PONTIFICIA UNIVERSIDAD CATOLICA DE CHILE
SCHOOL OF ENGINEERING

IMPLEMENTATION OF A MAGNETIC RESONANCE ELASTOGRAPHY SYSTEM

ALLAN ANDRÉS CID OLIVARES

Thesis submitted to the Office of Research and Graduate Studies
in partial fulfillment of the requirements for the degree of
Master of Science in Engineering

Advisor:
PABLO IRARRÁZAVAL MENA
SERGIO URIBE ARANCIBIA

Santiago de Chile, July 2012

PONTIFICIA UNIVERSIDAD CATOLICA DE CHILE
SCHOOL OF ENGINEERING

IMPLEMENTATION OF A MAGNETIC RESONANCE ELASTOGRAPHY SYSTEM

ALLAN ANDRÉS CID OLIVARES

Members of the Committee:
PABLO IRARRÁZAVAL MENA
SERGIO URIBE ARANCIBIA
CRISTIÁN TEJOS NUÑEZ
MARCELO ANDIA KOHNENKAMPF
DOMINGO MERY QUIROZ

Thesis submitted to the Office of Research and Graduate Studies
in partial fulfillment of the requirements for the degree of
Master of Science in Engineering

Santiago de Chile, July 2012

A mi familia.

ACKNOWLEDGEMENTS

I would like to express my gratitude to my academical advisors Dr. Pablo Irarrázaval and Dr. Sergio Uribe, who guided me through the way of knowledge, research and science. I appreciate their expertise, understanding and patience. I would also like to thank the other members of my committee, Dr. Domingo Mery, Dr. Cristián Tejos and Dr. Marcelo Andia, for their assistance on the development of this research project.

I cannot forget to acknowledge my colleagues and friends Iván Chaparro, Rafael Muñoz, Juan Muñoz, Roberto Pérez, Carlos Milovic and Cristóbal Arrieta for their motivation and encouragement on this process. I would also like to thank my colleagues and professors from the Ecole Centrale Marseille.

A very special thank goes to my mother Marcela, my brother Samir and all my family for the support, values and courage they provided me through all my life. I would also like to thank Nadjet Merabet, my girlfriend, without her love, encouragement and help, I would not have finished this thesis.

In conclusion, I express my gratitude to the Anillo ACT79 project and the Biomedical Imaging Center (CIB). This research would not have been possible without their support.

TABLE OF CONTENTS

LIST OF FIGURES

ABSTRACT

Most chronic liver diseases have in common a progressive hepatic fibrosis, and the degree and rate of progression of fibrosis are important prognostic factors in patients with chronic liver disease. The gold standard for staging of fibrosis is liver biopsy. However, this technique is invasive and dangerous. Several classic non-invasive techniques have been tested for possible use in hepatic fibrosis diagnose. Nevertheless, all these techniques are qualitative and may not be enough to measure the small differences that occur at detect early stages of the disease. It has been discovered that liver elasticity changes proportionally to the fibrosis stage of the liver. Non-invasive methods such as Ultrasound Elastography (USE) and Magnetic Resonance Elastography (MRE) have been introduced to measure elasticity in vivo. These methods produce elasticity maps (called elastograms) which can be used to assess the liver condition. In this work we present a low-cost pneumatic MRE system which is capable of measuring elasticity in phantoms and liver, based on the work developed at Mayo Clinic. The system uses a special hardware called Elastograph which is connected to the MRI scanner. The Elastograph mechanically stimulates the tissues of interest while obtaining dynamic images using the scanner. These images contain displacement information of the tissues, which is used to assess its elasticity. MRE systems available on the market are expensive. This is because of three main factors: (1) most developments have been patented, (2) only a few companies offer them and (3) their hardware and software design processes are complex. We built our system from scratch, which reduced up to 10 times its cost. Its low-cost allows medical and research institutions to build and set their own experimental MRE equipment and software.

Keywords: MRE, MRI, Elastography, Elastogram, Wave image, Palpation, Mechanical properties, Shear modulus.

RESUMEN

La mayoría de las enfermedades hepáticas tienen como factor común el desarrollo de una fibrosis hepática progresiva, y el grado y la tasa de progresión de la fibrosis son factores importantes de pronóstico para pacientes con un trastorno hepático crónico. La técnica estándar para medir el grado de fibrosis es la biopsia. Sin embargo, esta técnica es invasiva y peligrosa. Varias técnicas no-invasivas clásicas se han probado para un posible uso en el diagnóstico de fibrosis hepática. No obstante, estas técnicas son cualitativas y pueden no ser suficientemente sensibles para medir las pequeñas diferencias que ocurren en etapas tempranas de esta enfermedad. Se ha descubierto que la elasticidad del hígado cambia proporcionalmente a su grado de fibrosis. Métodos no-invasivos como la Elastografía por Ultrasonido (Ultrasound Elastography en inglés, o USE) y la Elastografía por Resonancia Magnética (MAgnetic Resonance Elastography en inglés, o MRE) han sido desarrollados para medir la elasticidad de tejidos en vivo. Estos métodos producen mapas de elasticidad (también llamados Elastogramas), los que pueden ser usados para estimar el estado de salud del hígado. En este trabajo presentamos un sistema de MRE de bajo costo que permite medir la elasticidad de tejidos en phantoms y el hígado, basados en el trabajo realizado en la Clínica Mayo. Este sistema utiliza un hardware especial llamado Elastógrafo, y un scanner de MRI. Este Elastógrafo estimula mecánicamente el tejido de interés para obtener imágenes dinámicas utilizando el escanner. Estas imágenes contienen información de desplazamiento de los tejidos, la que es utilizada para estimar su elasticidad. Los sistemas de MRE actualmente disponibles en el mercado son caros. Esto sucede debido a tres factores: (1) la mayor parte de los desarrollos se encuentran patentados, (2) son escazos en el mercado y (3) los procesos de desarrollo del hardware y software son complejos. Nosotros construimos nuestro sistema desde cero, lo que redujo en un orden de 10 veces su costo. Su bajo costo permite a instituciones médicas y de investigación a construir su propio equipamiento experimental de MRE.

Palabras Clave: MRE, MRI, Elastografía, Elastograma, Imagen de ondas, Palpación, Propiedades mecánicas, Módulo de elasticidad.

