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Monitoring MR-guided High Intensity Focused Ultrasound therapy using transient Supersonic Shear Wave MR-Elastography

Authors

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Abstract

Objective

The aim of the paper is to propose an all-in-one method based on Magnetic Resonance-Supersonic Shear wave Imaging (MR-SSI) and Proton Resonance Frequency Shift (PRFS) to monitor High Intensity Focused Ultrasound (HIFU) thermal ablations.

Approach

Mechanical properties have been shown to be related to tissue damage induced by thermal ablations. Monitoring elasticity in addition to temperature changes may help in ensuring the efficacy and the accuracy of HIFU therapies. For this purpose, an MR-SSI method has been developed where the ultrasonic transducer is used for both mechanical wave generation and thermal ablation. Transient quasi-planar shear waves are generated using the acoustic radiation force, and their propagation is monitored in motion-sensitized phase MR images. Using a single-shot gradient-echo echo-planar-imaging (EPI) sequence, MR images can be acquired at a

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3 sufficiently high temporal resolution to provide an update of PRFS thermometry and MR-SSI
4 elastography maps in real time.
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7 8 **Main results** 9

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11 The proposed method was first validated on a calibrated elasticity phantom, in which both the
12 possibility to detect inclusions with different stiffnesses and repeatability were demonstrated. The
13 standard deviation between the 8 performed measurements was 2% on the background of the
14 phantom and 11%, at most, on the inclusions. A second experiment consisted in performing a
15 HIFU heating in a gelatin phantom. The temperature increase was estimated at 9 °C and the shear
16 modulus was found to decrease from 2.9 to 1.8 kPa, reflecting the gel softening around the HIFU
17 focus, whereas it remained steady in non-heated areas.
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28 **Significance** 29

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31 The proposed MR-SSI technique allows monitoring HIFU ablations using thermometry and
32 elastography simultaneously, without the need for an additional external exciter as those used in
33 MR Elastography.
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42 **Keywords:** MR Elastography, Supersonic shear wave, Thermal ablation, HIFU therapy, Therapy
43 monitoring, Interventional MRI
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Introduction

High Intensity Focused Ultrasound (HIFU) is a non-invasive, non-ionizing thermal ablation method that has shown great potential for cancer treatment, when combined with appropriate and accurate monitoring methods (Geoghegan *et al* 2022). To date, Magnetic Resonance Thermometry (MRT) is considered as the gold standard for non-invasive monitoring of temperature changes during thermal ablations. Proton Resonance Frequency Shift (PRFS) MRT is a relatively tissue-independent temperature estimation method, with the exception of adipose tissue (Ishihara *et al* 1995, Quesson *et al* 2000, Rieke and Butts Pauly 2008). In practice, temperature is used as an instantaneous biomarker of the ablation, but this information does not directly inform the physician about the structural integrity of the tissue (Odéen and Parker 2019). In theory, thermal dose can provide such information (Sapareto and Dewey 1984), but it has the drawback of being highly sensitive to noise since it relies on the integration of temperature over time. Therefore, systematic errors on temperature will also be integrated in time and thus, may lead to amplified errors on thermal dose estimates. Therefore, complementary biomarkers are needed to ensure permanent feedback on the state of the tissue during and after ablation.

It has been shown that mechanical properties are sensitive to tissue damage caused by heating and thus, can be used as a complementary biomarker for HIFU monitoring (Arnal *et al* 2011, Sapin-de Broses *et al* 2010, Corbin *et al* 2016, Chen *et al* 2014, Mariani *et al* 2014, E. Konofagou *et al* 2012). Elastography monitoring methods were first introduced in ultrasound imaging to assess tissue damage during thermal ablations (Arnal *et al* 2011, Mariani *et al* 2014). Magnetic Resonance Elastography (MRE), initially used for diagnostic purposes (Lewa 1991, Muthupillai *et al* 1996), has also been shown to be applicable for monitoring thermal ablations. MRE has been used for assessing elasticity changes before and after the procedure (Chen *et al* 2014, Wu *et al*

2001), as well as during the ablation in real time (Corbin *et al* 2016, Yuan *et al* 2007). Nonetheless, MRE requires a mechanical exciter for shear wave generation, which significantly complicates the experimental setup of the MR-guided HIFU procedure. From this perspective, Acoustic Radiation Force (ARF)-based harmonic MRE approaches have been proposed where the mechanical excitation is generated using the radiation force of the focused ultrasound itself with no need for supplementary equipment (Wu *et al* 2000, Chatelin *et al* 2016, Vappou *et al* 2018b). Yet, these harmonic MRE approaches are limited because of shear wave near-field effects, making them highly inaccurate in the central zone around the ultrasound focus.

Several transient ARF-based methods have been proposed as an alternative (Souchon *et al* 2008, Nightingale *et al* 2002, McDannold and Maier 2008, Larrat *et al* 2010, Bour *et al* 2017, de Bever *et al* 2018, Hofstetter *et al* 2019, Bercoff *et al* 2004, Liu *et al* 2015, 2017). ARF Imaging (ARFI) is a transient elastography method that was first introduced in ultrasound imaging (Nightingale *et al* 2002). MR-ARFI was initially proposed for the targeting of MR-guided HIFU, since it allows visualizing the focal spot on the image (McDannold and Maier 2008, Larrat *et al* 2010). Several studies have reported the use of the MR-ARFI displacement to monitor thermal ablations, relying on the fact that this displacement is related to the underlying tissue elasticity (Auboiroux *et al* 2012, Bour *et al* 2017, de Bever *et al* 2018). One limitation of these methods is that MR-ARFI displacement is not only related to local tissue elasticity, but also to the surrounding mechanical environment and to potential changes of tissue acoustic absorption (and hence radiation force) due to temperature. This has led to the development of shear-wave-based methods, as shear wave velocity (c) is quantitatively related to the underlying shear modulus (μ) by:

$$\mu = \rho c^2 \quad (1)$$

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3 with ρ being the density (considered to be equal to $1000 \text{ kg}\cdot\text{m}^{-3}$) and assuming that the biological
4 tissue is purely linear elastic, isotropic, locally homogeneous, and incompressible (Manduca *et al*
5 2001). Hofstetter *et al* proposed an ARF-based method that consists in applying multiple ARF
6 pushes at several positions to allow for composite shear modulus mapping and real-time
7 elastography monitoring during HIFU procedures (Hofstetter *et al* 2019). An alternative method
8 to map mechanical properties over an extended area is the Supersonic Shear wave Imaging (SSI)
9 technique proposed by Liu *et al* (Liu *et al* 2015, 2017) based on the ultrasonic SSI method (Bercoff
10 *et al* 2004). It consists in generating quasi-planar shear waves in the tissue with ARF and tracking
11 their spatial propagation over time, so as to compute the shear wave velocity map from which an
12 elastogram can be deduced. Although the MR-SSI method was shown to be particularly accurate
13 with high field small animal preclinical setup (Liu *et al* 2015, 2017), reported acquisition times of
14 several minutes prevent this method from being used in its current form for real-time HIFU
15 monitoring.

