Published in final edited form as:

J Magn Reson Imaging. 2015 February ; 41(2): 369–375. doi:10.1002/jmri.24572.

Feasibility of Using 3D MR Elastography to Determine Pancreatic Stiffness in Healthy Volunteers

Yu Shi, MS^{1,2}, Kevin J. Glaser, PhD¹, Venkatesh K. Sudhakar, MD¹, Ephraim I. Ben-Abraham, BS³, and Richard L. Ehman, MD^{1,*}

¹Department of Radiology, Mayo Clinic, 200 1st St SW, Rochester, MN 55905

²Department of Radiology, Shengjing Hospital, China Medical University, No.36, Sanhao Street, Heping District, Shenyang, 110004, P.R. China

³Mayo Graduate School, Mayo Clinic, 200 1st St SW, Rochester, MN 55905

Abstract

Purpose—To evaluate the feasibility of using 3D Magnetic Resonance Elastography (MRE) to determine the stiffness of the pancreas in healthy volunteers.

Materials and Methods—Twenty healthy volunteers underwent 1.5-T MRE exams using an accelerated echo planar imaging (EPI) pulse sequence with low-frequency vibrations (40 and 60 Hz). Stiffness was calculated with a 3D direct inversion algorithm. The mean shear stiffness in five pancreatic subregions (uncinate, head, neck, body and tail) and the corresponding liver stiffness were calculated. The intrasubject coefficient of variation (CV) was calculated as a measure of the reproducibility for each volunteer.

Results—The mean shear stiffness (average of values obtained in different pancreatic subregions) was (1.15±0.17) kPa at 40 Hz, and (2.09±0.33) kPa at 60 Hz. The corresponding liver stiffness was higher than the pancreas stiffness at 40 Hz [(1.60±0.21) kPa, mean pancreas-to-liver stiffness ratio: 0.72], but similar at 60Hz [(2.12±0.23) kPa, mean ratio: 0.95]. The mean intrasubject CV for each pancreatic subregion was lower at 40 Hz than 60 Hz ($P<0.05$ for all subregions, range: 11.9–15.7% at 40 Hz and 16.5–19.6% at 60 Hz).

Conclusion—3D pancreatic MRE can provide promising and reproducible stiffness measurements throughout the pancreas, with more consistent data acquired at 40 Hz.

Keywords

MR elastography; feasibility; pancreas; healthy volunteers

Current imaging modalities are not sensitive enough to detect the early stages of either chronic pancreatitis (CP) or pancreatic ductal adenocarcinoma (PDAC). Conventional computed tomography has proven to be unreliable for detecting early-stage CP and PDAC (1, 2). Endoscopic retrograde pancreatography (ERCP) and MR cholangiopancreatography

* Address reprint requests to: R.L.E, 200 First St. SW, Rochester, MN FAX:507-284-9778, Tel:507-774-9781 55905. Ehman.Richard@mayo.edu.

(MRCP) provide excellent details and clear visualization of the ductal system, but mild disease may remain undetectable. Endoscopic ultrasound (EUS) is a sensitive procedure for evaluating and staging these diseases, but is invasive and detection of early-stage diseases without pathology is also controversial (3). In an attempt to overcome these deficiencies of modalities that only detect morphological changes, MR Elastography (MRE) offers a different approach for detecting diseases based on changes in tissue mechanical properties. MRE is a phase-contrast MRI technique for quantitatively assessing the stiffness of biological tissues by visualizing propagating shear waves in soft tissues (4). It has been shown to accurately assess hepatic fibrosis in patients with chronic liver diseases (5). Moreover, inflammation has also been shown to elevate tissue stiffness (6). Theoretically, both CP and PDAC, due to a build-up of fibrotic tissue and inflammatory changes, are likely to result in higher pancreatic stiffness compared to normal pancreas. It is worth exploring the feasibility of measuring pancreatic stiffness first in a cohort of normal subjects before investigating its potential as a clinical tool for detecting CP and PDAC.

