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A novel magnetic resonance elastography transducer concept based on a rotational eccentric mass: preliminary experiences with the gravitational transducer

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Abstract

Background. Magnetic resonance elastography (MRE) is used to non-invasively estimate biomechanical tissue properties via the imaging of propagating mechanical shear waves. Several factors including mechanical transducer design, MRI sequence design and viscoelastic reconstruction influence data quality and hence the reliability of the derived biomechanical properties.

Purpose. To design and characterize a novel mechanical MRE transducer concept based on a rotational eccentric mass, coined the gravitational transducer.

Materials and methods. Table top measurements were performed using accelerometers to characterize the frequency response of the new transducer concept at different driving frequencies (f_{VIB}) and different rotating masses. These were compared to a commercially available pneumatically driven MRE transducer. MR data were acquired on a 3T scanner using a fractionally encoded gradient echo MRE sequence in three healthy volunteers. Acceleration and displacement spectra were plotted in units of g and mm, respectively, and visually compared, emphasizing the ratio between the peaks at f_{VIB} and its 2nd harmonic, a known cause of error in the reconstruction of biomechanical properties as is explored in more detail in numerical simulations here. No formal statistical testing was performed in this proof-of-principle paper.

Results. The new transducer concept shows—as expected from theory—a quadratic or linear increase of acceleration amplitude with increase in f_{VIB} or mass, respectively. Furthermore, different versions of the transducer show markedly lower 2nd harmonic-to- f_{VIB} ratios compared to the commercially available pneumatically driven transducer. Displacement was constant over a range of f_{VIB} , in accordance with theory. Phantom and *in vivo* data show low nonlinearity and excellent data quality.

Conclusion. The table top measurements are in concordance with the theory behind a transducer based on a rotational eccentric mass. The resulting constant displacement amplitude irrespective of f_{VIB} and low 2nd harmonic-to- f_{VIB} ratio result in low nonlinearity and high data fidelity in both phantom and *in vivo* examples.

Abbreviations

EM	Electro-magnetic
$f_{\rm VIB}$	Vibrational frequency
$f_{\rm MEG}$	Motion-encoding gradient frequency
MEG	Motion-encoding gradient
MRE	Magnetic resonance elastography
MS	Multislice
PEEK	Polyether ether ketone
PTFE	Polytetrafluorethylene
TTL	Transistor-transistor logic

Introduction

Magnetic resonance elastography (MRE) is a non-invasive MR imaging (MRI) technique, capable of quantifying soft tissue biomechanical properties by applying vibrations to the tissue of interest using an MRE transducer (Muthupillai et al 1995, Glaser et al 2012). For clinical applications, low-frequency vibrations (e.g. 10–100 Hz) are employed. The longitudinal transducer vibrations are converted into shear waves at the borders of the transducer piston as well as on interfaces within the body. Subsequently, the total displacement—consisting of compressional and shear waves—is encoded in the phase of the MR signal using a synchronized motionencoding gradient (MEG). The biomechanical properties of soft tissue are then calculated from the MR phase data using a mathematical inversion based on the principles of wave propagation in a viscoelastic medium (Sinkus et al 2005, Hirsch et al 2016b, Fovargue et al 2018). MRE is FDA-approved (FDA) for liver fibrosis grading since 2009 (Loomba et al 2016) and has the potential to make a significant impact in other diagnostic areas, e.g. cancer staging and therapy response monitoring (Pepin et al 2015). To further facilitate this development, more reliable estimates of soft tissue biomechanical properties are desirable. This can be achieved (i) by mechanical transducers of higher fidelity to improve the quality of the source, (ii) by improved data acquisition methods providing more SNR and less motion artefacts from residual respiration for instance (Guenthner et al 2018), and (iii) by improved reconstruction algorithms for obtaining the complex shear modulus (Fovargue *et al* 2018). Improved acquisition schemes and reconstruction algorithms can do only so much to overcome the deleterious effects of suboptimal vibrational wave generation, hence improving the wave generation has priority.

To this end, several transducer setups have been proposed (Uffmann and Ladd 2008, Hirsch *et al* 2016a), generally speaking based on either pneumatic (Yin *et al* 2007, Dittmann *et al* 2017) air or electro-magnetic (EM) engine-coils (Sahebjavaher *et al* 2014, Forsgren *et al* 2015, Runge *et al* 2016). Other approaches are based on piezoelectric elements. These setups can have either rigid (rods) or more flexible (air or water) transmission of the shear waves into the object of interest (Hirsch *et al* 2016a). The only available commercial transducer setup is based on pneumatic air and uses a passive acoustic driver (Yin *et al* 2007). Its major advantage is its elegant handling and positioning due to its flexible design, while its drawbacks include the possibility of silencing the passive driver when too much pre-load is applied (i.e. when strapped too tight or when the patient lies on top) and the inaccuracies of the generated vibrational waves related to the inherent nonlinear properties of a pneumatically driven system, generating upper harmonics. Both issues deteriorate wave quality and seriously hamper proper reconstruction of the final viscoelastic maps. The main drawback of an EM engine-coil set-up is that it suffers from image distortions in the field-of-view (FOV) near the engine-coil.