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34 The aim of this work is to develop a new HIFU ablation monitoring method capable of providing
35 temperature and elasticity maps simultaneously and in real time, both information being derived
36 from the same MR phase images. In the next section, we present the experimental setup and the
37 developed algorithm for MR-SSI elastography. Then, we expose the results obtained on a
38 calibrated phantom as well as on a gelatin phantom undergoing HIFU heating.
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Methods

Focused ultrasound sequence

As introduced by Bercoff *et al*, the SSI method is based on the generation of a quasi-planar mechanical wave in the medium (Bercoff *et al* 2004). Thanks to the electronic steering abilities of the multi-element ultrasound transducer, several ARF pushes are generated successively, hence creating multiple quasi-punctual shear wave sources. When the pushes are generated at a sufficiently high speed, these waves interfere and form a quasi-planar shear wave, the propagation of which reflects the mechanical properties of the tissue. Subsequently, the HIFU heating is performed directing the focus a few millimeters away from the SSI push line (Figure 1). This set of pushes constitutes a cycle that will be repeated for each MR image acquisition (i.e., each TR) during the whole heating procedure. Once the heating completed, SSI pushes alone will be generated to maintain elasticity monitoring during the cooling down period.

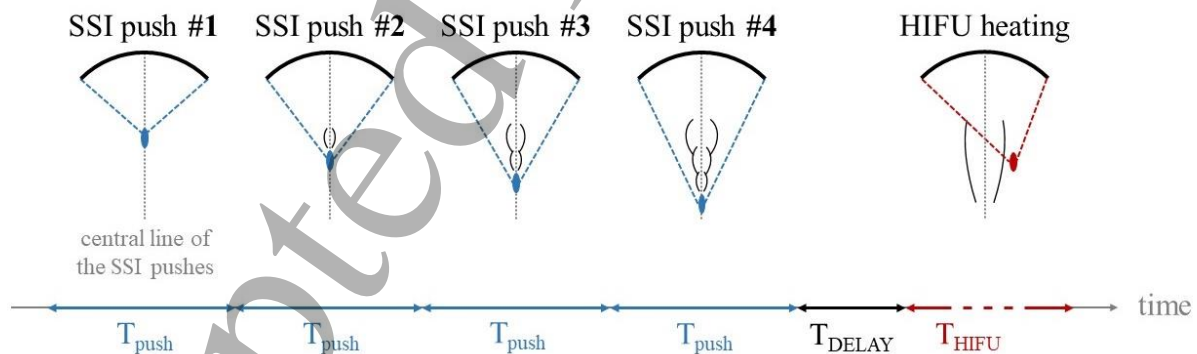


Figure 1 – Operating sequence of the focused ultrasound transducer. A quasi-planar supersonic shear wave is generated using 4 successive SSI pushes aligned on a straight line, followed by the HIFU heating located a few millimeters on the side of the SSI push line. Note that T_{HIFU} is sensitively longer than T_{push} (typically hundreds of ms vs a few ms). This time sequence is repeated in each MR image acquisition, i.e., each TR.

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3 As the generated mechanical wave is cylindrical, the wavefronts will have different appearances
4 depending on the orientation of the imaging plane (Figure S1 – supplementary material). When
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6 the imaging plane is parallel to the SSI push line, the shear wave appears as 2 quasi-planar
7
8 wavefronts propagating in opposite directions. When the imaging plane is perpendicular to the SSI
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10 push line, a circular wavefront is observed.
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14 15 **MR-SSI sequence** 16

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18 The displacements induced by the quasi-planar shear waves are encoded with bipolar Motion-
19 Sensitizing Gradients (MSG) added to a single-shot gradient-echo echo-planar-imaging (EPI)
20 sequence. The bipolar MSG is synchronized with the mechanical excitation using a variable time
21 offset Δt_{TRIG} ; thus, tissue displacement is encoded at different moments of the wave propagation,
22 below designated by “phase offsets” (Figure 2). Knowing the displacement field induced by the
23 shear wave, the wave equation can be used to estimate the shear modulus of the underlying
24 medium. It should be noted that for each Δt_{TRIG} value, two acquisitions are performed with opposed
25 MSG polarities. The average between these two phase images is used for PRFS thermometry, since
26 they bear opposite motion encoding that null each other, whereas their difference is used for shear
27 wave reconstruction, which doubles the displacement encoding and suppresses the background
28 phase in the phase difference image (Corbin *et al* 2016, Le *et al* 2006a, 2006b).
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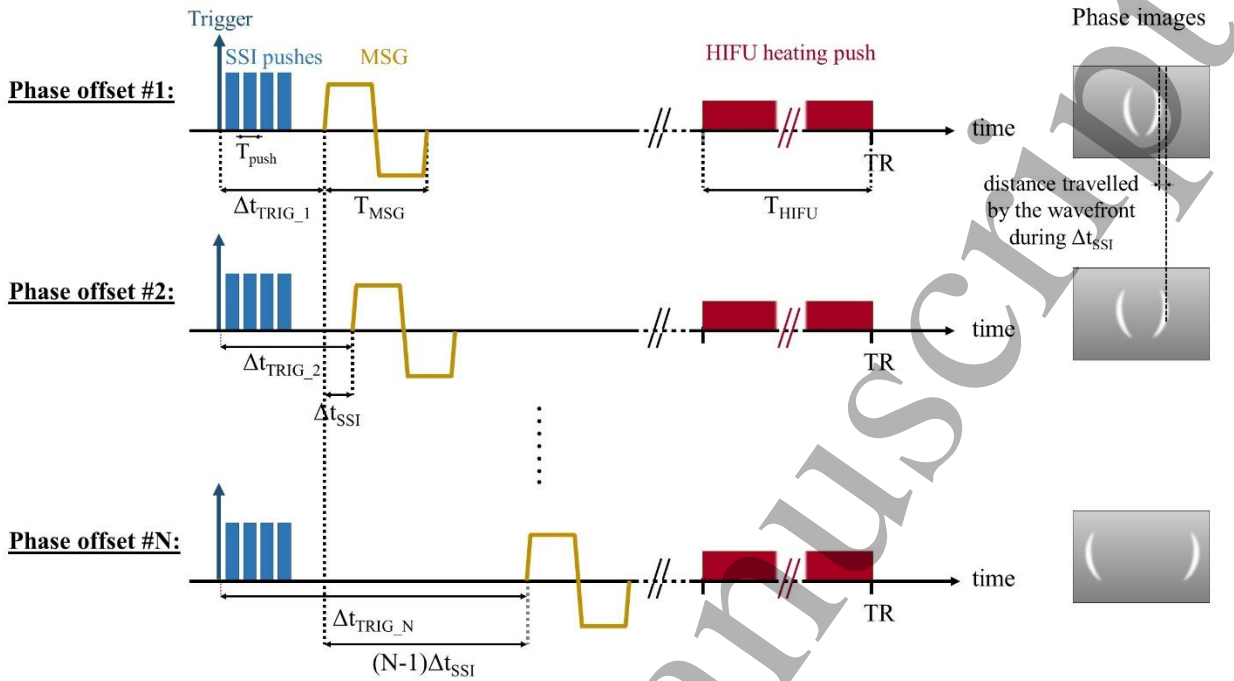