In hepatic MRE, a 2D inversion model is typically enough to provide valid stiffness estimates due to the controlled and reproducible method used for introducing the motion into the liver. However, the location of the pancreas, its small size and complex shape, and the impact of the geometric boundary conditions and wave transmission factors on the propagation of the waves through the abdomen and into the pancreas require a 3D analysis of wave field data in the pancreas. This is analogous to work done in evaluating 3D vector MRE wave fields in the brain which showed that vibrating the head produced a zone where the waves propagated approximately in-plane and could be analyzed in 2D with only small biases in the results. Analysis outside of that zone requires a 3D analysis to account for deviations in the wave propagation direction that occur in other parts of the brain (7). Performing MRE of the pancreas presents unique technical challenges, including the introduction of shear waves deep into the body as well as performing efficient sampling and processing of a 3D vector displacement field. We have designed a passive driver, with a large area in tight contact with the body, to introduce shear waves into the pancreas and implemented a multislice spin-echo echo planar imaging (EPI) MRE pulse sequence to allow for fast volumetric acquisitions (8). Hence, the goal of this study was 1) to assess the feasibility of performing 3D pancreas MRE using a tailored driver to deliver low-frequency (40, 60 Hz) vibrations, 2) to compile preliminary normative values for the shear stiffness of the healthy pancreas, and 3) to evaluate the stiffness measurements of different subregions of the pancreas (tail, body, neck, head and uncinate).

MATERIALS AND METHODS

Healthy Volunteer Study

To determine if 3D MRE may be feasible in the pancreas for future clinical examinations, a healthy volunteer study was developed. Twenty healthy volunteers (10 men, 10 women) with no history of personal or familial pancreatic disease were enrolled in the study which was approved by our institutional review board. All volunteers gave written informed consent after the nature of the study had been explained to them. Their mean age, weight, height and body mass index were 33.60 ± 7.14 years (23–48 years), 70.25 ± 20.01 kg (45–

95 kg), 170.2 ± 10.2 cm (156–190 cm) and 24.05 ± 5.34 kg/m² (18.3–42.2 kg/m²), respectively. The volunteers were instructed to fast for 2–3 hours before the examination to avoid compression by a full stomach.

MRE Acquisitions

All examinations were performed on a 1.5T MR Scanner (HDx, GE Healthcare, Milwaukee, WI, USA). Each volunteer was imaged in the supine position (feet first). An external, 8-channel phased-array torso coil was used for signal reception. Low-amplitude mechanical waves at 40 Hz and 60 Hz were generated in the upper abdomen using an active acoustic generator located outside the scanner room. An ergonomic, soft driver was designed as a pillow-like mechanical transducer which conformed to the abdomen. It was centered at the epigastrium (partially on the rib cage), close to the pancreas and secured by a 20-cm wide elastic band wrapped around the abdomen, as shown in Fig. 1. The soft driver has two components: a flexible rectangular bag (14 cm×19 cm×2 cm) and a 3-dimensional structural filling material. The bag membrane has a builtin mesh in the material to prevent stretching. The filling material is springy and porous so that it can store air pressure and let air flow through the bag freely even under load. The soft driver is engineered to be airtight with a 60-cm long, 1.75-cm diameter antikink supply tube which connects to the pneumatic active driver via a 5-m-long, flexible, polyvinylchloride tube.

