shear stiffness; (H–J) displacement maps. Image (A) shows the location of actuator plate, which moves in perpendicular direction to the image plane.

The application of the bed actuator setup is demonstrated with the gel phantom made from 0.5% agarose poured into a semi-rigid plastic container. A small cylinder-shaped inclusion of ~ 10 mm diameter made from 0.75% agarose was located at the center of the phantom. The container with the phantom was positioned on the actuator tray as shown in Fig. 1D and attached to the tray with double-sided adhesive tape to prevent sliding during vibrations. The experimental parameters were identical as in the previous experiment except for a smaller FOV of 100×100 mm and a different MEG amplitude of 1, 7 and 20 mT/m. The results from these measurements are shown in Fig. 4.

Fig. 4. MRE data obtained with the bed-type actuator and agarose phantom poured into a plastic container. The concentrations of agarose are 0.5% and 0.75% for matrix and inclusion, respectively. (A) Density image; (B–D) shear wave images at 50, 100 and 150 Hz; (E–G) calculated shear stiffness; and (H–J) displacement amplitude maps. Image (A) indicates the location and motion direction of the actuator.

It should be pointed out that, despite the fact that all phantom measurements were able to clearly depict the presence of the embedded inclusion, the actual size and shear stiffness value exhibit frequency-dependent variation. This can be explained by the properties of the LFE algorithm such as limited resolution and the fact that the correct estimate is reached only half a wavelength into a given region [21] and [22].

3.2. Brain elastography experiments

The head actuator setup was used to collect in vivo data from healthy volunteers. MRE images of three axial slices with the center slice going through the genu and splenium of the corpus callosum were acquired with SE EPI, using the following parameters: TR=6.0 s, TE=102 ms, FOV=220 mm, resolution of 128×128 pixels, slice thickness of 5 mm, and a fat saturation pulse was applied to suppress ghost artifacts from lipids. Four data sets using 50, 65, 80 and 90 Hz for mechanical frequency excitation were acquired,

while the test subject was in the supine position. The one (50, 65 Hz) or three (80, 90 Hz) MEG cycles of 32 mT/m amplitude in the phase-encode direction (parallel with anterior–posterior direction) were used to encode wave propagation at 20 different phase offsets within one wave cycle. The data obtained with the lower mechanical frequencies exhibits a substantially higher displacement and a better penetration into the deep brain areas when compared to the higher frequency data sets (see Fig. 5C, F, J and M). However, the data sets acquired with higher mechanical excitation could provide better spatial resolution, especially when a wavelength-sensitive reconstruction approach such as, e.g., LFE is exploited. This is well demonstrated in Fig. 5D, G, K and N.

Fig. 5. Results obtained from in vivo human brain experiments using 50, 65, 80 and 90 Hz excitation. (A) Fast spin-echo image showing the anatomy of the examined slice; (H) EPI image indicates the location of the actuators; (B, E, I, L) images of the shear wave propagating in the brain; (C, F, J, M) displacement maps; and (D, G, K, N) the shear stiffness estimates.

4. Conclusion

The new convertible actuator design allows for an easy and convenient experimental setup of the various MRE measurements. Experiments with phantoms and volunteers (over 10 healthy volunteers participated in the setup testing so far) showed that the actuator produces suitable shear waves, which can be used for the calculation of the elastic properties of tissues and materials. We demonstrated that the usable frequency range for the experiments with phantoms is up to 150 Hz, and for in vivo human brain measurements, vibrations up to 80 Hz have been exploited. The actuator design avoids the use of any materials that could disturb the homogeneous field of the MR scanner. MRE examinations with volunteers showed that the actuator is comfortable and does not produce any additional inconvenience for the subject. The subject's head is resting on the actuator naturally and there is no necessity for any specific mechanical setup, which could possibly affect the acquired data. The actuator concept is easy to integrate with most commercially available RF coils, which are open on both sides. These features make the actuator design especially suitable for research and clinical studies where high subject throughput, consistency and patient safety are required.