1. INTRODUCTION

Most chronic liver diseases have in common a progressive hepatic fibrosis (Yin et al., 2008) and the degree and rate of progression of liver fibrosis are important prognostic factors in patients with chronic liver disease (Talwalkar et al., 2008). The standard for staging of fibrosis is liver biopsy (Yin et al., 2008). However, as an invasive procedure, it has inherit inconvenients as sampling error, cost and pain, among others (Rouvière et al., 2006; Huwart et al., 2008; Talwalkar et al., 2008).

Several non-invasive imaging techniques, such as standard Magnetic Resonance Imaging (MRI), Computed Tomography (CT) and Ultrasound (US), have been tested for possible use in hepatic fibrosis diagnose (Mariappan, Glaser, & Ehman, 2010). However, all these techniques are qualitative and may not be enough to measure the small differences that occur at detect early stages of the disease (Talwalkar et al., 2008). A significant need exists for developing a safe, accurate and non-invasive technique for assessing the degree of fibrosis (Yin et al., 2008).

It has been demonstrated that liver elasticity changes as product of its health, particularly in presence of fibrosis. It means that the elasticity of the tissue changes proportionally to the fibrosis stage. This change is gradually done, so elasticity measures can give us a clue about liver health in earlier stages of the disease. (Muthupillai et al., 1995)

Methods such as Ultrasound Elastography (USE) and Magnetic Resonance Elastography (MRE) have been introduced to assess elasticity in vivo (Mariappan et al., 2010). Both methods produce elasticity maps (called elastograms) by applying a stress to a tissue and measuring its displacement values using US or MRI. While MRE measures the tissues shear modulus, USE does the same with bulk modulus. It has been proven that shear modulus can assess a wider range of tissue elasticity compared to bulk modulus (Mariappan et al., 2010). A comparison among the different imaging contrast methods is shown on figure 1.1. As MRE can quantitatively assess elasticity, it has been used in several clinical fields, such as tumor and disease detection in places as abdomen, brain, breast, heart, lungs, muscles, prostate and thyroid glands, among others (Mariappan et al., 2010). The way MRE works

is: (1) A mechanical actuator applies a cyclic stress (usually a monochromatic wave) on the patient's abdomen. (2) The induced strain is measured using MRI for different wave positions or phases creating a wave image. (3) An elastogram is reconstructed using the displacement information found in the offsets.

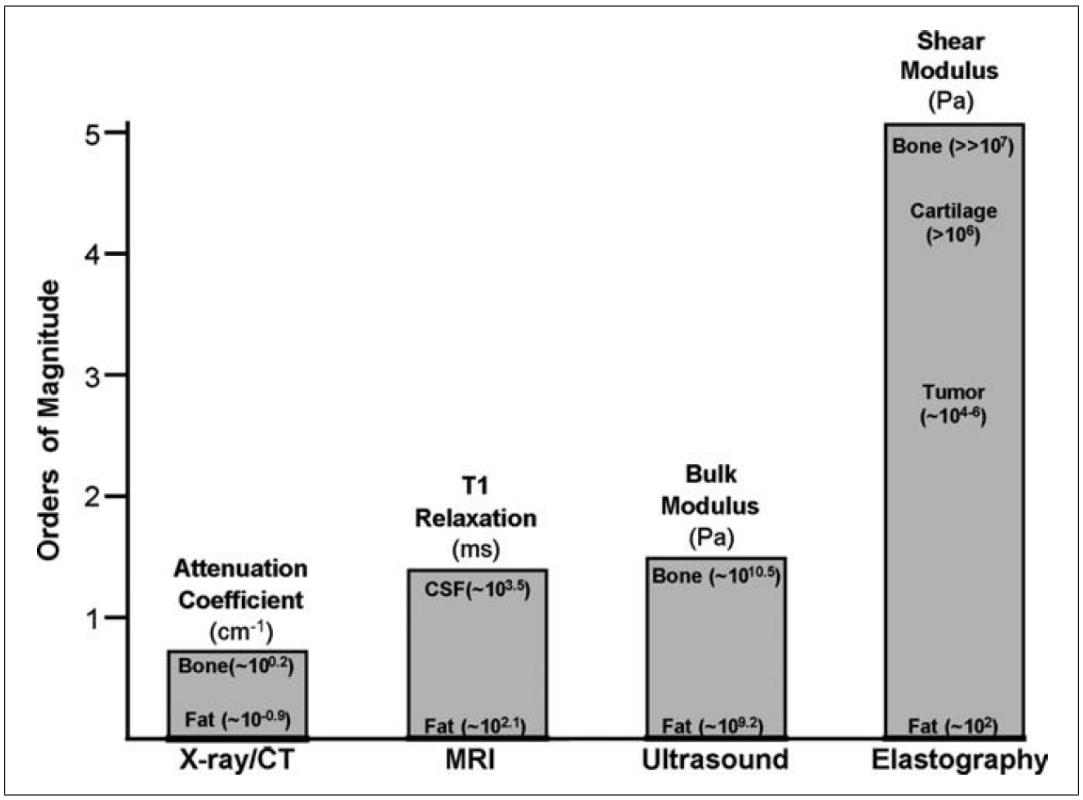

FIGURE 1.1. Comparison between different imaging modalities and the spectrum of contrast mechanisms utilized by them. The shear modulus has the higher contrast sensitivity for over 5 orders of magnitude among various physiological states of normal and pathological tissues. (Mariappan et al., 2010)

Several MRE systems and techniques have been developed (Tse et al., 2009; Mariappan et al., 2010). Until now, most of these systems are expensive and unreachable for most medical and research institutions. This is due to three important factors: (1) most hardware and software developments have been patented; (2) the low availability of these devices in the market; and (3) the complexity of the software development (Tse et al., 2009).

In this work we present a pneumatic MRE system, based on Talwalkar et al. (2008) work, which is capable of measuring elasticity in phantoms and liver. It consists of a loudspeaker for vibration generation, which is located outside the scanner room, a pneumatic tube, which receives and transports the vibration into the room, and a drum-like passive driver placed on the patient body. The wave generation is synchronized using a TTL pulse signal coming from the scanner, which is necessary to capture different offsets. The signal comes from a wave generator and it is boosted by a power amplifier. To register displacement information, we programmed an MRE sequence based on a Gradient Echo (GE) Phase Contrast (PC) sequence. To build the elastograms we used a reconstruction software developed by the Mayo Clinic (Mayo Clinic, Rochester, MN, USA).