The propagating shear waves were imaged with a 2D multislice EPI pulse sequence modified to include additional motion-encoding gradients (MEGs) to record the tissue motion as phase in the MR images. The imaging parameters were TR/TE = 1875/39.6 ms (40 Hz), 2084/39.4 ms (60 Hz); phase offsets = 3; FOV = 38.4 cm; acquisition matrix = 96×96 (reconstructed to 256×256); number of signal averages = 1; frequency-encoding direction = RL; parallel imaging acceleration factor = 3; number of contiguous axial slices = 50 (interleaved; collected in 2 passes); slice thickness = 3 mm; MEG amplitude = 40 mT/m; bipolar MEG duration = 7.14 ms (1 on each side of the spin-echo refocusing pulse); receiver bandwidth = ±250.00 kHz. The acquisitions were performed at the end of expiration. The total acquisition time was split into six periods of suspended respiration of 15 seconds. Care was taken to monitor the level of expiration to obtain a consistent position of the pancreas for each acquisition.

In addition to the MRE acquisitions, we also performed two acquisitions to image the anatomy. The first was an axial two-dimensional fast imaging employing steady-state acquisition (2D FIESTA) scan with the following parameters: TR/TE = 3.9/1.7 ms; flip angle = 70°; FOV = 38.4 cm; acquisition matrix = 224×256; phase FOV = 0.65; 50 3-mm slices registered with the MRE acquisitions; and 2 signal averages. The acquisition was performed at the end of expiration with 3 17-second breath holds. The second was an axial T2-weighted fast-recovery fast spin-echo (T2 FRFSE) scan with the following parameters: TR/TE = 13636.4/109 ms; FOV = 38.4 cm; acquisition matrix = 320×192; number of signal averages = 2; phase FOV = 0.65; and 50 3-mm slices registered with the MRE acquisitions. Respiratory triggering was performed and the acquisition time was approximately 2–4 minutes. The total scan time was about 15–20 minutes.

MRE Inversion

The data were processed with custom MRE software. To remove slice-to-slice phase perturbations from the wave data, the data were lowpass filtered along the slice direction using a 1D 4th-order Butterworth lowpass filter with a cut-off frequency of 4 cycles/FOV_z. To remove longitudinal wave effects from the data, the vector curl of the displacement data was calculated using 3×3×3 derivative kernels on the wrapped phase data (9). The curl data were directionally filtered (DF) using 20 3D directional filters oriented isotropically with a radial 4th-order Butterworth bandpass filter with cut-off frequencies of 0.001 and 24 cycles/FOV_x (10). Each of the filtered datasets was further smoothed with a 5×5×5 quartic kernel (11) and a direct inversion (DI) of the Helmholtz wave equation was performed in 3×3×3 windows to get an estimate of the tissue stiffness (12). A weighted sum of the stiffness estimates from each directionally filtered dataset was performed using the squared-amplitude of the filtered curl data to produce the final elastogram.

Measurement of Stiffness

After 3D MRE processing, the processed images were analyzed in additional in-house software. The magnitude images were checked before the 1D Z-direction filter to verify that there were no noise, flow, or motion artifacts affecting the pancreas. The X, Y, and Z components of the displacement and curl vector fields were checked to verify wave propagation throughout the pancreas. The elastogram after DI with 3D DF was used for measuring stiffness. Regions of interest (ROIs) were drawn on the magnitude images showing the largest cross-section of each subregion of the pancreas. The ROIs were drawn to encompass as much of the pancreatic parenchyma as possible for each part, excluding the boundary of the pancreas (to avoid edge effects from the processing) and peripancreatic tissues (duodenum, stomach, portal vein and splenic vessels, etc). The ROIs were simultaneously drawn on the elastogram and the three wave images as well so that care could be taken to avoid areas with magnitude, wave, or stiffness artifacts. The shear wave images were checked to make sure that waves could be visualized in at least one component of the wave field in each portion of the pancreas (Fig. 2). One ROI was chosen for each pancreatic subregion (the tail, body, neck, head and uncinate process), as shown in Fig. 3. The pancreatic head was defined as the portion of the gland that lies to the right of the superior mesenteric vein and gives rise to the uncinate process. The pancreatic neck lies immediately anterior to the superior mesenteric vessels. The body is the portion of the pancreas that lies to the left of the superior mesenteric vessels. Since there is no clear border between the body and tail, for this study, we followed the method published by Kimura et al. (13), using one-half of the distance between the neck and the end of the pancreas to divide the body and tail. As a reference, liver stiffness was also measured manually following the technique reported by Venkatesh et al. (14).