In general, tissue absorbs higher frequencies more efficiently. As most transducer concepts yield reduced amplitude with increasing frequency, generally less efficient wave penetration is observed when operating at higher frequencies. In addition, most transducers are designed for use with a particular body part of interest and are not necessarily easy to adapt for other applications. In all, there are several transducer-related issues that should be addressed or improved upon in order to provide high-fidelity waves: (i) vibrational frequency (f_{VIB}) accuracy, (ii) induced image artefacts, (iii) preserved transducer amplitude at higher f_{VIB} , (iv) modularity of the setup and (v) intrinsic capability of generating several frequencies simultaneously, i.e. multi-frequency MRE.

The purpose of this study is to develop and demonstrate the principle of a new MRE transducer based on the concept of a rotational eccentric mass that addresses the aforementioned points.

Methods

Theoretical basis of the gravitational transducer

This concept is coined a 'gravitational transducer' as it directly utilises the generic equivalence of acceleration to force. The gravitational transducer is based on the concept of a rotational eccentric mass, similar to those found in mobile phones' vibrating alert motors or compaction plate motors (figure 1).



Figure 1. Conceptual drawing of the gravitational transducer. The outer transducer casing (shown here as the light grey square) is connected by bearings and an axis (shown as circle in slightly darker grey) to an eccentric mass (shown as dark grey triangle). As the axis rotates at a specific driving frequency ($f_{\rm VIB}$, shown by the curved, solid arrow), a centripetal force keeps the eccentric mass on its plane of rotation. The results is a time-varying force in all directions (shown by the numbered dotted arrows), transmitted via the axis and bearings to the transducer casing, which will vibrate at the desired frequency $f_{\rm VIB}$. Note that the arrows point outwards to highlight the force generated by the rotating mass, not by the rotating axis.

The basic premise is that a small eccentric mass rotates at a given angular frequency (f_{VIB}); the centripetal force keeping it on its circular path in the plane orthogonal to the axis of rotation is following

$$F(t) = -m \cdot r \cdot \omega^2 e^{i(\phi + t\omega)},\tag{1}$$

where *m* is the mass, *r* is the distance from the centre of gravity of the mass to the axis of rotation, ω is the angular frequency (in our setup this is $2\pi \cdot f_{\text{VIB}}$) and ϕ is an arbitrary phase. According to Newton's third law, an equivalent opposing force will act on the whole transducer system in the same two-dimensional plane orthogonal to the axis of rotation. Neglecting the coupling to the body, the system of total mass M + m is accelerated according to

$$\frac{d^2x(t)}{dt^2} = -\frac{F(t)}{M+m},\tag{2}$$

where M is the mass of the object to be shaken. The displacement x(t) of the transducer system is then given by

$$x(t) = -\frac{m}{M+m} \cdot r \cdot e^{i(\phi+t\omega)}.$$
(3)

Hence, the displacement amplitude is given by

$$|x(t)| = \frac{m}{M+m} \cdot r, \tag{4}$$

which approximates to

$$|\mathbf{x}(t)| = \frac{m}{M} \cdot \mathbf{r} \tag{5}$$

in first-order Taylor series expansion in m/M, which is accurate to 1%, if the mass ratio m/M is less than 10%. Consequently, the amplitude of the transducer is independent of f_{VIB} . Assuming therefore that the eccentric mass is much smaller than the weight of the casing, the force exerted by the transducer scales linearly with mass and quadratically with the rotational frequency.

Design of the gravitational transducer

The gravitational transducer setup (figure 2) consists of (A) a 4 nm stepper motor (Nema23 1.8° Dual shaft, 60BYGH401-03, CNC4YOU Ltd, Milton Keynes, UK) and stepper controller unit (MCC-1-32-48 MINI, Phytron GmbH, Gröbenzell, Germany) and (B) a rotating shaft now made from fully flexible 3 m polyether ether ketone (PEEK) sections interconnected via locknut connections attached to (C) the gravitational transducer casing. The setup is modular in design and allows the transducer to be detached from the rotating shaft and interchanged with body part specific designs. The PEEK rotating axis allows operation at extreme bends, yielding extreme flexibility in terms of placement in relationship to the MRI scanner but is also highly robust and reliable in terms of absolute phase stability. Figure 2 shows designs for upper abdominal applications such as liver and spleen MRE. Other designs that have been tested include breast (Hoelzl *et al* 2017) and brain. The transducer casing (C) is made of CNC-lathed polyoxymethylene copolymer and consists of two parts that are connected with polyether ether ketone (PEEK) screws. The casing houses a (D) piece of CNC-lathed polytetrafluorethylene (PTFE) acting as eccentric mass, which is glued onto a PEEK rod which itself is glued to (E) PEEK bearings (BB-626-A500-70-GL, igus[®] GmbH, Köln, Germany) containing ceramic bearing balls. A second PEEK rod glued to PEEK bearings