Unlike commercial system available, its low cost allows medical and research institution to build and set their own experimental MRE equipment and software.

2. THEORY

Studies have proven that the condition of some human tissues is related to their elasticity (or stiffness). Measuring the elastic properties can be done by using three different approaches (Tse et al., 2009): (1) the static approach, (2) the quasi-static approach and (3) the dynamic approach.

The static approach compares the displacements that encounter different tissues while a constant force is applied. The quasi-static approach compares those displacements in a times series obtained by applying low motion frequencies around 1Hz. The dynamic approach compares local displacements by applying higher motion frequencies than the quasi-static approach.

Although the three approaches can be used to calculate elasticity, the dynamic approach is more practical in the clinical environment and gives a better resolution (because its frequency range can be measured using an MRI scanner). Most of MRE research has been done based on this approach (Mariappan et al., 2010).

2.1. Elasticity

The elasticity is the physical property of a material to return to its original shape after a deformation induced by stress. Physically, elasticity is described by Hooke's law:

$$\sigma_{ij} = C_{ijkl} \cdot \epsilon_{kl} \tag{2.1}$$

where σ is the stress tensor, C is the elasticity tensor and ϵ is the strain tensor. In the general formulation, C is a 4^{th} order tensor and it has 81 elastic coefficients defined by σ and ϵ. Most of the research in the MRE field assume that liver tissue is an isotropic linear elastic material domain (Mariappan et al., 2010). Under this assumption we can reduce these 81 coefficients to just 2 (λ, μ) called the Lamé constants. The simplified expression of 2.1 is:

$$\sigma_{ij} = 2\mu\epsilon_{ij} + \lambda\epsilon_{ii}\delta_{ij} \tag{2.2}$$

where λ and μ are called the bulk and shear modulus respectively. It has been proven that the shear modulus is proportional to the stiffness of the material.

Other way to calculate elasticity is using the Local Estimated Frequency (LFE) approach (Manduca, Muthupillai, Rossman, Greenleaf, & Ehman, 1996). This works by estimating the local wavelength on the acquired wave images. As the local frequency is known, we can use the Oberst's formula to calculate the shear stiffness:

$$\mu = \nu^2 l^2 \rho \tag{2.3}$$

Where ν is the mechanical wave frequency, l is the local wavelength estimated, and ρ is the density of the medium (Muthupillai & Ehman, 1996).

Regardless of the used assumption, the problem is to measure the local displacements ϵ suffered by the protons within the liver while an stress σ is applied. Once we have measured them, we will be able to build an elastogram based on the mechanical approach which better fits the studied tissue (Mariappan et al., 2010).

2.2. MRE

MRE is a non-invasive technique developed to assess human tissue stiffness. As mentioned in section 2.1, we can calculate a material's stiffness by measuring displacements experimented by a particle while being mechanically excited.

In MRI, we can assess flow or motion using Motion Encoding Gradients (MEGs). An MEG is a waveform that encodes information about coherent motion into the phase ϕ of the MR signal. MEGs can encode information about velocity, acceleration, or higher derivatives of motion. If the type of motion is known (e.g. sinusoidal waveform), displacements can be calculated from the velocity information (Muthupillai et al., 1995).

In MRI, the accumulated phase for a spin isochromat is given by:

$$\phi(t) = \gamma \int_0^t \mathbf{G}_r(u)\mathbf{r}(u)du \tag{2.4}$$

where γ is the gyromagnetic constant, $\mathbf{G}_r(u)$ is a time dependent magnetic field gradient and $\mathbf{r}(u)$ is the position vector of the moving spin (Muthupillai et al., 1995).

In MRE, $\mathbf{r}$ is usually described by a sinusoidal mechanical waveform. If $\mathbf{G}_r(u)$ has the same frequency as $\mathbf{r}$ (e.g. a sinusoidal or rectangular pulse train), the phase is:

$$\phi_{sinusoidal}(t) = \frac{\gamma NT(\mathbf{G}_0 \cdot \boldsymbol{\xi}_0)}{2} \cos(\mathbf{k} \cdot \mathbf{r} + \theta) \tag{2.5}$$

$$\phi_{rectangular}(t) = \frac{2\gamma NT(\mathbf{G}_0 \cdot \boldsymbol{\xi}_0)}{\pi} \sin(\mathbf{k} \cdot \mathbf{r} + \theta) \tag{2.6}$$

where T is the period of the wave, N is the number of MEGs applied, $\mathbf{k}$ is the wave vector, θ is the phase between the MEGs waveform and the mechanical waveform, and $\mathbf{G}_0 \cdot \boldsymbol{\xi}_0$ represents the projection of the displacement $\boldsymbol{\xi}_0$ in the MEG direction (Muthupillai & Ehman, 1996). $\boldsymbol{\xi}_0$ is the displacement encoded if those vectors are parallel (measured un μm) (Muthupillai & Ehman, 1996).

The Maximum Displacement (or Motion) Encoded by the system (MENC) can be determined by:

$$MENC_{sinusoidal} = \frac{2\pi}{\gamma NT|\mathbf{G}_0|} \tag{2.7}$$

$$MENC_{rectangular} = \frac{\pi^2}{2\gamma NT|\mathbf{G}_0|} \tag{2.8}$$

and it is measured in μm.

2.3. Actuator

In MRE, mechanical waves are created by an external vibration driver. This driver must generate longitudinal or transverse waves and must be synchronized to the MR scanner. Furthermore, it has to be MRI compatible. A survey of MRE hardware design can be found in Tse et al. (2009). Tse classifies MRE hardware in three types, depending on the technology used to guarantee MR compatibility:

Type I: Actuation is generated far away from the scanner or outside the scanner room. Movement is then transmitted by a pneumatic tube or a long rod to a passive wave generator located near the magnetic isocenter.

Type II: Actuation is generated in the scanner room using electromagnetic coils by making use of the scanner's main field. This happens because a current passes through the coil and motion is induced according the Lorentz law for actuation.

Type III: Actuation is generated by an active wave generator near to the object of interest. This actuator must be MR compatible to prevent forces or torques induced by the magnetic field, such as piezoelectric materials.

From the list above, type I longitudinal pneumatic actuator fits better to hepatic MRE (Tse et al., 2009).

2.4. Sequence

MRE sequences have in common the use of MEGs to register the displacement generated by the traveling waves. In MRI, there are two sequence families (Spin-Echo (SE) and Gradient-Echo (GE) sequences) and both are suitable for the inclusion of MEGs (Muthupillai & Ehman, 1996; Asbach et al., 2008; Huwart et al., 2008).