Statistical Analysis

After testing for normality using the Shapiro-Wilks test ($P > 0.05$ for the pancreatic subregions at 40 Hz and 60 Hz), the data were described as the mean $\pm$ standard deviation (SD). The intrasubject coefficient of variation (CV) was used as a measure of variability in the stiffness measurement of each pancreatic subregion within the same subject,

representing the intrinsic variability for each measurement. The intersubject CV was used to report the variability across the subjects, calculated as the SD divided by the total mean for all subjects. Comparisons of the mean shear stiffness were performed using the nonparametric Wilcoxon matched-pairs signed-ranks test. A *P* value <0.05 was considered significant.

RESULTS

1. Pancreatic Shear Stiffness at 40 Hz and 60 Hz

MRE was successfully performed in the pancreas of all volunteers. The wave images showed better wave illumination, a more planar wave pattern, and higher amplitude of motion with less attenuation and interference at 40 Hz than at 60 Hz (Fig. 2). The mean shear stiffness of the pancreatic tail, body, neck, head and uncinate were significantly lower at 40 Hz than at 60 Hz (all *P* < 0.001, Wilcoxon matched-pairs signed-ranks test for each subregion). The overall mean shear stiffness (average of values obtained in different pancreatic subregions) was (1.15 ± 0.17) kPa at 40 Hz and (2.09 ± 0.33) kPa at 60 Hz, as shown in Table 1. After a pairwise comparison of stiffness between the pancreatic subregions at both frequencies, no statistical difference was found except in the neck, which presented a lower stiffness than the other regions (*P* < 0.05 as compared with the stiffness of the tail, body, head and uncinate at 40 Hz; and with the tail, body and uncinate at 60 Hz). The corresponding liver shear stiffness was (1.60 ± 0.21) kPa (range: 1.22–2.00 kPa) at 40 Hz and (2.12 ± 0.23) kPa (range: 1.78–2.59 kPa) at 60 Hz. The mean ratio of pancreatic stiffness over liver stiffness was 0.72 at 40 Hz and 0.95 at 60 Hz. The shear stiffness for each individual subject in each pancreatic region was consistent among the subjects (approximately 1.0–1.5 kPa at 40 Hz and 1.5–2.5 kPa at 60 Hz), with a stiffness gap of (0.93 ± 0.28) kPa.

2. Intrasubject and Intersubject CV

The mean intrasubject CV of each pancreatic subregion was significantly lower at 40 Hz than that at 60 Hz (*P* = 0.001–0.012 for different subregions), with overall mean CVs of 13.4% at 40 Hz and 18.1% at 60 Hz, as shown in Table 1. At both 40 Hz and 60 Hz, the shear stiffness of the neck showed the highest intrasubject CV, with statistical differences (*P* < 0.05) compared with the body, head and uncinate at 40 Hz, and with the head at 60 Hz. Across individual subjects, the intrasubject CV was slightly lower at 40 Hz than at 60 Hz. The overall intersubject CV was also slightly lower at 40 Hz (14.9%) than at 60 Hz (15.6%), though without statistical significance (*P* > 0.05).

3. BMI and Stiffness Measurements

Generally, compared with the surrounding retroperitoneal fat, the elastograms delineated the pancreas more clearly at 40 Hz than at 60 Hz. For the subjects with relatively small or normal BMI, the elastograms showed comparable size with the magnitude or T2W images at both 40 Hz and 60 Hz, though they were noisier at 60 Hz. For the obese subjects, the pancreas size on the elastograms at 60 Hz became smaller with a little distortion. However, the sample size in this study was too small to be classified by BMI or to show any statistical results between BMI and stiffness, as shown in Fig. 4.