Figure 2 – Illustration of the acquired images at different SSI steps. By varying the offset Δt_{TRIG} between the SSI pushes and the encoding gradient (MSG), different phases of the propagating shear wave can be captured. Note that for simplicity, only one of the two MSG polarities has been represented. In practice, one wavefront image is obtained from the difference between two phase images acquired with the same Δt_{TRIG} but with opposed MSG polarities.

Post-processing and elastogram reconstruction

Elastograms are reconstructed using an edge detection approach followed by a time-of-flight-based method for the computation of the shear wave velocity (McCracken *et al* 2005, Liu *et al* 2017). The whole process is implemented using Matlab (The Mathworks Inc., MATLAB R2021a).

Inputs

The processing program is provided with phase difference images acquired for a set of Δt_{TRIG} values (Figure 3A). Thus, the main input of the process is a 3-dimensional matrix containing displacement images over time (2 spatial dimensions and 1 temporal dimension).

Preparation of the data and wavefront enhancement

The spatial domain is truncated to a region of interest (ROI) containing shear wavefronts. A Hessian-based multiscale filtering approach is applied to enhance the wavefronts on the images (Frangi *et al* 1998).

Shear wave velocity estimation on each line of the field of view (FOV)

To estimate the shear wave velocity, the position of each wavefront is tracked across all the propagation lines. In other words, the shear wave velocity is mapped on all the lines that are perpendicular to the wavefront. For each line, a chronogram represents the line in question over the different phase offsets sampling its propagation over time (Figure 3B). An edge detection approach (Canny filter (Canny 1986)) is implemented to delineate both edges of the wavefront (Figure 3C). These edges are referred to as “inner” and “outer” edges with respect to the SSI push line. For both edges, the arrival time of the wavefront is represented against its distance from the SSI push line. This curve is smoothed with a local regression (Figure 3D), then the local slope is estimated using a simple differential approach (Figure 3E). Finally, the shear wave velocity is locally deduced as it is inversely proportional to the previously estimated slope (Figure 3F).

Elastogram reconstruction

Depending on the imaging slice, the process will be repeated either in Cartesian or radial lines of the image. Once the shear wave velocity is estimated on all the lines of the image, we obtain two distinct maps: one using the inner edges of the wavefront and the other one using the outer edges (Figure 3G). A compound map is then reconstructed from these complementary maps, with a mean value taken on the pixels where both maps overlap (Figure 3H). The shear modulus map is deduced

using equation (1). Finally, a median filter (4×4 kernel) is applied to attenuate potential outliers (Figure 3I).

