IN VIVO LIVER TISSUE MECHANICAL PROPERTIES BY TRANSIENT ELASTOGRAPHY: COMPARISON WITH DYNAMIC MECHANICAL ANALYSIS

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ABSTRACT

Understanding the mechanical properties of human liver is one of the most critical aspects of its numerical modeling for medical applications or impact biomechanics. Generally, model constitutive laws come from *in vitro* data. However, the elastic properties of liver may change significantly after death and with time. Furthermore, *in vitro* liver elastic properties reported in the literature have often not been compared quantitatively with *in vivo* liver mechanical properties on a same organ.

In this study, both steps are investigated on porcine liver. The elastic property of the porcine liver, given by the shear modulus *G*, was measured by both Transient Elastography (TE) and Dynamic Mechanical Analysis (DMA). Shear modulus measurements were realized on *in vivo* and *in vitro* liver to compare the TE and DMA methods and to study the influence of testing conditions on the liver viscoelastic properties.

In vitro results show that elastic properties obtained by TE and DMA are in agreement. Liver tissue in the frequency range from 0.1 to 4 Hz can be modeled by a two-mode relaxation model. Furthermore, results show that the liver is homogeneous, isotropic and more elastic than viscous. Finally, it is shown in this study that viscoelastic properties obtained by TE and DMA change significantly with *post mortem* time and with the boundary conditions.

Keywords: Soft Tissues, Animals, Instruments, Dynamics.

DURING IMPACTS, THE SOLID ORGANS OF THE ABDOMEN APPEAR TO BE MORE AT RISK THAN THE HOLLOW ORGANS. Moreover, injury to the liver and spleen can be life threatening. Injury occurs when the local mechanical load, exerted on the organ, exceeds certain tolerance levels. Understanding how an external mechanical load on the abdomen is transferred to a local mechanical load in a solid organ is needed to improve injury protecting devices and diagnostic methods. To obtain this understanding, finite element (FE) modelling is often used. Current FE abdomen models contain a detailed geometrical description of the organs but lack an accurate description of material behaviour. Thus, characterizing the *in vivo* mechanical properties of solid organs of the abdomen and in particular of the liver is of a great interest.

Data on the biomechanical properties of the liver generally include two distinct stages. In a first step, experimental curves linking strain and stress can be obtained from a specific experimental set up. In a second step, the parameters of the constitutive laws governing the behaviour of a tissue can be identified from these curves. On the initial stage, there are in general two different possible data sources: *in vitro* data extracted from rheological tests where a liver sample is positioned on a testing instrument and *in vivo* data obtained by indentation or by imaging technique-based methods.

Concerning the *in vitro* investigations, Sakuma et al. (2003) conducted tests in compression and elongation. As the boundary conditions are easily identified with this testing method, they can represent the properties of the material in the form of stress-strain curves. Liu and Bilston (2000) identified *in vitro* properties of bovine liver in a linear viscoelastic domain (less than 0.2% strain) by shear relaxation experiments. They used a generalized Maxwell model to get the linear viscoelastic

behaviour of tissues. They also conducted similar experiments to measure the *in vitro* behaviour of bovine liver under large strains (Liu and Bilston, 2002). Kerdok et al. (2005) developed an instrument to measure *in vitro* inner local mechanical properties of porcine liver with a perfusion system. Although only preliminary results have been obtained, they determined that the perfusion system can approach the *in vivo* conditions for at least two or three hours after the perfusion. The Young's modulus of *in vitro* liver is not often determined. Generally, the viscoelastic shear modulus is used to characterize the liver.

In vivo tests are based on a system of position and force sensors set inside the abdomen to perform an indentation or on elastometric images where an imaging modality - ultrasound imaging for transient elastography (TE) (Sandrin et al., 2003) or magnetic resonance imaging for magnetic resonance elastography (MRE) (Kruse et al., 2000) - can provide information on the Young's modulus of living materials. Nava et al. (2004) measured the properties of the liver of a human body using an aspiration device which consists of a small vacuum tube and an imaging device. They found a Young's modulus equal to 90 kPa with the Glisson's capsule (the membrane which surrounds the liver). Elastography techniques are methods increasingly used for the characterization of liver tissue and particularly for the diagnosis of liver diseases. Indeed, in MRE, Huwart et al. (2006) reported an average *in vivo* shear modulus with values between 2.24 ± 0.23 kPa (substantial fibrosis) and 4.68 ± 1.61 kPa (cirrhosis) in patients at 65 Hz, Klatt et al. (2006) an *in vivo* shear modulus between 1.99 ± 0.16 and 3.07 ± 0.88 kPa in healthy volunteers at 51 Hz, Rouviere et al. (2006), an *in vivo* shear modulus of 2.0 ± 0.3 kPa in volunteers at 80 Hz and Yin et al. (2007) an *in vivo* shear modulus through the liver in healthy persons of 2.20 ± 0.31 kPa at 60 Hz. In imaging acoustic radiation force, Palmeri et al. (2008) has found a shear modulus between 0.9 and 3.0 kPa, with a mean accuracy of ± 0.4 kPa in volunteers. In TE, the human *in vivo* shear modulus of a healthy liver is usually within the range from 0.8 to 2.3 kPa at 50 Hz (Castera et al., 2008; Fraquelli et al., 2007).