DISCUSSION

In this study, pancreatic MRE using EPI MRE was shown to be feasible in healthy volunteers at low mechanical frequencies. The data from each main region of the pancreas was consistent among the volunteers, with generally lower intrasubject and intersubject variability at 40 Hz. The total mean pancreatic stiffness was nearly three quarters of the liver stiffness at 40 Hz and almost identical to the liver stiffness at 60 Hz.

1 Feasibility of 3D EPI Pancreatic MRE

The liver is an ideal organ for MRE because it has a homogenous parenchyma and extends close to the body wall, which allows for good shear wave penetration. A 2D gradient-echo acquisition with 2–4 slices centered at the hilar level is usually enough to represent the diffuse changes of the whole liver (14). On the contrary, the pancreas is an elongated, tapered organ located across the back of the abdomen, resulting in a significant challenge to get adequate shear wave penetration throughout the whole pancreas. The pattern of wave propagation in the pancreas is extremely complex and includes waves propagating at oblique angles relative to an axial plane of section. Given these characteristics, a 3D MRE analysis, in this case using a 2D multislice EPI acquisition with a 3D inversion, is a necessity for measuring pancreatic stiffness as it is in other organs where the shear waves cannot be assumed to propagate only within the imaging plane (8). As shown in liver studies, 3D MRE using spin-echo EPI can have increased signal and better illumination with fewer artifacts from the bowels (14–16). Moreover, this EPI-based technique can significantly reduce the acquisition time for volumetric MRE scans (17). In 6 short breath-holds, 50 slices covering the entire pancreas, both kidneys, and a significant part of the liver is possible.

Ideally, the mechanical waves produced during MRE at the surface of the body would be transmitted through the stomach, the bowels and the liver to the different regions of the pancreas. However, compared with the pancreas that is fixed tightly in the retroperitoneal space, the gastrointestinal tracts with gas and luminal contents are very flexible, limiting the wave propagation to the pancreas. While the liver might transmit the waves directly to the right half of the pancreas, it is too far away for adequate transmission to the body and tail. It is also possible that mechanical waves could be introduced from the back. However, this can also be a long distance for the waves to transverse while also being impeded by the vertebrae, musculature in the back and the kidneys. For this reason, we used a long and soft passive driver with improved abdomen-driver mechanical coupling that covers a wide range of the upper abdomen, enabling the waves to be generated uniformly and allows the entire abdomen to vibrate and generate shear waves, rather than vibrating one spot with excessive energy and significant phase wrapping in the images. Additionally, the pancreas winds behind the stomach and is inclined to be compressed by a full stomach. Therefore, a fasted state is strongly recommended to avoid artifacts from the stomach and deformity of the pancreatic tail and body.

2 The Shear Stiffness of Pancreatic Subregions at Low Frequencies

In our study, two low vibration frequencies were tested to avoid the rapid decrease of wave amplitude with penetration depth that occurs at high frequencies. The chosen frequencies

included 60 Hz, which is commonly used for liver MRE, and the lower 40-Hz frequency. Our results indicated that 40-Hz vibrations might be more suitable for pancreatic MRE based on lower intrasubject CV and better shear wave penetration. Theoretically, the data at 60 Hz should give rise to improved resolution since the shorter shear wavelength at the higher frequency can improve the calculation of stiffness. However, the shear wave amplitude decreased more rapidly in deeper tissue at 60 Hz than at 40 Hz. By reviewing the literature reporting 2D MRE results in the liver using 60-Hz vibrations, the wave amplitude is weaker in the inner half of the liver, especially for healthy livers which have shorter shear wavelengths (18, 19). A 3D MRE acquisition or processing can improve the interpretation of the wave field generated in the tissue, but cannot change the mechanical propagation of the waves itself. This also accounts for the increased attenuation of waves in obese subjects, which causes noise biases and false stiffness estimates. Hence, we speculate that the challenge of producing effective, reproducible wave propagation at 60 Hz in the pancreas will limit its usefulness for routine, clinical imaging.