Figure 2. Details of the gravitational transducer setup. The *upper left* panel shows (A) the gravitational transducer mainboard housing a 4 nm stepper motor (60BYGH401-03, CNC4YOU) with controller unit (MCC-1-32-48 MINI, Phytron). The *upper right* panel shows (C) the transducer fixed to (B) the flexible PEEK rotating axis. The *lower left* and *right* panels show that the transducer casing houses two PEEK rods, one of which is glued to (D) a PTFE eccentric mass and both to (E) two PEEK bearings. The two PEEK rods are connected via (F) two PEEK timing pulleys and a timing belt. The smaller rod is connected to the rotating axis via a locknut connection.

(BB-608-A500-70-GL) and connected to the flexible rotating axis via a locknut connection is interconnected with (D) via a second PEEK timing pulley and belt (F) with a 1:3 ratio: three rotations of the mass for each rotation of the rotating axis. Depending on the application and desired range of frequencies, the ratio can be chosen accordingly. Polypropylene shims placed directly next to the timing pulleys keep the (F) timing pulley belt locked on the pulleys, restricting movement in the *z* direction (along the axis of rotation). The timing pulley gear box results in a decreased load on the rotating axis and stepper motor. By adding additional PEEK rods and eccentric masses with differently geared timing pulleys or planetary gears, intrinsic multi-frequency MRE experiments are possible: carefully chosen mass and gear ratios even allow one to construct a wave pattern where the individual frequencies have the same amplitude, which is not possible using current transducer concepts. The multi-frequency approach is not tested in this study.

Controlling the gravitational transducer

The stepper motor described earlier is controlled by a dedicated stepper motor controller that is synchronized to transistor–transistor logic (TTL) signals sent by the MR system at the start of specific acquisition loops, such as the slice selection loop in case of a multislice (MS) acquisition. The stepper motor control functions are implemented in the Phytron MiniLog format (Pythron GmbH 2010). A stepper motor uses microsteps to perform a full turn of the motor. In case of the gravitational transducer, this is set to 800 microsteps (stepsPerRotation), resulting in a desired number of steps per second, which follows:

stepsPerSecond = stepsPerRotation
$$\cdot f_{\text{VIB}} \cdot \text{gearRatio.}$$
 (6)

For example, at $f_{\rm VIB}$ of 45 Hz and gear ratio of 1:3 this results in 12000 microsteps s⁻¹. MRE sequence modifications were made to allow the stepper motor to accelerate to this velocity before the start of data acquisition. During a user-defined number of dummy shots following the first TTL-signal, the motor reaches a constant rotational velocity. Subsequently, the angular position of the mass must be maintained in synchrony with the MRI acquisition protocol to enable precise quantification of the propagating waves, i.e. the acquisition and vibrational waves are phase-locked. The length of an integer number of image shots (excitation, encoding and readout) exactly matches the length of the vibrational wave, provided the stepper motor reaches a constant

rotational velocity. Hence, any angular misalignment of the mass throughout the MRE acquisition will lead to uncorrectable errors in the final displacement maps and subsequently to unpredictable biases for the reconstruction of the biomechanical properties.

Initial tests indicated that nominal motor settings resulted in a loss of correct angular position relative to the reference position. This is related to timing errors in the frequency converter and processor timing, which accumulate in the wave form generator during the experiment. Normally, these issues are of little practical consequence for industrial applications of stepper motors like CNC machines, which do not need to operate at such precision for such sustained periods of time. Our experiments however require far more stability over a prolonged period of time. We therefore further customised the stepper motor control functions to ensure the angular stability and synchrony between the angular position and the MRI acquisition protocol by verifying the stepper motor position relative to the reference position—stored at reception of the first TTL-signal after the dummy shot(s)—and de- or accelerating as required throughout the next run period between TTL-signals.