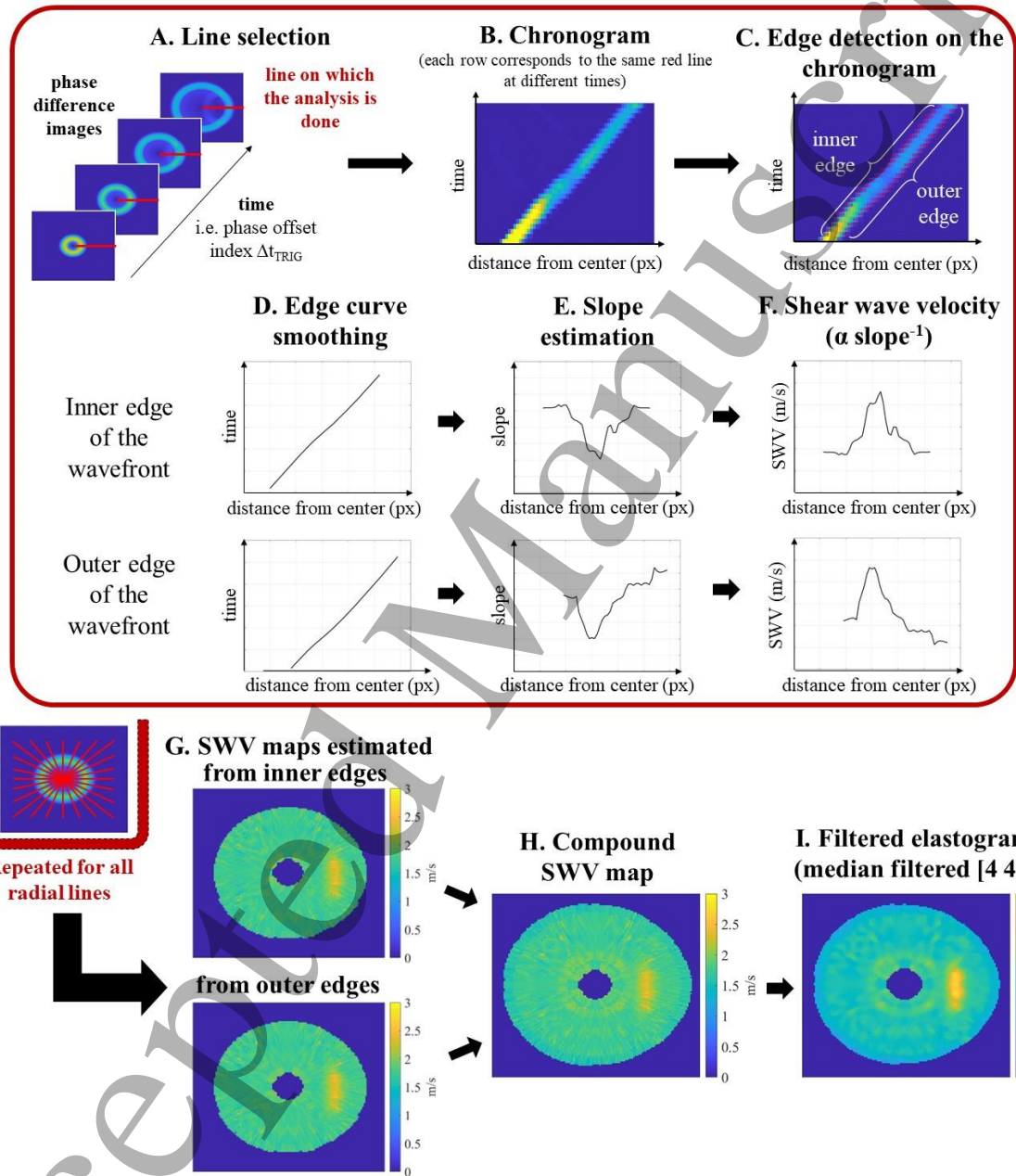


Figure 3 – MR-SSI reconstruction process. Illustration in the case of a circularly propagating SSI wavefront, i.e. the imaging plane is parallel to the SSI push line. (A) For each radial line (or Cartesian line in the case of linearly propagating SSI wavefronts, i.e., the imaging plane is perpendicular to the SSI push line), (B) a chronogram is built

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3 showing the selected line over the different SSI steps. (C) An edge detection step allows identifying inner and outer
4 edges of the wavefront in the chronogram, each edge representing a 1D curve. After (D) smoothing the curve, (E) a
5 slope estimation leads to (F) the estimation of the shear wave velocity. After applying steps A-E to all the lines of
6 the FOV, (G) two shear wave velocity (SWV) maps are obtained, one corresponding to the inner and the other one
7 to the outer edges of the wavefronts. (H) A compound SWV map is computed from which (I) the final elastogram
8 can be deduced. The images on this figure correspond to a simulated dataset on a 3 kPa-numerical phantom
9 containing a 5 kPa-inclusion.
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17 **Experimental protocol**

18 **MRI acquisitions**

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21 All MR experiments were performed in a 1.5 T MRI system (MAGNETOM Aera, Siemens
22 Healthcare, Erlangen, Germany) using a combination of Spine and 18-element body matrix coils.
23 The single-shot gradient-echo EPI sequence was modified for real-time monitoring with MR-SSI.
24 Relevant parameters for the acquisition were: TR/TE 500/29 ms, FOV 256 mm × 176 mm, slice
25 thickness 5 mm, matrix 128 × 88 (80% phase lines), MSG duration 10 ms at 30 mT/m, bandwidth
26 1502 Hz/px, flip angle 55°, GRAPPA factor 2.
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38 **SSI experimental setup**

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41 A 128-element spherical HIFU transducer (1 MHz central frequency, 6 cm radius of curvature, 6
42 cm in aperture; Imasonic, Voray sur l'Ognon, France) was used to generate SSI pushes and HIFU
43 heating (Figure 1) driven by the MR-compatible HIFU generator (Image Guided Therapy, Pessac,
44 France). The transducer was positioned on the upper surface of the phantom with its focal axis
45 perpendicular to the phantom surface. The acoustic coupling between the transducer and the
46 phantom was realized using a balloon filled with degassed water, and a thin layer of degassed
47 water over the phantom surface.
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Proof-of-concept of MR-SSI in CIRS phantom

Phantom

The first experiment was realized on a commercial elastography phantom (Computerized Imaging Reference Systems, Inc. (CIRS), Norfolk, USA). The phantom contains 4 inclusions of 2 cm in diameter, having each a different shear modulus. The shear modulus values estimated with MR-SSI were compared to those provided by the manufacturer in the datasheet (obtained by static indentation), as well as to the results obtained using conventional MRE.

MR-SSI

The SSI wave was provoked by four pushes respectively at 50, 60, 70 and 78 mm from the top of the transducer active surface. Each push lasted 1.6 ms with an estimated acoustic power of 230 W.

MR-SSI images were acquired in a plane perpendicular to the SSI push line, corresponding to a coronal slice for a patient in supine position. The imaged slice was chosen to cross all 4 inclusions and, at the same time, to pass through the center of the SSI pushes (i.e., at approximately 64 mm from the top of the transducer active surface). At this point of the study, we could not determine yet the number of SSI steps necessary for the shear modulus reconstruction. Therefore, a total of 46 phase offsets were acquired with Δt_{TRIG} starting from 0 ms to 22.5 ms with increment $\Delta t_{\text{SSI}} = 0.5$ ms. The longer Δt_{TRIG} were chosen so that all 4 inclusions were covered by the propagating shear wave. Table 1 summarizes the relevant temporal parameters used in the experiments.

This experiment had three objectives: (i) to assess the ability to map the 4 inclusions of the CIRS phantom, (ii) to evaluate the repeatability of the MR-SSI process and (iii) to investigate the number of required SSI steps. For the second objective, the acquisition was repeated 8 times. As for the

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3 last objective, we decimated the acquired dataset, and identified the minimal number of SSI steps
4 providing similar values to those obtained with the entire dataset.
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7 8 *MRE* 9

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11 The commercial MR Elastography system (Resoundant, Rochester, MN) was set at a mechanical
12 frequency of 150 Hz. Relevant parameters of the MRE gradient-echo sequence were: TR/TE
13 50/12.4 ms, FOV 300 mm \times 300 mm, slice thickness 5.5 mm, matrix 128 \times 128, MSG duration
14 6.7 ms at 28 mT/m, bandwidth 1000 Hz/px, flip angle 20°. The shear modulus was reconstructed
15 using a Local Frequency Estimation algorithm (Knutsson *et al* 1994, Manduca *et al* 1996) under
16 the assumption of a purely linear elastic material.
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25 26 *Definition of ROIs* 27

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29 For each inclusion, the shear modulus was averaged within an ROI defined on the magnitude
30 image. The same process was applied to the 8 datasets acquired for MR-SSI as well as to MRE
31 images.
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35 36 *Monitoring of HIFU heating in a gel phantom* 37

38 39 *Phantom* 40

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42 The second experiment consisted in monitoring a HIFU heating in a gelatin phantom using MR-
43 SSI elastography and PRFS thermometry. The phantom was made of 2 kg of water, 400 g of
44 condensed milk, and 100 g of gelatin. Before refrigeration, the mixture was degassed using a
45 vacuum chamber.
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MR-SSI

Images were acquired in a plane parallel to the SSI push line, corresponding to a transverse slice for a patient in supine position, centered at the ultrasound focus. SSI pushes were generated at depths of 42, 54, 66 and 78 mm from the top of the transducer active surface with a duration of 1.6 ms for each push and an estimated acoustic power of 230 W. The HIFU heating focus was located at a depth of 60 mm from transducer active surface (coinciding with the midpoint of the segment containing the SSI pushes) and 16 mm laterally from the SSI push line. The SSI push line and the HIFU heating focus are both contained in the chosen transverse monitoring plane. The HIFU heating shot was performed during 250 ms (duty cycle 50% compared to 1.3% for the SSI push sequence) at an estimated acoustic power of 92 W.