While GE sequences are susceptible to chemical shift and field inhomogeneities, they are dramatically faster than SE sequences while maintaining diagnostic performance for assessing liver fibrosis (Weaver, Van Houten, Miga, Kennedy, & Paulsen, 2001).

2.5. Reconstruction Method

There are several MRE reconstruction methods (Manduca et al., 1996; Bishop, Samani, Sciarretta, & Plewes, 2000; Oliphant, Manduca, Ehman, & Greenleaf, 2001). The main idea for all the reconstruction algorithms is to measure the wavelength of the traveling vibrations. Once measured, we can use this information to build an elasticity map. The methods can

be classified by its underlying mechanical model or the technique for generating the elasticity map. While mechanical approaches variate between elastic, viscoelastic and poroelastic; the elasticity map generating techniques variate from texture analysis (Manduca et al., 1996) to the inversion of mechanical equations (Oliphant et al., 2001).

3. MATERIALS AND METHODS

3.1. MRE Hardware

The vibration actuator we used is a type I longitudinal pneumatic passive actuator. This design is based in four fundamental components which are connected in cascade: (1) a signal generator, (2) a power amplifier, (3) a loudspeaker and (4) a passive driver (Ehman, Rossman, Hulshizer, & Dresner, 2005).

The system works as follows: the electric signal generated is amplified and then transduced by the loudspeaker (active driver). The vibration is collected using a polycarbonate plate (which completely covers the loudspeaker) with a hole in the middle. Then, it is transmitted using a pneumatic tube connected to a drum-like driver (passive driver) which is placed on the patient's body.

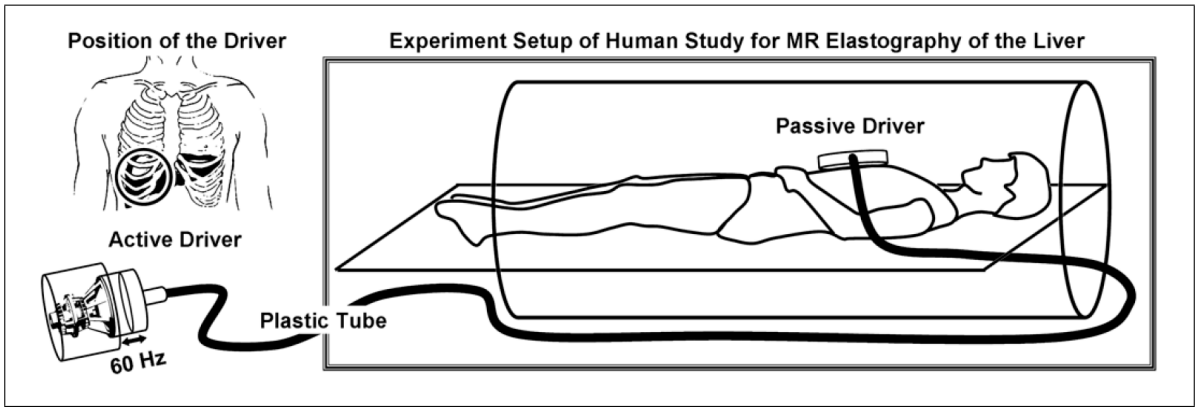

FIGURE 3.1. System for applying shear waves for MRE of the liver. The acoustic waves (60Hz) applied are generated by a loudspeaker located away from the MRI cage, and transmitted via a flexible tube to a drum-like passive driver. In the left is shown the correct position of the passive driver with respect to the liver. (Yin et al., 2008)

Triggering Device

Signal generator synchronization was achieved using the Philips Functional Brain Imaging Box (FBI Box) (Philips Healthcare, Best, The Netherlands) which generates trigger pulses (TTL pulses of 4 V of around 5 microseconds) from an optical signal coming from the scanner.

FIGURE 3.2. Triggering device for the MRE system. (Philips Healthcare, Best, The Netherlands)

Signal Generator

The signal generator we used is a 100 MHz Tektronix AFG3101 (Tektronix Inc., Beaverton, OR, USA) with a 50Ω output which can be synced by a TTL pulse. This study uses a 60Hz sine wave signal for human volunteers and a 125Hz and 250Hz sine wave signal for phantoms, with a voltage level of 5V peak-to-peak for all the experiments.

FIGURE 3.3. Signal generator to create the mechanical wave. (Tektronix Inc., Beaverton, OR, USA)

Power Amplifier

We used a 600W Crest CPX 900 Power Amplifier (Crest Audio Inc., Meridian, MS, USA) with an input impedance of 50Ω and an output impedance of 8Ω using the bridge mono configuration to reach 600W.

FIGURE 3.4. 600W power amplifier. (Crest Audio Inc., Meridian, MS, USA)

Loudspeaker

As active driver, we used a 18 inches Peavey Black Widow loudspeaker (Peavey Electronics Corporation, Meridian, MS, USA), with an input impedance of 8Ω and 600W output.

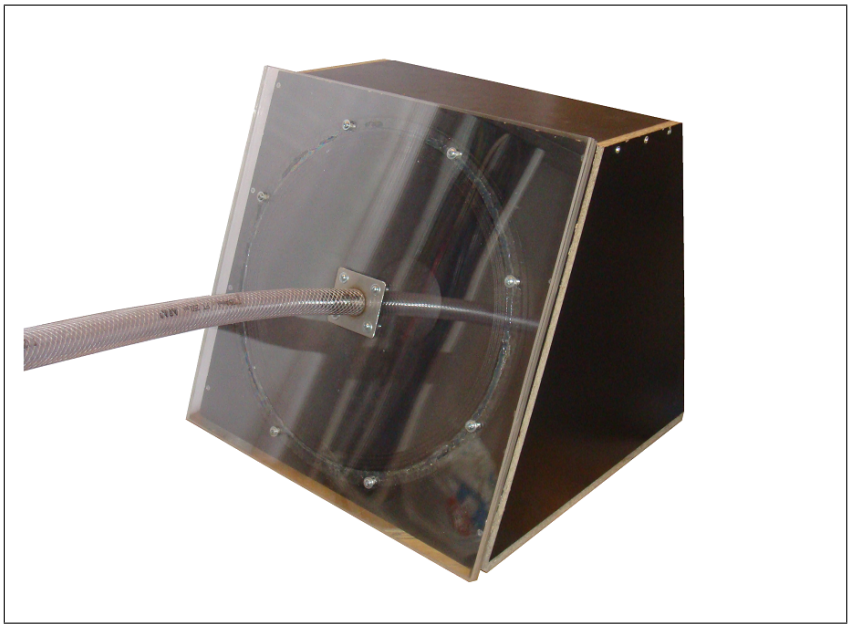

FIGURE 3.5. Loudspeaker used as an active driver to generate motion. (Peavey Electronics Corporation, Meridian, MS, USA)