Characterisation using accelerometers

We tested the frequency response of two MRE transducers: the gravitational transducer in two variants (liver and brain) and an air-based commercially available and commonly used device (Resoundant, Inc., Rochester, MN, USA) by measuring the according vibrational spectra. For the gravitational transducer we used a liver transducer with a 1:3 gear ratio and a 38.24 g mass, and a brain transducer with 1:2 gear-ratio and 7.45 | 10.48 | 13.50 | 16.52 g masses. The brain transducer is smaller and thinner and fits easily inside a head coil for brain experiments. A commercially available ADXL345 accelerometer (Analog Devices Inc., Norwood, USA) controlled by a Raspberry Pi 3 Model B (Raspberry Pi Foundation, Cambridge, United Kingdom) was fixated to the transducer casing to measure its acceleration. In turn, this was fixated firmly to the quadriceps muscle of a volunteer using an elastic strap. Experiments were performed with driving frequencies $f_{\rm VIB}$ of 40, 50, 60 and 70 Hz, representing the likely spectrum of frequencies used for *in vivo* applications. Acceleration spectra were recorded for 10 s at 500 Hz sampling frequency, leading to a spectral resolution of 0.1 Hz after Fourier transform and a bandwidth of ± 250 Hz. Recorded data were transformed to acceleration presented in units of g (9.8 m s⁻²) and to displacement in units of mm.

Numerical simulations

Numerical wave simulations were performed to demonstrate possible detrimental effects to viscoelastic reconstruction from the presence of significant upper harmonics in the source. A finite element method was used within the CHeart software (Lee *et al* 2016) to model steady-state vibrational waves at 40 Hz in a 2D liver mesh using 2684 nodes and 5009 triangular elements. Linear viscoelasticity and isotropy were assumed and the storage and loss components of the shear modulus were set to 1.5 and 0.2 kPa, respectively, and assumed constant across the entire domain as well as over all frequencies (i.e. no power law assumption). The simulation solves a static time-harmonic linear isotropic viscoelastic problem with Dirichlet boundary conditions enforced on one section of the boundary to model the transducer. Simulations were also run with 80, 120 and 160 Hz waves to ascertain the behaviour of the 1st, 2nd and 3rd upper harmonics. The solutions (complex-valued steady-state displacement fields) can be transformed to real-valued time-dependent displacement fields and summed, mimicking what is measured by MR.

Phantom measurements

The gravitational transducer was tested on a dedicated Elastography phantom (Model 049, Computer Imaging Reference Systems Inc., Norfolk, VI, USA) using the brain transducer (1:2 gear ratio) with a 7.45 g mass. Data acquisition was performed using a clinical 3T MR Scanner (Achieva, Philips, The Netherlands) with motion encoding gradients (MEGs) in three orthogonal directions and a fourth reference scan without MEGs to compensate for motion encoding from the imaging gradients. A fractionally encoded gradient echo MRE acquisition was performed with $f_{\rm VIB} = 90$ Hz, $f_{\rm MEG} = 180$ Hz, TE = 6.9 ms (in-phase) and TR = $\left(N_{\rm SLICES} \cdot T_{\rm VIB} / N_{\rm SHOTS/VIB}\right) + \left(\frac{1}{N_{\rm WAVEIMAGES}} \cdot T_{\rm VIB}\right) = 90.3$ ms with 1 image shot per vibration period and 8 wave images. An image shot here refers to excitation, encoding and readout (Garteiser *et al* 2013). Reconstructed voxel size was $2.0 \times 2.0 \times 2.0$ mm³ with acquisition and reconstruction matrices of $96 \times 96 \times 8$. The readout bandwidth was 163 kHz, balancing SNR requirements while minimising image artefacts generated from a long readout. SENSE factor was set to 2. The acquisition time per encoding direction was 35 s.

Phase images were processed offline. The processing pipeline performs phase unwrapping, reference scan subtraction and phase offset corrections according to slice number originating from the fact that one mechanical oscillation period encompasses several slice acquisitions (Garteiser *et al* 2013). We only accepted voxels with magnitude >15% of the average magnitude of the area of interest and a total wave displacement amplitude >10 μ m. The mechanical properties were derived using a reconstruction with a localized divergence free finite element model (Hirsch *et al* 2016b, Fovargue *et al* 2018). The curl of the displacement fields is calculated by local least

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squares polynomial fitting and shown here for presentation purposes as this operator will remove the compressional wave component from the data. To assess data quality, we estimated the nonlinearity in the acquired data sets. Nonlinearity is defined as the ratio of the amplitude of the 2nd harmonic vibrational frequency to the base frequency (f_{VIB}). Hence, a perfect sinusoid without any noise will lead to a nonlinearity of 0. On the contrary, if the 2nd harmonic is as strong as the base frequency, the ratio will be 1, i.e. 100%. It therefore represents the percentage of deviation from the expected sinusoid and characterises thereby the quality of the vibration and the noise. In this paper, the elasticity values reported are the magnitude of the complex shear modulus (Gabs).