A pre-heating acquisition was performed to determine the phase offsets needed in order to cover the desired ROI with the spatial extent of the wave propagation. Based on the results of the first experiment and the pre-heating acquisition, a total of 10 SSI steps, ranging from $\Delta t_{\text{TRIG}} = 1.5$ ms to 15 ms with increment $\Delta t_{\text{SSI}} = 1.5$ ms (Table 1), were acquired for each SSI cycle, allowing the reconstruction of one elastogram. A temporal interpolation of factor 3 was used in the MR-SSI process.

Knowing that 2 images with opposite MSG polarities are acquired for each of the 10 SSI steps, and with a TR set to 500 ms, the elasticity map is updated every 10 s, corresponding to a refresh rate of 0.1 Hz. The refresh rate for thermometry was 1 Hz, because a thermometry map is obtained for each SSI step from averaging the 2 phase images acquired with opposed MSGs.

The HIFU heating phase lasted 120 s, followed by a cooling down phase of 120 s, in which the HIFU heating shot was not performed while SSI pushes continued to be applied so that both tissue temperature and elasticity were still measured.

Experimental data processing

Phase images were spatially unwrapped prior to SSI computations and temporally unwrapped for thermometry. B₀ field drift effects in PRFS were corrected by applying phase subtraction with respect to a non-heated ROI chosen close to the HIFU heating focus.

Time (ms)	T_{push}	T_{MSG}	Δt_{TRIG_1}	Δt_{TRIG_N}	Δt_{SSI}	T_{HIFU}
Validation on CIRS (N = 46)	1.6	10	0	22.5	0.5	No heating
Monitoring on gel (N = 10)	1.6	10	1.5	15	1.5	250

Table 1 – Relevant temporal parameters of ultrasound and MRI sequences (as defined in Figure 2). Recall

that N: number of SSI steps, T_{push} : duration of each of the 4 successive SSI pushes, T_{MSG} : duration of the MSG,

Δt_{TRIG_1} : temporal offset between the SSI pushes and the MSG at the first SSI step, Δt_{TRIG_N} : temporal offset

between the SSI pushes and the MSG at the last SSI step, Δt_{SSI} : temporal increment between 2 successive SSI steps,

T_{HIFU} : duration of the HIFU heating push.

Results

Validation on CIRS phantom

Circular wavefronts acquired with increasing phase offsets can be seen from left to right in Figure 4 with the corresponding detected edges represented in red (all images are provided in Video 1 – supplementary material). In the two phase difference images corresponding to longer phase offsets (Figure 4), one can notice that the wave propagates faster on the right side (green arrow), and slower on the left side (blue arrow). These observations are consistent with the location of the rigid and soft inclusions across the imaged slice, to the right and left side, respectively.

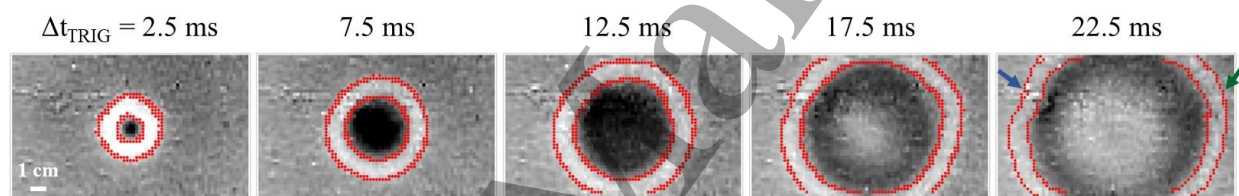
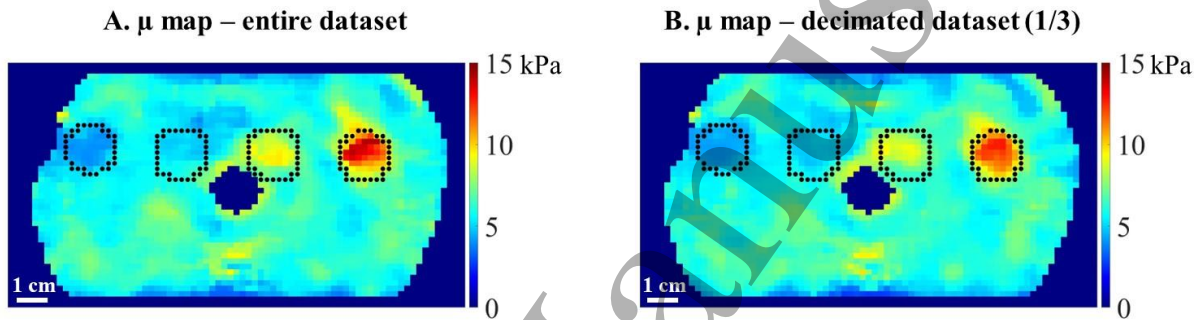


Figure 4 – Wavefronts observed at different SSI steps on the CIRS phantom. Phase difference images for increasing phase offsets (Δt_{TRIG}) between SSI push and MSG phase encoding. Red circles represent the detected inner and outer edges of the wavefronts. Arrows on the last frame show the regions where the wave propagated faster (green) and slower (blue) due to the presence of inclusions with different shear moduli.

By applying the SSI reconstruction process, we obtained the shear modulus map presented in Figure 5A (obtained with one of the 8 acquired datasets). Mean shear moduli in each ROI are given in Table 2 in kPa with standard deviations as percentages. Note that the values shown for MR-SSI represent the average of the 8 datasets (at each cell, are given the average of the 8 obtained shear moduli and the average of the 8 standard deviations).