In this study, the pancreas stiffness at 60 Hz was very similar to the liver stiffness, though our liver stiffnesses were a little lower than those reported in the literature (18, 19). Previous work has also reported a similar discrepancy in brain stiffness between 2D and 3D analysis (8). We speculate that our 3D MRE data had lower hepatic stiffness for two reasons. First, through-plane propagating waves can appear in 2D data as waves with longer wavelengths, thereby resulting in an overestimate of the tissue stiffness. Second, the thin-slice acquisition used in this work to produce a 3D sampling of the wave field can have lower SNR than the standard thick-slice 2D MRE acquisitions. This introduces more noise into the images, which the image processing can interpret as the short-wavelength waves of soft tissue. There are very few existing studies reporting MRE results in the pancreas. Yin et al. reported that the mean shear stiffness of pancreatic tissue in ten normal volunteers was (2.0 ± 0.4) kPa at 60 Hz, which is consistent with our result of (2.09 ± 0.33) kPa (20). The data in our study from each pancreatic subregion was almost identical, corresponding to the histological homogeneity of the pancreas. The neck was estimated to be slightly softer than the other regions at 40 Hz with higher inter- and intrasubject variability. The partial volume effect due to the small size of the pancreas in this region rather than the real mechanical properties may explain this variation.

3 Possible Application for Pancreatic Diseases

Histologic changes of the normal pancreatic architecture due to chronic pancreatitis and pancreatic ductal adenocarcinoma include both fibrosis and inflammatory cell infiltrates. Recent studies with ultrasonic endoscopic elastography reported different mean strain ratios in the healthy pancreas, mass-forming pancreatitis and pancreatic cancer. Strain ratio was defined as the quotient B/A by analyzing the elastogram in the representative areas of the mass (A) and soft reference areas (B). Both Itokawa (21) and Iglesias-Garcia (3) have reported results indicating that healthy pancreas has the lowest strain ratio, and inflammatory masses have a higher strain ratio than healthy pancreas, but lower than pancreatic adenocarcinoma and endocrine tumors. Although the comparison of these results with MRE is difficult due to the intrinsic differences between the different modalities, these

results provide motivation for exploring MRE as a potential tool for detecting the mechanical properties of pancreatic diseases.

4 Limitations

There are several limitations to the current study that will need to be addressed in future studies to validate the results of this study and to make the technique practical for clinical applications. First the sample size in this study was small and future studies will need to involve more subjects with larger BMI and older subjects. Second, evaluating the effectiveness or quality of the shear wave transmission into the pancreas is still complicated and not normative, calling for a confidence map or error metric that reflects the quality of the wave data and the stiffness estimates from the 3D wave data. Finally, in patients with atrophy and fatty infiltration of the pancreas, the current mechanical properties reported by MRE will reflect the composite properties of adipose and pancreatic tissue.

CONCLUSION

3D pancreatic MRE using a multislice EPI acquisition can provide promising and stable stiffness measurements throughout the pancreas. The mean pancreatic stiffness was (1.15 ± 0.17) kPa at 40 Hz and (2.09 ± 0.33) kPa at 60Hz, with lower variation and better wave propagation at 40 Hz. Our results provide data that will enable future investigations of this technique as a possible clinical tool.

Acknowledgments

Contract grant sponsor:

NIH grant EB001981.

National Natural Science Foundation of China (No. 81271566).