In vivo human abdominal measurements

As proof of principle, we performed liver MRE in three healthy volunteers using the liver-MRE model gravitational transducers. Scanning was performed under an IRB approval for sequence validation and informed consent was obtained from all volunteers. Liver disease was ruled out by a thorough medical history by an MD (JHR), no blood tests were performed. The gravitational transducer was strapped to the right flank overlying the liver, with a medical device class 2a gel pad to mould to the rib cage and body contour and to provide comforting cushioning. While absorbing shear motion, the gel pad will transmit without any attenuation the compressional motion sent by the transducer. In this lateral position, the transducer does not influence RF coil positioning. Data were acquired with the same fractionally encoded gradient echo MRE acquisition protocol as the phantom experiment with the following settings: $f_{\rm VIB} = 40 | 50 | 60 \text{ Hz}$, $f_{\rm MEG} = 165 \text{ Hz}$, TE = 6.9 ms (in-phase) and TR = 81 | 85 | 71 ms with 3 | 2 | 2 image shots per vibration period and 4 wave images. Readout direction of these transverse scans was in the AP direction. Reconstructed voxel size was $4.0 \times 4.0 \times 4.0 \text{ mm}^3$ with acquisition and reconstruction matrices of $76 \times 53 \times 8$ and $80 \times 80 \times 8$, respectively. SENSE factor was set to 1.4. The readout bandwidth was 243 kHz. The acquisition times per encoding direction were 18 | 19 | 16 s, respectively. Data were processed similarly to the phantom data.

Results

Transducer and controller characterisation

The custom-made functions for the stepper motor controller allowed control of the stepper motor's rotational velocity and position with high accuracy (see figure 3). The median (interquartile range) eccentric mass' angular position for each subsequent TTL-signal is 0° (-1.4° to 2.7°) relative to the reference position (i.e. the start position stored at reception of the first TTL-signal after the dummy shot(s)). Without stabilisation, there is a 69° min⁻¹ shift. The loss of synchrony between MRE acquisition protocol and rotating mass introduces uncorrectable biases in the phase data, strongly hampering reliable biomechanical properties reconstruction.

The accuracy of stepper motor control is further highlighted in figure 4, which shows the results of table top frequency response measurements. Panels (A) and (B) show the acceleration in units of g and amplitude in mm for the brain transducer with a 16.52 g mass at f_{VIB} of 40, 50, 60 and 70 Hz. Note in B how acceleration increases quadratically with f_{VIB} , where the fit-line of $m \cdot r \cdot \omega^2$ in blue through the maxima yields a R^2 of 0.992 and a realistic value of r of 0.0104 m. Panel (C) compares the 50 Hz spectra of the commercially available air-based transducer (line (A)) with both the brain (lines (B)–(E), with increasing rotating mass respectively) and liver (line (F)) gravitational transducer. All transducers showed a main peak on resonance with a full width at half maximum of ~0.5 Hz. Whilst the air-based method (line (A)) has a sturdy punch on resonance, it also produces serious peaks at the upper harmonics. For the gravitational transducers (lines (A)–(F)), the noise occurs mainly on non-harmonic frequencies due to inaccuracies in the pulley system and rotational axis, indicating further room for refinement. Percentagewise the 2nd harmonics at 150 Hz for (A)–(F) were respectively 17.9%, 7.7%, 5.8%, 3.8%, 1.6% and 3.0% of the on resonance amplitudes. Finally, panel (D) indicates the linear increase in acceleration with increase in mass volume (R^2 of 0.998), in line with the theory behind the system.

Numerical simulations

Figure 5 shows the results of the numerical simulations, where figure 5(A) is the liver mesh, and figures 5(B) and (C) show the displacements for 40 Hz and 120 Hz waves, respectively. The measurement of wave displacements at four time points over the cycle of the 40 Hz wave allows to obtain 40 Hz signal as well as the 1st harmonic (80 Hz) using a voxel-wise Fourier transform. The 3rd harmonic (160 Hz) adds only to the zero-frequency component as it cycles four times per 40 Hz cycle, while the 2nd harmonic (120 Hz) will corrupt the 40 Hz signal and lead to pollution of the derived elasticity values. Figures 5(D) and (G) show the reconstructed elasticity and viscosity with a 40 Hz signal in absence of noise. Figures 5(E) and (H) show the same but with added (pseudorandom, uniformly distributed) noise ($\pm 3\%$ of the transducer amplitude). The ROI in figure 5(E) indicates the approximate region, which is still accurately reconstructed. Figures 5(F) and (I) show the elasticity and viscosity when a 2nd harmonic at 120 Hz is added to the 40 Hz signal at 20% of the intensity and with the same added



Figure 3. Angular position of rotating mass. Plot of the angular position of the rotating eccentric mass as function of the total number of TTL-signals received by the stepper controller for nominal settings (*red*), +1 microstep s⁻¹ offset (*yellow*) and nominal with stabilisation (*blue*). Note how nominal position starts to deviate after approximately 10 TTL-signals, while already a +1 microstep s⁻¹ causes an immediate deviation, with substantial angular offset (69° after 70 TTL-signals = ~1 min) and consequently phase errors. With stabilisation, the mean (interquartile range) angular position is very stable with median (interquartile range) position: 0° (-1.4° to 2.7°).

noise. The ROI in figure 5(F) highlights the region that is significantly impacted by the 2nd harmonic which is shown to be a large portion of the (in absence of the 2nd harmonic) accurately reconstructed region.