Driver Design

The driver system is connected to the loudspeaker through an 12mm thick acrylic plaque which covers the entire loudspeaker and the joints were sealed using silicone. The plaque is connected to a 1-inch diameter 2-mm thick rubber tube using a metallic joint. The tube goes to the scanner room through a waveguide and it is connected to a drum-like actuator which is placed over the patient's body. The passive drum-like actuator consists of an acrylic body with a polycarbonate membrane. The body is comprised of a cylindrical shaped enclosure connected to the end of the tube. The size of this enclosure is 2mm thick, and varies from 7cm to 15cm in diameter, depending on the particular clinical application. The membrane is made of a 0.2mm thick polycarbonate sheet. The tube and the actuator are made from MR compatible materials.

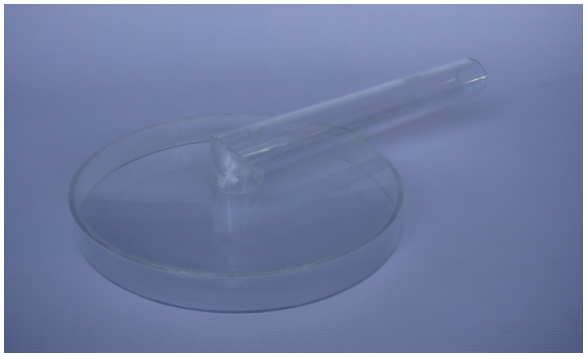

FIGURE 3.6. Drum-like passive driver which generates motion across the tissues.

3.2. MRE Sequence

We have programmed an MRE sequence based on a PC GE sequence using the Goal-C Pulse Programming Environment for Philips MR scanners (Figure 3.8). A diagram of the sequence can be seen in Figure 3.7.

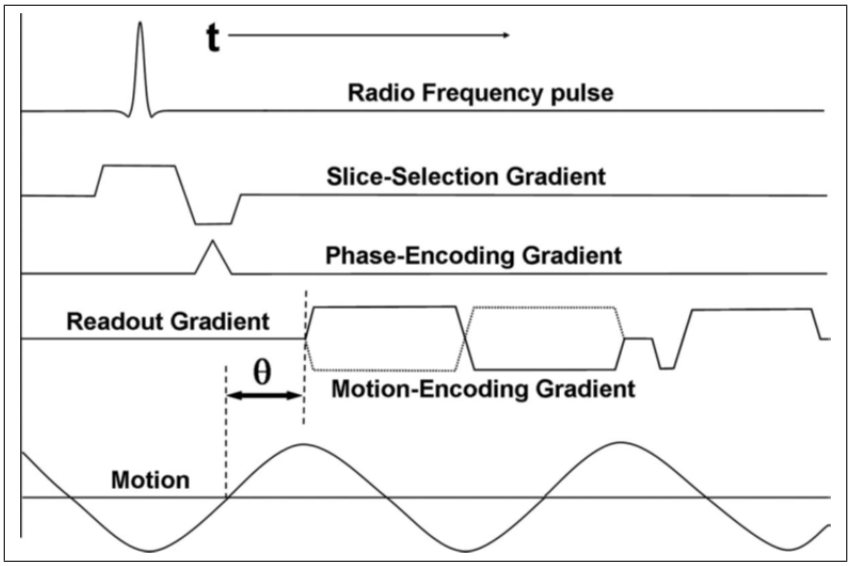

FIGURE 3.7. MRE sequence diagram. (Mariappan et al., 2010)

We made the following changes:

(i) The sequences was modified to input MRE parameters from both scanner menus (ScanList and ExamCards).
(ii) Trapezoidal shaped MEGs were put between the RF excitation and the beginning of the readout gradient on all three axes to capture motion information.
(iii) A trigger signal was added to synchronize wave generation to the scanner MEGs to get different phase images.

The MRE input parameters are:

(i) MEGnum: Number of motion encoding gradients
(ii) MEGstr: Motion encoding gradients amplitude
(iii) MEGfreq: Motion encoding gradients frequency

All these parameters are used to calculate the MENC mentioned in equations (2.7) and (2.8).

3.3. Acquisitions

Four normal individuals were recruited prospectively on a volunteer basis. Inclusion criteria were (1) age $\geq$ 18 years, and (2) no prior history of chronic liver disease. None of the normal volunteers had a previous diagnose of steatohepatitis or any other kind of liver disease. We used these criteria to compare our results to the state-of-art (Yin et al., 2008).

The MRE sequence with cartesian trajectories was performed to register the displacement information into the phase image. All the images were obtained using a Philips Intera 1.5T scanner (Philips Healthcare, Best, The Netherlands). This scanner has a maximum gradient strength of 21mT/m.

A cardiac body coil was used, and 9 acquisitions (with a 40^{o} phase between each one) were performed with the following parameters: MEGs frequency = 60Hz, 1 MEG pair, MEG strength = 10,5mT/m, matrix size 256x256, TR =50 ms, TE=19.4 ms, Flip Angle (FA) = 15^{o}, slice thickness = 10mm, Field of View (FOV) = 450mm, breathhold = 10 seconds per acquisition. While TR was chosen to be a multiple of the MEG's period, TE was chosen to be the minimum. The passive actuator was placed as indicated in figure 3.1.

3.4. Reconstruction

To calculate elasticity from the acquired images, we use the MRE/Wave software (Mayo Clinic, Rochester, MN, USA). This reconstruction software assumes that the liver behaves as an isotropic linear elastic material.

MRE/Wave uses a Local Frequency Estimate (LFE) algorithm (Manduca et al., 1996) to invert the acquired phase difference wave data (wave images) into an Elastogram to provide an estimate of the stiffness (Muthupillai et al., 1995).

The wave images were calculated by subtracting the negative polarity image from the positive polarity image independently for each phase image directly on the scanner. We used magnitude images to mask the wave images.