The MR-SSI computed shear modulus of the background was 6.1 kPa (compared to 5.9 kPa from the datasheet). For inclusions I-IV, the shear moduli were measured to 4.5, 4.8, 7.3 and 11.5 kPa

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3 respectively. In comparison with the CIRS datasheet, MR-SSI provided higher values for soft
4 inclusions (types I and II) and lower values for rigid ones (types III and IV). A better consistency
5 was found between SSI and MRE results, with a maximum difference of 1.3 kPa between these
6 two methods. In terms of homogeneity, the shear moduli estimated with MR-SSI have standard
7 deviations of 15%, 12%, 22%, 22% and 17% for inclusions I-IV and the background respectively.
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Figure 5 – Shear modulus maps obtained on the CIRS phantom. (A) using the entire dataset and (B) the decimated dataset (decimation 1/3 then interpolation $\times 3$). Black circles represent the selected ROIs in the inclusions.

Inclusions are referred to as type I, II, III and IV (from left to right) by the manufacturer.

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Shear modulus (kPa)	Type I	Type II	Type III	Type IV	Background
CIRS datasheet	1.9	3.4	11.6	23.8	5.9
MRE (150 Hz)	3.6 $\pm 23\%$	3.7 $\pm 11\%$	7.5 $\pm 7\%$	10.2 $\pm 13\%$	6.1 $\pm 15\%$
MR-SSI (entire dataset)	4.5 $\pm 15\%$	4.8 $\pm 12\%$	7.3 $\pm 22\%$	11.5 $\pm 22\%$	6.1 $\pm 17\%$
MR-SSI (decimation 1/3 then interpolation $\times 3$)	4.5 $\pm 17\%$	4.8 $\pm 12\%$	7.2 $\pm 22\%$	11.6 $\pm 23\%$	6.1 $\pm 19\%$

Table 2 – Shear moduli for the 4 inclusions and the background of the CIRS phantom, obtained from the datasheet, MRE 150 Hz, MR-SSI with the entire datasets (phase offset 0 to 22.5 ms with 0.5 ms step) and MR-

SSI with the decimated datasets (1/3 followed by interpolation $\times 3$). Note that the values shown for MR-SSI

correspond to the average of the 8 datasets \pm the average of the 8 standard deviations in the ROI.

Figure 6 represents the boxplots obtained from the 8 shear modulus maps acquired on the CIRS phantom (using the full datasets i.e., without decimation). The standard deviations between the 8 results are of 4%, 9%, 6%, 11% and 2% for inclusions I-IV and the background respectively which shows the high repeatability of our process.

In order to increase the temporal resolution of MR-SSI for real-time monitoring, we tested whether the number of SSI steps could be reduced while maintaining sufficient information. It was found that with a decimation of factor 3 (i.e. keeping 1 out of each 3 SSI steps) followed by an interpolation of the same factor (to compensate the missing phase offsets), the elasticity map (Figure 5B) and the mean shear moduli (Table 2, last line) remained close to those obtained with the full dataset. The maximum difference between the two datasets is equal to 0.1 kPa. Δ_{SSI} was therefore set to 1.5 ms (3×0.5 ms) for the monitoring experiment.

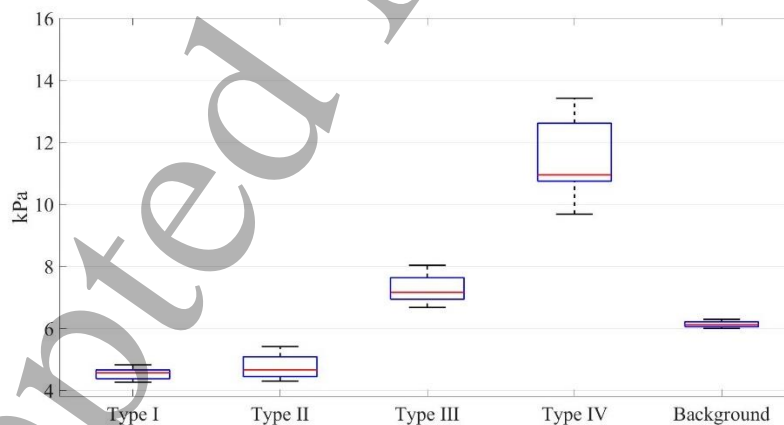


Figure 6 – Boxplot of the shear moduli obtained with the MR-SSI process in the CIRS phantom (8 datasets)

Monitoring of HIFU heating in a gel phantom

Figure 7 presents the SSI wavefronts before and at the end of the HIFU heating. It can be noticed that at the end of the heating, the wavefront on the right side did not go as far as it had gone before

the heating (red vertical line in Figure 7) suggesting that planar shear waves slow down over time (Video 2 – supplementary material). This decrease of the shear wave velocity was expected because the gel softens during heating. This is consistent with the elasticity maps obtained with MR-SSI where the shear modulus is found to decrease on the right side, at the same position where the temperature increased during the HIFU heating (Video 3 – supplementary material).

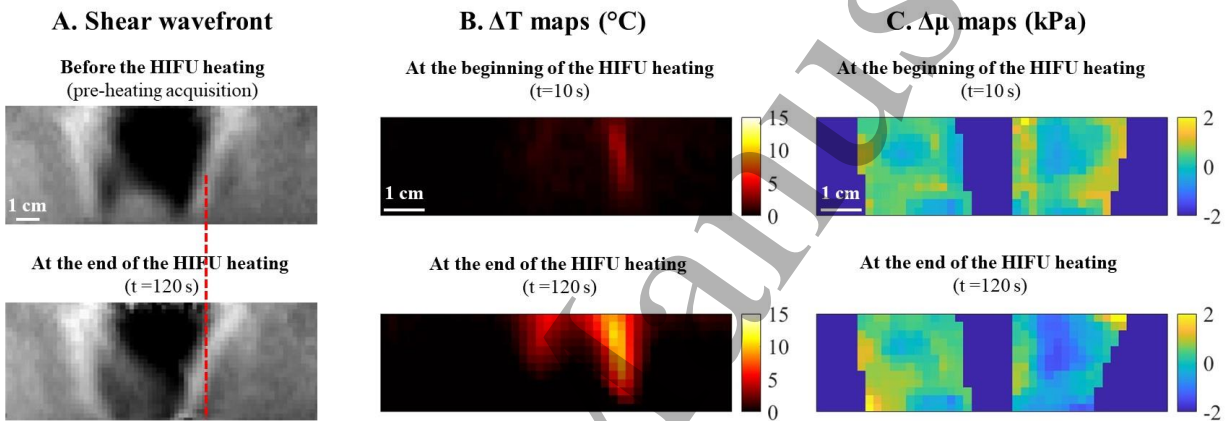


Figure 7 – Shear wavefront, thermometry and elastography maps before and after the HIFU heating. (A) The red dotted line indicates the inner edge of the right-side wavefront at the beginning of the heating.