From this wide variety of studies, it is difficult to choose a particular model among the proposed ones to describe the mechanical behaviour of the liver because each experiment has its advantages and disadvantages. For example, a significant perfusion of liver strongly affects its rheology (the liver receives a fifth of the total flow of blood at any time) and remains an open question: can *in vitro* experiments be sufficient to evaluate the properties of living tissue, even if attention is taken to prevent dehydration or swelling of tissues? Conversely, data obtained from *in vivo* experiments must also be treated with caution, because the tissue response may depend on the testing area (linked to specific boundary conditions or non-homogeneity of the material) and the influence of the loading tool on the strain may not be well understood. Moreover, respiratory and circulatory movements may also affect the *in vivo* data. Moreover, another important source of uncertainty in measurements is the state of liver deformation during indentation. Indeed, Brown et al. (2003) pointed out that most researchers condition their liver samples using several cycles of indentation to have repeatable estimates of the elasticity and hysteresis. However, during surgery or in case of impact, the liver is not preconditioned or if it is, it is negligible. Besides, the rheology of the liver is not only influenced by the perfusion, but also by the Glisson's capsule. Carter et al. (2001) showed that the elasticity of cylindrical liver samples parenchyma having kept part of the Glisson's capsule is double those without Glisson's capsule, conducting similar rheological tests. They identified a Young's modulus of 270 kPa with the Glisson's capsule.

Therefore, data from the literature on mechanical properties of the liver are highly variable. They depend not only on testing conditions, but also on measurement methods. The probably most important limitation is that most experiments were conducted *in vitro* on excised samples, while knowledge of the mechanical behaviour of liver tissue in its environment (blood, active metabolism, etc) remains unknown. Although some studies (Brouwer et al., 2001; Tay et al., 2006; Valtorta et al., 2005) give results on the *in vivo* mechanical properties of human liver, the invasiveness of such devices limits conclusions about the mechanical behaviour of the liver in its natural environment owing to the result discrepancy.

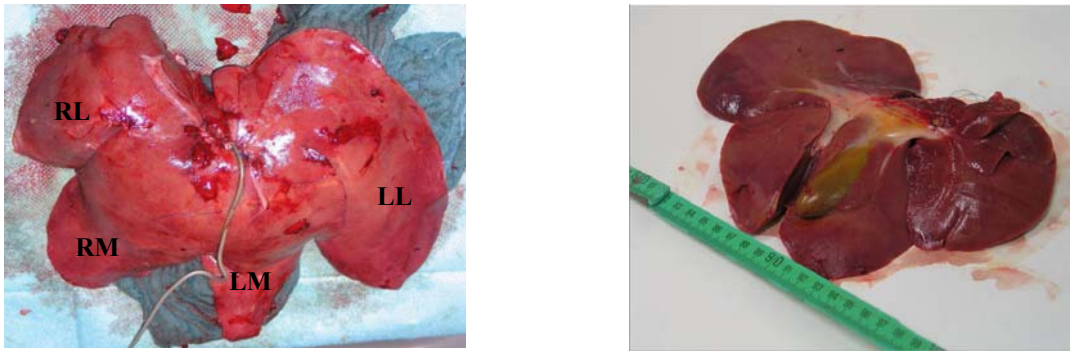
In this paper, a preliminary study on the characterization of porcine liver is performed in order to obtain *in vivo* and *in vitro* liver viscoelastic properties and to highlight the influence of testing conditions and measurement techniques on a single organ. The measurements are carried out by

transient elastography (TE) which is a non-invasive technique and by dynamic mechanical analysis (DMA).