The inversion was performed following the mechanical model explained on equation 2.3.

The parameters needed to perform inversion are FOV, MEG frequency, number of MEGs, MEG amplitude and number of phase offsets acquired. A Gaussian Band-Pass Filter was used to reduce noise.

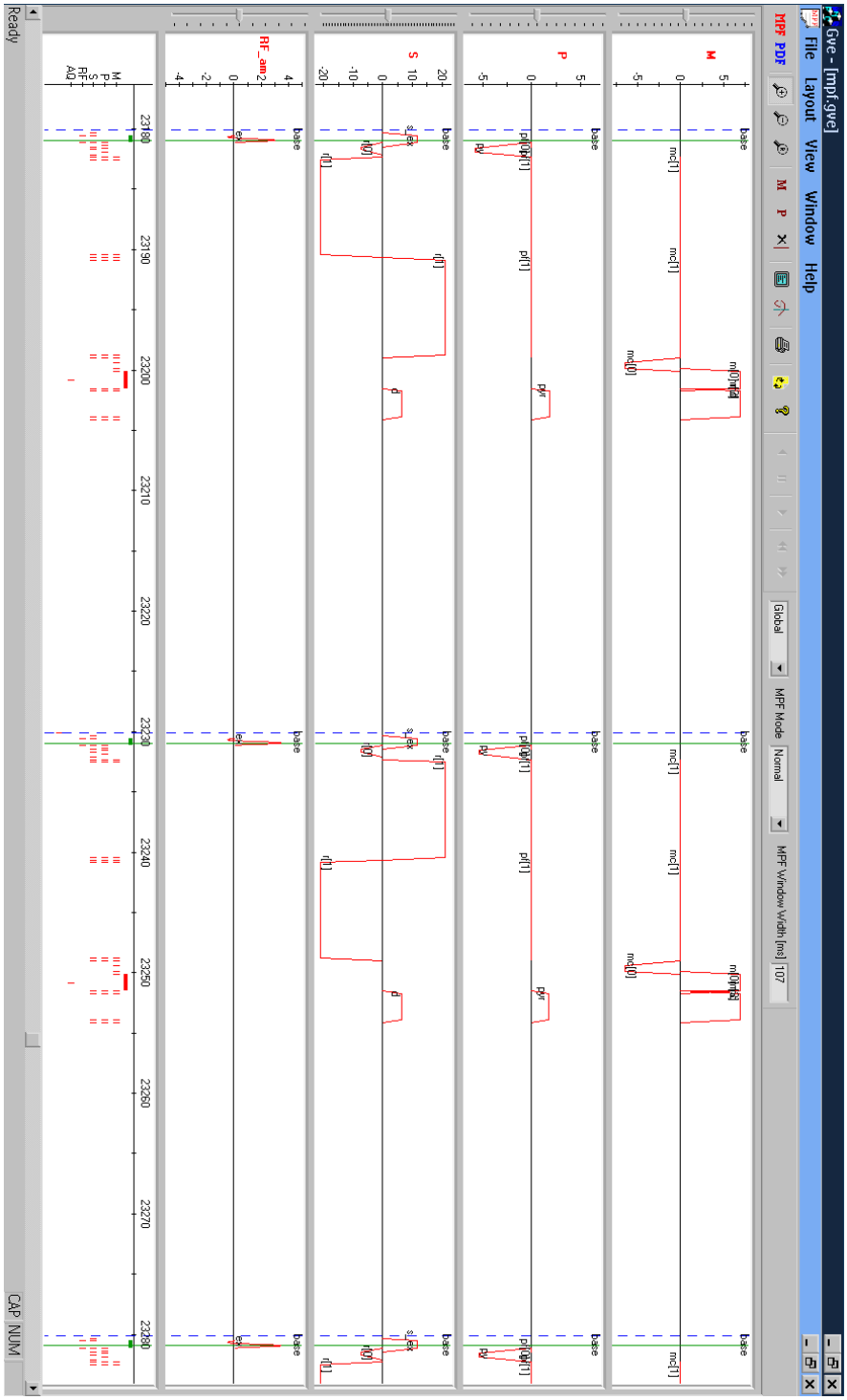

FIGURE 3.8. MRE sequence programmed using the Goal-C Pulse Programming Environment for Philips MR scanners. A MEG can be observed in the slice select gradient.

4. RESULTS AND DISCUSSION

4.1. Image Obtention Performance

The MRE driver system generated shear waves that could be readily imaged throughout the visualized upper abdomen in normal volunteers. Figures 4.1 and 4.2 demonstrates shear wave images and elastograms obtained from four normal volunteers. No problems were encountered in applying the MR Elastography technique with the surface driver in fat volunteers, which could potentially prevent adequate generation of shear waves in the liver.

4.2. Wave Images

The wave seems to be homogeneous in the whole liver extension area, but it disappears in the spleen area. As we can see, these waves seem to have a similar wavelength in all the volunteers, so we can hypothesize that the elasticity will be comparable for them.

4.3. Elastograms

Elasticity images for four volunteers are shown on 4.2. These images show elasticity measures calculated from a series of wave images. For the four volunteers, we got the following stiffness values in the liver area: 2.6 ± 0.8 kPa, 2.7 ± 1.0 kPa, 2.8 ± 1.1 kPa and 2.5 ± 1.1 kPa. As the wave images tells us about similar wavelength on the liver area, elastograms present also comparable elasticity measures. Also, all the individuals seem to be healthy as none of their mean stiffness is over the stiffness upper limit of normal (Taouli, Ehman, & Reeder, 2009).