At $t = 120$ s, the wavefront on the right side is slowed down, suggesting local gel softening as a result of the HIFU heating. Temperature increase and shear modulus decrease occurring on the heating side can be seen on ΔT (B) and $\Delta\mu$ (C) maps.

Figure 8 presents elasticity and temperature changes averaged within a 5×3 -pixel ROI selected around the heating focus on the thermometry map. As expected, temperature and shear modulus remained nearly constant during the whole experiment on the left side (symmetrical ROI with regards to the SSI push line). As for the right side, the temperature increase ΔT reached 9°C while the shear modulus μ decreased from $2.9 \text{ kPa} \pm 5\%$ to $1.8 \text{ kPa} \pm 17\%$ between the beginning and

the end of the heating. While the temperature started decreasing once the HIFU heating stopped, the shear modulus remained steady at 1.8 kPa during the following 120 s.

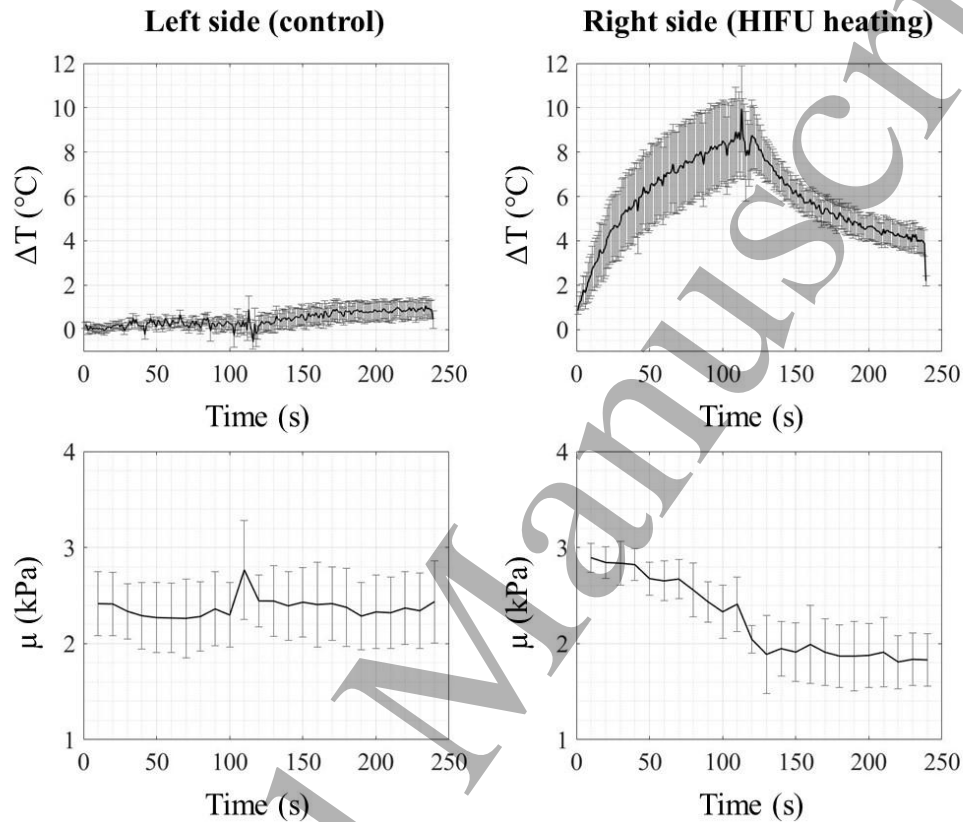


Figure 8 – Changes in temperature and shear modulus averaged within a 5×3 -pixel ROI on the left side (control) and on the right side (HIFU heating) of the SSI push line. Error bars represent the standard deviation of temperature over the selected ROI. On the right-side, heating the gelatin results in significant softening, whereas no significant changes are observed on the left side (control).

Discussion

In this paper, an MRI protocol and a reconstruction process have been proposed for the MR-SSI elastography method for the monitoring of elasticity and temperature changes during HIFU ablation. The proposed reconstruction process relies on two main steps: (i) shear wavefront tracking using an edge detection approach and (ii) shear wave velocity mapping based on a time-of-flight method. This elastography technique was first validated on a calibrated phantom where inclusions with different shear moduli could be mapped with high repeatability. Second, the HIFU heating experiment emphasized the ability to map in real time the irreversible elasticity changes caused by HIFU in a gel phantom. The shear modulus decrease at the heating focus was monitored on elasticity maps refreshed every 10 s. The shear modulus remained steady during the whole cooling down phase, which emphasizes the relevance of mechanical properties as biomarkers of thermal damage.

As presented previously, our MR-SSI elastogram reconstruction process is based on a time-of-flight computation, a principle also proposed by McCracken *et al* and Liu *et al* (McCracken *et al* 2005, Liu *et al* 2017) for shear wave velocity estimation. However, instead of extracting the maximum peak on the wave profiles, we chose to detect both the inner and outer edges of the wavefront. By doing so, the information is duplicated and 2 distinct elastograms can be obtained which allows averaging the shear modulus on the area where both maps overlap and so, reduce the number of potential outliers in this region.

Several limitations of the proposed method need to be highlighted. First, elasticity estimation may be subject to significant averaging effects. This is mainly explained by the fact that spatial resolution of the SSI elastograms is entirely dependent on the temporal resolution of SSI images.