Phantom measurements

The elastography phantom contains four large (\emptyset 20 mm diameter) and four small (\emptyset 10 mm) round inclusions of known elasticity. The MRE imaging plane was centred on the large inclusions, perpendicular to the gravitational transducer. Figure 6(A) shows the magnitude image of the two central slices through the large inclusions. Figures 6(B)–(D) show the (B) accompanying *x*-component of the curl of the displacement, (C) elasticity (kPa) and (D) nonlinearity (%) maps. ROI measurements in the inclusions and background as found here and as reported by the manufacturer are shown in table 1.

In vivo human abdominal measurements

In vivo liver MRE data were successfully acquired in healthy volunteers. None of the volunteers reported any ill effects of or pain caused by the vibrations. Figure 7 illustrates MRE magnitude images, total wave amplitude, *z*-component of the displacement curl, nonlinearity, elasticity and viscosity maps for $f_{\rm VIB} = 50$ Hz for a single volunteer (supplementary figure E1 shows abdominal data for two volunteers as well as an example of brain elastography (stacks.iop.org/PMB/64/045007/mmedia)). The elasticities of liver and spleen were in concordance with normal values reported elsewhere (Mannelli *et al* 2010, Bohte *et al* 2013) for healthy individuals at 1.73 (±0.04) and 2.78 (±0.52) kPa, respectively. Note the deep transmission of the shear waves, with >20 μ m waves present in the spleen even with the transducer strapped to the right flank and the low level of nonlinearity in liver (7%) and spleen (14%). Also note that the map of viscosity shows rather homogeneous distributions in liver and spleen with average values of 0.47 kPa and 0.87 kPa (volunteer 1), respectively.

Discussion

Here, we present a novel MR Elastography transducer concept based on gravitational forces generated by an eccentrically rotating mass: it produces highly accurate vibrational waves without relevant 2nd harmonic vibrations and can be accurately phase-locked with an MRE sequence. The concept combines the accuracy of EM transducers with the artefact-free generation of shear waves in the body of pneumatic approaches. In addition, the use of a rotational eccentric mass results in a constant transducer casing vibration amplitude which is independent of the driving frequencies as opposed to other transducers where the amplitude decreases with increasing driving frequency. The practical consequence is that waves are transmitted deep into the body or organ parenchyma, even at higher frequencies where the decreased amplitude typically limits other transducers. Certainly, other transducer concepts such as the commercially available device (using pneumatic air and a



Figure 4. Frequency response of the gravitational transducer. Plot of transducer casing acceleration in units of g (panel (A)) and displacement amplitude in mm (panel (B)) for four different driving frequencies $f_{\rm VIB}$ (40, 50, 60 and 70 Hz) for a 16.52 g mass. Note the constant displacement amplitude in (B) over a range of $f_{\rm VIB}$, as predicted by the theory for a rotating eccentric mass and the absence of substantial upper harmonics (location of the 2nd harmonics indicated by arrows). Panel (C) shows the 50 Hz spectra of an air-based commercially available (line (A)) and five gravitational transducer spectra (lines (B)–(F)) with either a 1:2 gear ratio ((B)–(E): 7.45 | 10.48 | 13.5 | 16.52 g mass) or a 1:3 gear ratio ((F): 38.24 g mass). Panel (D) shows the linear increase in transducer acceleration with increase in mass.

passive acoustic driver) can compensate the loss in driving amplitude at higher frequencies by simply increasing the input power. Unfortunately, this typically leads to enhanced and pronounced nonlinearities inherent to acoustic systems. Recently, Neumann *et al* (2018) reported their approach by combining an eccentric mass with a pneumatically driven turbine to generate a centrifugal force, very similar to our concept presented in 2017 (Runge *et al* 2017). They demonstrate forces up to approximately 2 g (~20 m s⁻²) which is ten-fold smaller than those generated by our liver transducer which enables efficient transmission of waves deep into the abdomen. It remains to be shown whether the increased air pressure necessary for a translation of the pneumatic approach to human applications remains stable and safe for patients, and whether the absolute phase of the system can be correctly controlled, which is a mandatory requirement for multiple breath-hold scans in liver applications. Compared to the axis presented earlier, our current rotating axis is made from PEEK and is sectioned. This allows for a very modular design in terms of length and an axis that is extremely flexible and bendable, but robust and highly reliable in terms of absolute phase stability under many conditions (temperature, bending, etc).