References

1. Witt H, Apte MV, Keim V, Wilson JS. Chronic pancreatitis: challenges and advances in pathogenesis, genetics, diagnosis, and therapy. *Gastroenterology*. 2007; 132(4):1557–1573. [PubMed: 17466744]
2. Janssen J, Schlörner E, Greiner L. EUS elastography of the pancreas: feasibility and pattern description of the normal pancreas, chronic pancreatitis, and focal pancreatic lesions. *Gastrointest Endosc*. 2007; 65(7):971–978. [PubMed: 17531630]
3. Iglesias-García J, Lindkvist B, Lariño-Noia J, Domínguez-Muñoz JE. The role of EUS in relation to other imaging modalities in the differential diagnosis between mass forming chronic pancreatitis, autoimmune pancreatitis and ductal pancreatic adenocarcinoma. *Rev Esp Enferm Dig*. 2012; 104(6):315–321. [PubMed: 22738702]
4. Mariappan YK, Glaser KJ, Ehman RL. Magnetic resonance elastography: a review. *Clin Anat*. 2010; 23(5):497–511. [PubMed: 20544947]
5. Yin M, Chen J, Glaser KJ, Talwalkar JA, Ehman RL. Abdominal magnetic resonance elastography. *Top Magn Reson Imaging*. 2009; 20(2):79–87. [PubMed: 20010062]
6. Wang Y, Ganger DR, Levitsky J, et al. Assessment of chronic hepatitis and fibrosis: comparison of MR elastography and diffusion-weighted imaging. *AJR Am J Roentgenol*. 2011; 196(3):553–561. [PubMed: 21343496]

7. Hamhaber U, Sack I, Papazoglou S, Rump J, Klatt D, Braun J. Three-dimensional analysis of shear wave propagation observed by in vivo magnetic resonance elastography of the brain. 2007; 3(1): 127–137.
8. Murphy MC, Huston J 3rd, Jack CR Jr, et al. Decreased brain stiffness in Alzheimer's disease determined by magnetic resonance elastography. *J Magn Reson Imaging*. 2011; 34(3):494–498. [PubMed: 21751286]
9. Glaser, K.; Ehman, RL. MR Elastography Inversions Without Phase Unwrapping. Proceedings of the International Society for Magnetic Resonance in Medicine; Honolulu, HI. 2009. p. 4669
10. Manduca A, Lake DS, Kruse SA, Ehman RL. Spatio-temporal directional filtering for improved inversion of MR elastography images. *Med Image Anal*. 2003; 7:465–473. [PubMed: 14561551]
11. Romano AJ, Bucaro JA, Ehman RL, Shirron JJ. Evaluation of a material parameter extraction algorithm using MRI-based displacement measurements. *IEEE Trans Ultrason Ferroelectr Freq Control*. 2000; 47(6):1575–1581. [PubMed: 18238703]
12. Manduca A, Oliphant TE, Dresner MA, et al. Magnetic resonance elastography: non-invasive mapping of tissue elasticity. *Med Image Anal*. 2001; 5(4):237–254. [PubMed: 11731304]
13. Kimura W, Nagai H, Kuroda A, Muto T, Esaki Y. Analysis of small cystic lesions of the pancreas. *Int J Pancreatol*. 1995; 18(3):197–206. [PubMed: 8708390]
14. Venkatesh SK, Yin M, Ehman RL. Magnetic resonance elastography of liver: technique, analysis, and clinical applications. *J Magn Reson Imaging*. 2013; 37(3):544–555. [PubMed: 23423795]
15. Yin, M.; Grimm, RC.; Manduca, A.; Ehman, RL. Rapid EPI-based MR elastography of the liver. Paper presented at: Proceedings of the International Society for Magnetic Resonance in Medicine; May 6–12, 2006; Seattle, Washington. p. 2268
16. Yin, M.; Manduca, A.; Romano, AJ., et al. 3-D local frequency estimation inversion for abdominal MR elastography. Paper presented at: Proceedings of the International Society for Magnetic Resonance in Medicine; May 19–25, 2007; Berlin, Germany. p. 960
17. Nedredal GI, Yin M, McKenzie T, et al. Portal hypertension correlates with splenic stiffness as measured with MR elastography. *J Magn Reson Imaging*. 2011; 34(1):79–87. [PubMed: 21608066]
18. Venkatesh SK, Gang W, Teo LLS, Ang BWL. Magnetic resonance elastography of liver in healthy asians: Normal liver stiffness quantification and reproducibility assessment. *J Magn Reson Imaging*. 2013 Oct 2. Epub ahead of print. 10.1002/jmri.24084
19. Venkatesh SK, Xu S, Tai D, Yu H, Wee A. Correlation of MR elastography with morphometric quantification of liver fibrosis (Fibro-C-Index) in chronic hepatitis B. *Magn Reson Med*. 2013 Oct 28. Epub ahead of print. 10.1002/mrm.25002
20. Yin, M.; Venkatesh, SK.; Grimm, RC.; Rossman, PJ.; Manduca, A.; Ehman, RL. Assessment of the pancreas with MR elastography. Paper presented at: Proceedings of the International Society for Magnetic Resonance in Medicine; May 3–9, 2008; Toronto, Canada. p. 2627
21. Itokawa F, Itoi T, Sofuni A, et al. EUS elastography combined with the strain ratio of tissue elasticity for diagnosis of solid pancreatic masses. *J Gastroenterol*. 2011; 46(6):843–853. [PubMed: 21505859]