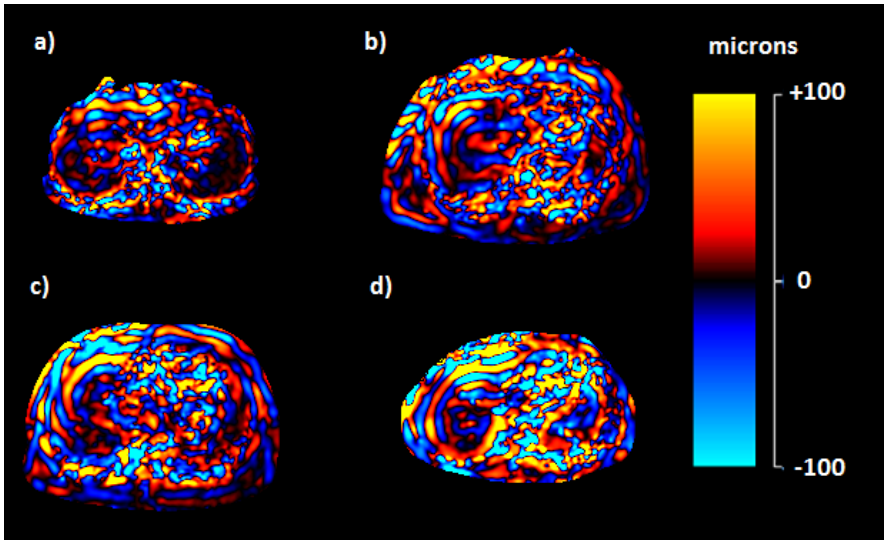

FIGURE 4.1. Wave images for the 4 healthy volunteers. The traveling wave can be observed in the liver extension area. For all the 4 volunteers the wavelength is similar.

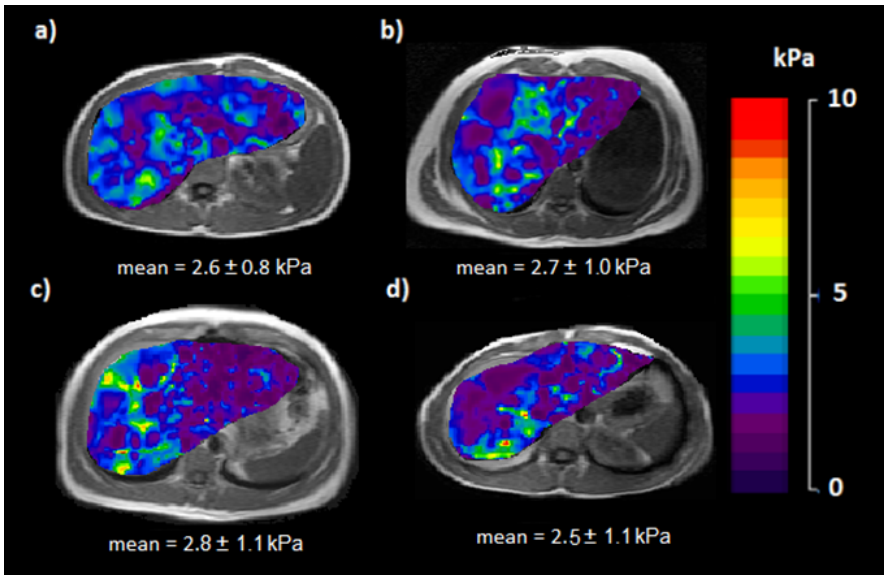

FIGURE 4.2. Elastograms for the 4 healthy volunteers. The measures obtained are comparable to state-of-art results (Yin et al., 2008) and confirm the health of the volunteers.

4.4. Comparison

To validate our system, we compared our results against the state-of-art results (Yin et al., 2008). From this article we took 3 images of a healthy volunteer: (1) an anatomic image, (2) a wave image and (3) an elastogram and we put our results in parallel. The comparison is shown in figure 4.3. We took the anatomic image to identify and manually segment the liver.

For the wave image, there is good correspondence between our results and those of the Mayo Clinic: wave shape, displacement order and wavelength are alike in the liver.

By comparing the elastograms, we can see a resemblance between the elasticity measures on the liver. Out of the liver the measures are dissimilar because of the attenuation of the wave. The mean liver stiffness value in the liver segmented image for the volunteers is 2.50 ± 0.98 kPa, which is close to the mean stiffness value found on the bibliography for normal volunteers: 2.20 kPa (For a group of patients with cirrhotic liver the mean liver stiffness value is 8.90 kPa) (Mariappan et al., 2010).