The unique characteristic of sustained penetration at higher vibrational frequencies can be used to assess the behaviour of elasticity values over a range of driving frequencies, as the measured elasticity changes with the applied frequency. For instance, measured elasticity is impacted by the presence of micro-obstacles such as microvasculature (Juge *et al* 2015). Hence, by measuring the effect of driving frequency on elasticity values, it can become possible to detect underlying properties. Furthermore, the basic concept of the gravitational transducer allows for intrinsic, i.e. simultaneous, multi-frequency experiments (not performed here) by adding one or more rotational eccentric masses with different gear ratios. Currently, multi-frequency experiments are either performed sequentially (as shown in this manuscript), or using a combined waveform (as shown by Garteiser *et al* (2013)). The former doubles or triples scan time, while the latter requires thorough preparation as the unweighted addition of multiple waveforms leads to an uneven contribution of frequencies in the output, where the lower frequencies typically dominate. To overcome this, weighted addition of the waveforms leads to an even contribution of all waveforms. This same problem would also occur for the gravitational transducer and requires



upper narmonics on viscoelastic reconstruction. (A) The domain of the simulation, extracted onto an image grid. The simulations were run at a higher resolution. (B) Wave displacement in *y*-direction at 40 Hz. (C) Wave displacement in *y*-direction at 120 Hz, representing the 2nd upper harmonic of the 40 Hz wave. (D) and (G) Reconstructed elasticity and viscosity from a 40 Hz signal with no noise. (E) and (H) Reconstructed elasticity and viscosity from a 40 Hz signal with added pseudorandom noise (\pm 3% of the transducer amplitude). The ROI indicates the approximate region, which is accurately reconstructed. (F) and (I) Reconstructed elasticity and viscosity from the 40 Hz signal with a 120 Hz signal added at 20% of the intensity and with added noise to simulate the presence of upper harmonics. The ROI indicates the approximate region which is significantly affected by the upper harmonic.

careful adjustment of the additional eccentric masses to generate the same centrifugal force for all frequencies. When properly implemented, the intrinsic multi-frequency approach could save valuable scan time.

We showed that the gravitational transducer is highly accurate as indicated by strongly suppressed upper harmonics compared to the commercially available transducer. This was accomplished by a stepper motor set-up that includes feed-back loops to account for lag in the control system and motor, nulling any angular position drift. The numerical simulations highlighted the importance of the pure character of the gravitational transducer: higher harmonics can seriously hamper proper viscoelastic reconstruction. While this issue can be overcome by measuring the wave at additional time points, and thereby properly accounting for the 2nd harmonic, this would lead to increased scan times that in breath-hold scenarios (e.g. liver, spleen, pancreas) is clearly undesired. Furthermore, the gravitational transducer delivers constant amplitude over a wide range of driving frequencies typical to the clinical application of MRE (i.e. 40–70 Hz). A breast (Hoelzl *et al* 2017) and brain version of the transducer are currently being tested in the context of a large scale EU-funded Horizon2020 project (FORCE). These developments indicate some of the many potential applications where a unique design of a transducer casing allows targeting specific organs of interest.

In this proof-of-principle paper, we successfully applied the transducer concept on a phantom and *in vivo*. The phantom data show an underestimation of the stiffer inclusions. This has been reported previously (Sahebjavaher *et al* 2014), with several factors playing a role including limitations of the inversion algorithm, ageing of the material, dynamic (this paper) compared to static (manufacturer) loading and temperature differences between the experiments. Given that ageing, dynamic versus static loading and temperature will affect all inclusions and that the softer inclusions were not under- or overestimated to the extent of the stiffer inclusions, limitations of the inversion algorithm are the most likely factor as numerical computations become less accurate with increasing wavelength since noise will begin to dominate calculations such as derivatives. Hence, the applied frequency directly affects reconstruction accuracy as larger wave lengths are present at lower driving frequencies (e.g. 40 Hz). For our phantom experiments, we therefore applied the highest possible frequency with the currently available masses with the gravitational transducer (90 Hz).

One important aspect of the gravitational transducer relates to safety. Several key concepts are in place as safeguards: (1) should the rotational eccentric mass jam for any reason, the suddenly strongly increased torque will



Figure 6. Elastography phantom MRE experiment at 90 Hz. Results of the MRE experiment at 90 Hz on a dedicated elastography phantom. Panel (A) shows the average magnitude image of the two central slices containing the four larger inclusions. Panels (B)–(D) show the accompanying (B) *x*-component of the displacement curl, (C) elasticity (kPa) and (D) nonlinearity (%). Mean nonlinearity in the phantom was 8%, with outliers centrally in the area where standing waves occur due to reflection. Elasticity values for the inclusions and background material are shown in table 1.