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References

1. R. Muthupillai, D.J. Lomas, P.J. Rossman, J.F. Greenleaf, A. Manduca, R.L. Ehman Magnetic resonance elastography by direct visualization of propagating acoustic strain waves *Science*, 269 (1995), pp. 1854–1857
2. R. Muthupillai, P.J. Rossman, D.J. Lomas, J.F. Greenleaf, S.J. Riederer, R.L. Ehman Magnetic resonance imaging of transverse acoustic strain waves *Magn Reson Med*, 36 (1996), pp. 266–274
3. C.J. Lewa Magnetic resonance imaging in the presence of mechanical waves *Spectroscopy Letters*, 24 (1991), pp. 55–67
4. Y.K. Mariappan, K.J. Glaser, R.L. Ehman Magnetic resonance elastography: a review *Clin Anat*, 23 (2010), pp. 497–511
5. L. Xu, Y. Lin, Z.N. Xi, H. Shen, P.Y. Gao Magnetic resonance elastography of the human brain: a preliminary study *Acta Radiol*, 48 (2007), pp. 112–115
6. G. Liu, P. Gao, Y. Lin, J. Xue, X. Wang, B. Sui *et al.* Brain magnetic resonance elastography on healthy volunteers: a safety study *Acta Radiol*, 50 (2009), pp. 423–429
7. E.C. Ehman, P.J. Rossman, S.A. Kruse, A.V. Sahakian, K.J. Glaser Vibration safety limits for magnetic resonance elastography *Phys Med Biol*, 53 (2008), pp. 925–935
8. K. Uffmann, M.E. Ladd Actuation systems for mr elastography: design and applications *IEEE Eng Med Biol Mag*, 27 (2008), pp. 28–34
9. A. Di Ieva, F. Grizzi, E. Rognone, Z.T.H. Tse, T. Parittotokkaporn, F. Rodriguez Y Baena *et al.* Magnetic resonance elastography: a general overview of its current and future applications in brain imaging *Neurosurg Rev*, 33 (2010), pp. 137–145 discussion 145
10. M.A. Green, L.E. Bilston, R. Sinkus In vivo brain viscoelastic properties measured by magnetic resonance elastography *NMR Biomed*, 21 (2008), pp. 755–764
11. S.A. Kruse, G.H. Rose, K.J. Glaser, A. Manduca, J.P. Felmlee, C.R.J. Jack *et al.* Magnetic resonance elastography of the brain *NeuroImage*, 39 (2008), pp. 231–237
12. I. Sack, B. Beierbach, U. Hamhaber, D. Klatt, J. Braun Non-invasive measurement of brain viscoelasticity using magnetic resonance elastography *NMR Biomed*, 21 (2008), pp. 265–271
13. P. Rossman Pneumatically actuated driver for use in mre of the brain, ISMRM-Flow and Motion Study Group Workshop (2006) New York
14. I. Sack, B. Beierbach, J. Wuerfel, D. Klatt, U. Hamhaber, S. Papazoglou *et al.* The impact of aging and gender on brain viscoelasticity *NeuroImage*, 46 (2009), pp. 652–657
15. O. Rouvière, M. Yin, M.A. Dresner, P.J. Rossman, L.J. Burgart, J.L. Fidler *et al.* Mr elastography of the liver: preliminary results *Radiology*, 240 (2006), pp. 440–448
16. Y.K. Mariappan, K.J. Glaser, A. Manduca, A.J. Romano, S.K. Venkatesh, M. Yin *et al.* High-frequency mode conversion technique for stiff lesion detection with magnetic resonance elastography (mre) *Magn Reson Med*, 62 (2009), pp. 1457–1465

17. D. Grenier, L. Milot, X. Peng, F. Pilleul, O. Beuf A magnetic resonance elastography (mre) approach for liver investigation Conf Proc IEEE Eng Med Biol Soc, 2007 (2007), pp. 2607–2610
18. Y. Zheng, G. Li, M. Chen, Q.C.C. Chan, S.G. Hu, X.N. Zhao *et al.* Magnetic resonance elastography with twin pneumatic drivers for wave compensation Conf Proc IEEE Eng Med Biol Soc, 2007 (2007), pp. 2611–2613
19. B.C. Goss, K.P. McGee, E.C. Ehman, A. Manduca, R.L. Ehman Magnetic resonance elastography of the lung: technical feasibility Magn Reson Med, 56 (2006), pp. 1060–1066
20. T. Oida, Y. Kang, T. Matsuda, J. Okamoto, T. Azuma, O. Takizawa *et al.* Bed-type oscillator for mr elastography ISMRM 12th Scientific Meeting, Kyoto (2004), p. 1773
21. H. Knutsson, C. Westin, G. Granlund Local multiscale frequency and bandwidth estimation IEEE International Conference on Image Processing (1994)
22. A. Manduca, T.E. Oliphant, M.A. Dresner, J.L. Mahowald, S.A. Kruse, E. Amromin *et al.* Magnetic resonance elastography: non-invasive mapping of tissue elasticity Med Image Anal, 5 (2001), pp. 237–254