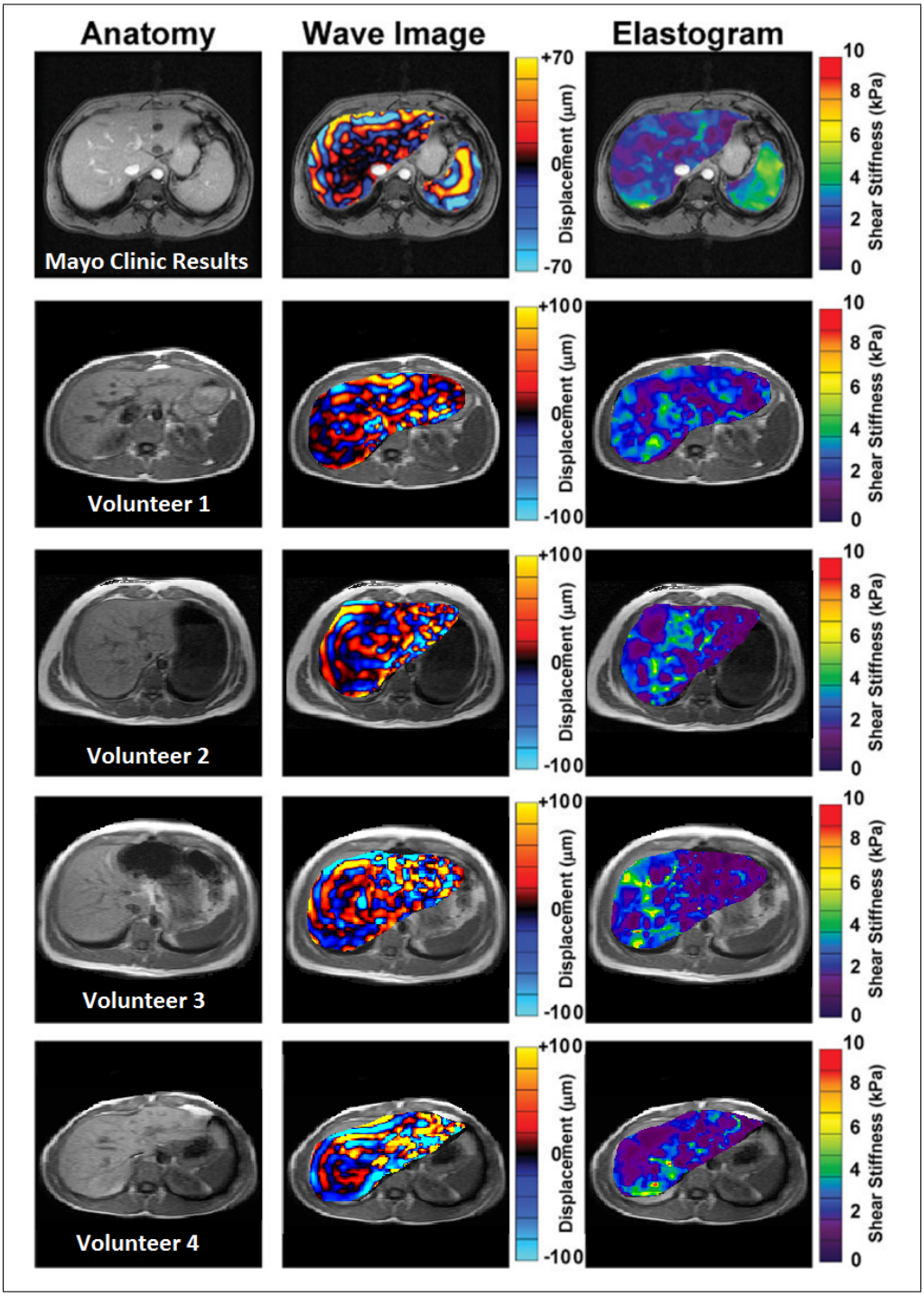

FIGURE 4.3. Comparison between Mayo Clinic (Yin et al., 2008) and our results. Anatomic images (left column), wave images (middle column) and elastograms (right column) of 5 normal volunteers are shown. While the first row of images was acquired by the Mayo Clinic team, the next 4 rows of images were acquired using our system. The resulting wave images and elastograms show similar measures in terms of wavelength and elasticity for all the 5 volunteers.

5. CONCLUSION AND FUTURE RESEARCH

We presented a modular system for obtaining Elasticity Maps using Magnetic Resonance, obtaining state-of-art comparable results for liver elasticity measuring. As we built the system from scratch by ourselves, our cost of acquisition was up to 10 times less than the available solutions in the market.

As this system is modular, improvement can be achieved separately on each subsystem.

This system can be used to assess elasticity on phantoms and human bodies. This allows research work on human volunteers, which is a value-added for our system.

Also, the elasticity values calculated while using our system are comparative to those found in the literature.

Although we implemented only a gradient-echo motion encoding, the theoretical framework enables the usage of any motion encoding sequence.

Future work can be based on several areas, including:

(1) The design of new geometries for the mechanical devices, where pressure drops usually can be avoided to improve the performance of the system.

(2) The development of new kind of sequences, while reducing the noise in the phase can improve the measure of local elasticity.

(3) The incorporation of more realistic elasticity models, to reach more accuracy in stiffness reconstruction.

References

Asbach, P., Klatt, D., Hamhaber, U., Braun, J., Somasundaram, R., Hamm, B., et al. (2008, August). Assessment of liver viscoelasticity using multifrequency MR elastography. *Magnetic resonance in medicine : official journal of the Society of Magnetic Resonance in Medicine / Society of Magnetic Resonance in Medicine*, *60*(2), 373–9.

Bishop, J., Samani, A., Sciarretta, J., & Plewes, D. B. (2000). Two-dimensional MR elastography with linear inversion reconstruction : methodology and noise analysis. *Strain*, *2081*.

Ehman, R. L., Rossman, P. J., Hulshizer, T. C., & Dresner, M. A. (2005, December). *Pressure activated driver for magnetic resonance elastography.*

Huwart, L., Salameh, N., Beek, L. ter, Vicaut, E., Peeters, F., Sinkus, R., et al. (2008, November). MR elastography of liver fibrosis: preliminary results comparing spin-echo and echo-planar imaging. *European radiology*, *18*(11), 2535–41.

Manduca, A., Muthupillai, R., Rossman, P., Greenleaf, J., & Ehman, R. (1996). Local wavelength estimation for magnetic resonance elastography. In *Image processing, 1996. proceedings., international conference on* (Vol. 3, pp. 527–530). IEEE.

Mariappan, Y. K., Glaser, K. J., & Ehman, R. L. (2010, July). Magnetic resonance elastography: a review. *Clinical anatomy (New York, N.Y.)*, *23*(5), 497–511.

Muthupillai, R., & Ehman, R. (1996). Magnetic resonance elastography. *Nature medicine*, *2*(5), 601.

Muthupillai, R., Lomas, D. J., Rossman, P. J., Greenleaf, J. F., Manduca, a., & Ehman, R. (1995, September). Magnetic resonance elastography by direct visualization of propagating acoustic strain waves. *Science (New York, N.Y.)*, *269*(5232), 1854–7.

Oliphant, T. E., Manduca, a., Ehman, R., & Greenleaf, J. F. (2001, February). Complex-valued stiffness reconstruction for magnetic resonance elastography by algebraic inversion of the differential equation. *Magnetic resonance in medicine : official journal of the Society of Magnetic Resonance in Medicine / Society of Magnetic Resonance in Medicine*, *45*(2), 299–310.

Rouvière, O., Yin, M., Dresner, M., Rossman, P., Burgart, L. J., Fidler, J., et al. (2006). MR Elastography of the Liver: Preliminary Results. *Radiology*, *240*(2), 440.

Talwalkar, J. a., Yin, M., Fidler, J. L., Sanderson, S. O., Kamath, P. S., & Ehman, R. (2008, January). Magnetic resonance imaging of hepatic fibrosis: emerging clinical applications. *Hepatology (Baltimore, Md.)*, *47*(1), 332–42.

Taouli, B., Ehman, R., & Reeder, S. B. (2009, July). Advanced MRI methods for assessment of chronic liver disease. *AJR. American journal of roentgenology*, *193*(1), 14–27.

Tse, Z. T. H., Janssen, H., Hamed, a., Ristic, M., Young, I., & Lamperth, M. (2009, April). Magnetic resonance elastography hardware design: a survey. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, *223*(4), 497–514.

Weaver, J. B., Van Houten, E. E. W., Miga, M. I., Kennedy, F. E., & Paulsen, K. D. (2001). Magnetic resonance elastography using 3D gradient echo measurements of steady-state motion. *Medical Physics*, *28*(8), 1620.

Yin, M., Talwalkar, J. A., Glaser, K. J., D, P., Manduca, A., Grimm, R. C., et al. (2008). A preliminary assessment of hepatic fibrosis with magnetic resonance elastography. *Clinics in Gastroenterology*, *5*(10), 1207–1213.