Figure 1.

Left: coronal depiction of the pancreatic MRE setup with the passive driver position indicated (square). Right: photo of the passive driver with ruler as reference (14×19×2 cm).

Figure 2.

The magnitude and wave images of the three components of the vector displacement field at 40 Hz (upper row) and 60 Hz (lower row). The wave images show better illumination of the pancreas at 40 Hz than 60 Hz. The red (magnitude images) and yellow (wave images) outlines highlight the pancreas.

Figure 3. Axial magnitude images (upper row) and elastograms (lower row) of the pancreatic tail, body, neck, head and uncinate at 40 Hz, respectively. One ROI was placed for each pancreatic subregion. The red outlines highlight different subregions of the pancreas.

Figure 4.

Axial FIELSTA T2-weighted images (T2WI) (1st column), EPI MRE magnitude images (2nd column) and maximum intensity projection (MIP) of the magnitude (3rd column) and elastograms (4th column at 40 Hz and 5th column at 60 Hz) of subjects with different body mass index (BMI). Row 1: BMI=19.02 kg/m²; Row 2: BMI=22.49 kg/m²; Row 3: BMI=42.16 kg/m². At 40 Hz, the entire pancreas was delineated clearly on the elastogram regardless of the BMI. At 60 Hz, the edges of the pancreas are less distinguishable at normal BMI. For the subject with the highest BMI, the shape of the pancreas on the elastogram did not strictly conform to the anatomic morphology on the T2WI, with smaller size and slight deformity.

Table 1

Shear stiffness of various pancreatic subregions measured at 40 Hz and 60 Hz

Frequency	Main subregions	Mean±SD (kPa)	Intersubject CV (%)	Minimum (kPa)	Maximum (kPa)	Intrasubject CV (%)
40 Hz	Tail	1.22±0.13	10.6	1.03	1.46	13.5
	Body	1.20±0.16	13.6	0.96	1.54	12.9
	Neck	1.02±0.17	16.8	0.72	1.42	15.6
	Head	1.14±0.17	15.1	0.87	1.50	11.9
	Uncinate	1.18±0.16	13.2	0.89	1.53	12.9
60 Hz	Total	1.15±0.17	14.9	0.72	1.54	13.4
	Tail	2.18±0.23	10.6	1.70	2.68	18.4
	Body	2.19±0.30	13.5	1.67	2.69	18.1
	Neck	1.93±0.31	16.1	1.42	2.49	19.6
	Head	2.04±0.41	20.3	1.63	3.47	16.3
	Uncinate	2.09±0.31	14.7	1.56	2.96	18.2
	Total	2.09±0.33	15.6	1.42	3.47	18.1

* CV: coefficient of variation