Table 1. Elasticity values of dedicated elastography phantom in
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	Manufacturer ^a	This study
Background	3.7	2.9 ± 0.23
1 (top)	1.3	1.6 ± 0.16
2	2.2	2.1 ± 0.14
3	7.8	4.2 ± 0.15
4 (bottom)	12.6	6.4 ± 0.10

^a Provided as Young's modulus hence divided by 3 to convert to shear modulus for comparison purposes.

stall the stepper motor; (2) the eccentric mass consists of a single piece of CNC-lathed PTFE fixated to a PEEK rod by glue; should the glue bond fail, the rotating axis would spin inside and not turn the mass; moreover, there is no room for the rotating mass itself to fully detach from the rotating axis; (3) the transducer casing is made of a very strong plastic closed tightly using strong and resilient PEEK screws, ensuring that in the very unlikely event that any part of the mass should come loose, little damage would be sustained by the casing. With respect to the eccentric mass, we plan to further improve this part by moving to a single piece of CNC-lathed PTFE that includes both rod and mass parts. This design would inherently preclude the possibility of the eccentric mass coming loose inside the casing.

With certain eccentric mass volumes, the generated force becomes particularly strong at higher f_{VIB} . Hence, object and desired f_{VIB} have to be matched in order to generate the best quality data possible: when the mass of the driver m is too large compared to *M*, this becomes detrimental to data quality (i.e. *m* should be a maximum of 10% of *M*). Optimising particular applications requires more extensive testing with differing eccentric mass volumes and/or density. As the modular design of the gravitational transducer allows straightforward replacement of the eccentric mass, a range of eccentric masses can be provided for easy, patient- and/or organ-specific transducer modification due to the lock-nut connections. Of note, the gravitational transducer is explicitly meant for *in vivo* human applications with f_{VIB} in an approximate range of 40–70 Hz, depending on the size of the mass. For example, at the upper end of this frequency range the generated force for the 38 g (liver) mass verges on becoming unbearable in subjects with normal body composition and we only use it at that frequency for obese individuals. In terms of the generated tissue displacement, Ehman *et al* showed that for an MRE examination at 60 Hz of 15 min the maximum allowable whole-body displacement according to EU law is 243 μ m (Ehman *et al* 2008).



Figure 7. *In vivo* MRE upper abdomen experiment of liver and spleen. Results of the *in vivo* MRE experiment for a single volunteer with the gravitational transducer placed over the liver with $f_{\rm VIB}$ of 50 Hz. Panel (A) shows the MRE magnitude image of the three central slices. Panels (B)–(D) show (B) the corresponding total displacement amplitude (μ m), (C) *z*-component of the curl of the displacement, (D) nonlinearity (%), (E) elasticity (kPa), and (F) viscosity (kPa). Displacement amplitudes are displayed on a log scale and are of the order of 200 μ m at the transducer. Due to mode-conversion from compressional to shear waves, a total amplitude around 20 μ m is still recorded on the opposite side of the body. Liver and spleen ROIs are shown in red and green, respectively. Nonlinearity is very low in the parenchymal organs of interest in these slices, i.e. liver and spleen and both organs are well visibly in the viscoelastic maps.

Our *in vivo* data show that even with the largest mass of 38 g, the gravitational transducer does not generate displacement over 200 μ m inside the liver at 50 Hz. Moreover, when a more realistic duration of 6 min is used for the calculation, the upper limit becomes 384 μ m.

As the purpose of this study was to prove the principle of this transducer concept, we did not perform a headto-head comparison with other transducer designs. Formal comparison studies are planned for the near future in both healthy volunteers as well as patients to determine whether the gravitational transducer provides substantially improved data for more reliable viscoelastic reconstruction.

The gravitational transducer for MR Elastography wave generation as presented here combines the benefits of two previous concepts: the EM approach with highly accurate on-resonance f_{VIB} and the acoustic approach with easy positioning. In addition, it lacks specific drawbacks of those concepts, such as image artefacts (EM approach with engine-coil close to imaged FOV); low accuracy and the presence of upper harmonics (acoustic approach). Uniquely, the gravitational transducer delivers a constant transducer casing displacement amplitude as f_{VIB} increases, completely opposite to the effect seen in other transducer concepts. The combination of these characteristics make the gravitational transducer well-suited for MR Elastography applications in a wide range of clinical contexts. Further clinical studies that include head-to-head comparisons are required to determine how the gravitational transducer functions relative to currently available transducers.

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