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3 In other words, the more SSI steps are acquired, the better the resolution of the computed elasticity
4 maps is. However, the number of SSI steps should be maintained reasonably low to ensure a high
5 temporal update rate of elasticity. Thus, it is necessary to find an acceptable compromise between
6 temporal and spatial resolutions of the estimated elastograms. In addition, operations such as
7 interpolation and smoothing have supplementary averaging effects on the final elasticity values.
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10 Second, several phenomena may limit the spatial coverage and the accuracy of the obtained
11 elastograms. As for spatial coverage, elasticity cannot be mapped neither near the SSI push line
12 because of shear wave near-field effects, nor too far from this same line because of wave
13 attenuation, which makes the wavefront detection more difficult. As for accuracy, guided wave
14 phenomena can lead to inaccuracy of shear wave velocity estimates at interfaces between media
15 with different mechanical properties. These limitations could justify the differences between the
16 MR-SSI results and the values provided in the CIRS datasheet. Indeed, the previously mentioned
17 averaging effects may partially explain why the shear moduli obtained for the inclusions softer
18 than the background were higher than expected, and inversely for the rigid inclusions. It is also
19 important to note that elasticity values reported by the manufacturer have been obtained by
20 mechanical indentation, a static mechanical testing method that could also account for some of the
21 observed differences. All these reasons may explain the better consistency between MR-SSI and
22 conventional MRE as they are both dynamic elastography approaches, based on shear wave
23 propagation. This was further highlighted by the repeatability study performed on the CIRS which
24 shows that MR-SSI and MRE results are in good agreement. Although its accuracy needs to be
25 further evaluated, particularly in heterogeneous media for the aforementioned reasons, MR-SSI
26 was shown to be a repeatable method providing a quantitative biomarker of structural damage
27 induced by heating.
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3 A key parameter of the SSI method is the temporal step Δt_{SSI} that must be determined with
4 attention. Indeed, the higher the rigidity of the tissue, the smaller the Δt_{SSI} should be to ensure a
5 sufficient temporal resolution within an SSI cycle. Therefore, preliminary acquisitions are required
6 to obtain an estimation of the expected tissue elasticity, since this will influence the value of the
7 SSI step Δt_{SSI} that will be chosen. In practice, a pre-ablation MR-SSI acquisition may be performed
8 so the user can estimate the range and number of SSI phase offsets that will provide adequate
9 spatial coverage of the propagating wave for the following monitoring acquisitions. The results
10 presented in this paper are limited since they do not include in vitro or in vivo tissues. Therefore,
11 further investigation is necessary to find the optimal range of SSI phase offsets as well as the Δt_{SSI}
12 step as they will depend on the elasticity of the tissue as well as its heterogeneities.
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17 In vitro or in vivo tissue experiments would allow reaching much higher temperatures such as
18 those used clinically during HIFU ablations. Gelatin is not a suitable model for high temperature
19 experiments as it melts, thus preventing the generation and propagation of shear waves. In practice,
20 higher temperatures can be achieved by increasing the acoustic power of the HIFU heating push
21 (which was set at only 40% of the maximal power of the transducer for the heating experiment on
22 gelatin) and/or its duty cycle (set to 50% here). It should be noted that higher duty cycles can be
23 achieved for the HIFU heating push. This duty cycle could be further increased through the use of
24 a real-time industrial version of the communication software between the HIFU generator and the
25 MRI console, or through the use of longer TR, though at the expense of the SSI temporal
26 resolution.
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31 Several approaches have already been proposed for MR-guided HIFU monitoring using the
32 mechanical properties of the tissue. As opposed to MRE, MR-SSI does not require an additional
33 mechanical exciter since the ARF is used to generate the pushes. Nonetheless, a potential risk of
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3 this method could be to induce some thermal effects along the SSI push line. This is mainly avoided
4 by the short duration as well as the low duty cycle of the pushes. The mild heating observed along
5 the SSI push line in the second experiment (+6 °C maximum) does not entail any change in the
6 phantom structure, and it is deemed to be lower in a perfused organ. Furthermore, this heating
7 along the SSI push line was mainly observed in experiments where a HIFU heating was performed
8 simultaneously with the MR-SSI elastography measurements. No significant heating was observed
9 in MR-SSI experiments without HIFU heating phase. This could be explained by the fact that the
10 HIFU wave delivers a certain amount of energy along the whole ultrasound cone. Although this
11 energy is mostly absorbed at the focus, the summation of energy on the areas where several
12 ultrasound cones are superimposed may induce a mild temperature increase (especially in the
13 presence of a long HIFU heating push, and in the absence of perfusion). This is visible in Video
14 3, as mild heating occurs along the SSI push line only during the HIFU heating phase, followed
15 by a plateau during the cooling down phase despite the fact that the SSI pushes were still generated.
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17 When comparing to other ARF-based approaches, MR-SSI is distinguished by its ability to provide
18 an extended spatial mapping of elasticity whereas other methods such as MR-ARFI estimate a
19 single value at the focus (Vappou *et al* 2018). The method proposed by Hofstetter *et al* (Hofstetter
20 *et al* 2019) also allows for spatial elasticity mapping. It relies on the generation of multiple pushes
21 in a plane perpendicular to the ultrasound propagation axis, allowing reconstructing elasticity maps
22 every 12 seconds. Our method instead relies on the generation of quasi-cylindrical supersonic
23 shear waves, which, in theory, are more contrasted and less attenuated than spherical waves. This
24 may explain the larger spatial extent of our MR-SSI method in a coronal slice (a circular field of
25 view of about 13 cm in diameter on the CIRS phantom) compared to the method of Hofstetter *et*
26 *al* where elasticity has been measured on a circular field of view of 5 cm in diameter. Another
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3 relevant point of comparison is the applicability of our MR-SSI method in different imaging planes
4 as the composite shear wave used by Hofstetter *et al* makes the method only applicable in coronal
5 slices. Nonetheless, this latter was also demonstrated in 3D providing elasticity measurements over
6 several slices with an acquisition time of 144 seconds. Finally, both approaches allow mapping
7 elasticity in 2D with similar refresh rates compatible with real-time monitoring.
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12 Being equivalently applicable in different imaging planes adds a supplementary degree of freedom
13 to MR-SSI. A transverse imaging slice has been chosen for the HIFU heating experiment as the
14 HIFU focus has an ellipsoidal shape of about 1 cm in length and 1-2 mm in diameter. Thus,
15 mapping the elasticity in a transverse plane is much more relevant in order to detect the variations
16 along the length of the ellipsoidal lesion. Therefore, the possibility to choose the imaging plane
17 according to the shape and size of the expected tissue heterogeneities is an advantage of the
18 proposed method.
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32 Although theoretically, the MR-SSI method can be used with non-EPI sequences, the need for
33 high temporal resolution virtually imposes the use of single-shot EPI acquisitions, with all related
34 artifacts and limitations. Single-shot EPI with longer TR also offers optimal duty cycle balance
35 between the expected low duty cycle at the SSI push line (limited heating), and the high duty cycle
36 necessary at the HIFU ablation spot.
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Conclusion

The proposed MR-SSI method allows simultaneous temperature and elasticity mapping. This method was first evaluated on a heterogeneous phantom and was then used for the monitoring of HIFU ablations, with MR-SSI elastograms updated every 10 seconds and relative temperature maps updated every second. In the case of a gelatin phantom, the shear modulus was found to decrease during the HIFU heating, thus reflecting that MR-SSI can be a promising technique to assess the structural changes of tissue during a HIFU treatment.

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