MATERIALS AND METHODS

In the present study, a brief outline of the theoretical background which underlines the test methods for measuring the mechanical properties of soft tissues is presented. The *in vivo* and *in vitro* mechanical properties of porcine liver, given by the shear modulus G, were studied on a single organ. *In vivo* measurements were performed by transient elastography and *in vitro* experiments by both transient elastography and dynamic mechanical analysis to compare the results obtained with both techniques.

The study was carried out on the liver of 5 female pigs weighting between 25 and 35 kg. Contrary to human liver, porcine liver has four lobes as illustrated in Fig. 1. However, its metabolic functions are similar as the human ones. Fig. 1a and Fig. 1b show the diaphragmatic and visceral faces of a porcine liver, respectively. RL, LL, RM and LM correspond to the right lateral lobe, the left lateral lobe, the right median lobe and the left median lobe, respectively.



a) Diaphragmatic Face **b) Visceral Face**

Fig. 1 - View of Porcine Liver.

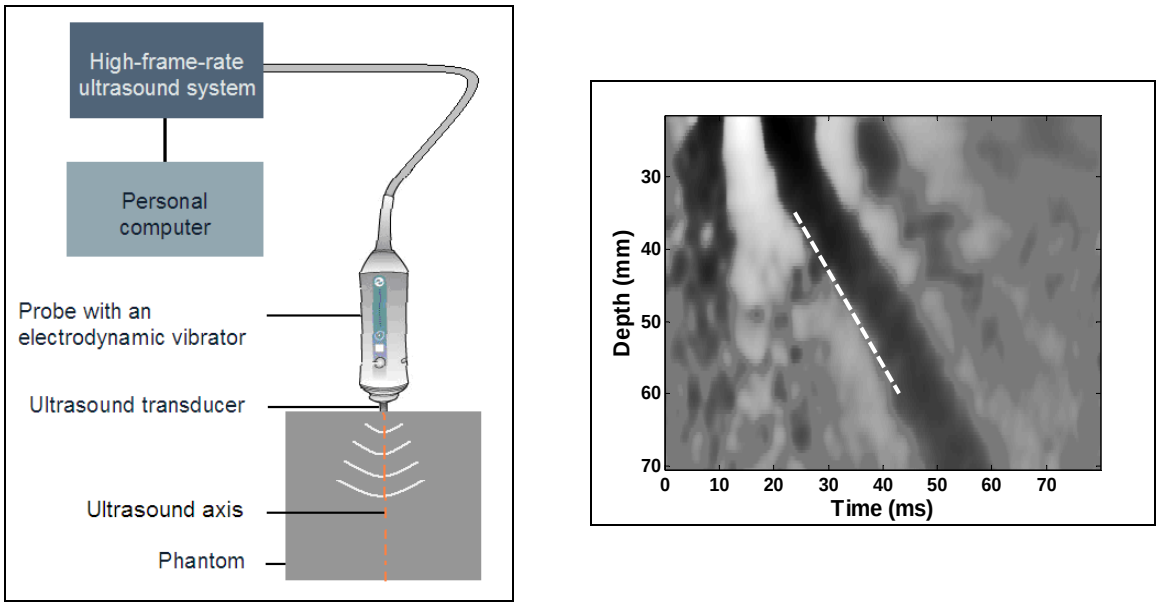
TRANSIENT ELASTOGRAPHY-BASED METHOD

The transient elastography technique consists of applying a low-frequency transient mechanical vibration to the studied medium and following the propagation of an elastic shear wave induced by this vibration. The low-frequency transient vibration is a period of a sine wave at a 50 Hz frequency and 2 mm peak-peak amplitude. The TE device is composed of a probe and an ultra-fast ultrasound imaging system. The probe comprises an electrodynamic vibrator and a single ultrasound transducer. A schematic diagram of the TE set up is given in Fig. 2a. The propagation of the shear wave is monitored via the ultrasound imaging system at a high frame rate of about 6 kHz. The acquired ultrasound lines were cross-correlated to calculate the displacement induced by the shear wave propagation. Displacements were measured along the ultrasound axis which intercepted the vibration axis. Typical displacements are about 10-100 µm. A spatial-temporal strain map was then computed from the recorded displacements. The shear wave speed V_s was calculated based on the slope of the wave front visualized in the strain map as in Sandrin et al. (2003). An example of a spatial-temporal strain map obtained on *in vivo* liver is given Fig. 2b. The slope of the dashed line corresponds to the shear wave speed.

The elastic properties of the tissue were obtained from the measurement of the elastic wave propagation parameters. The shear modulus is estimated from the measured shear wave speed and is expressed in kPa. In an isotropic, homogeneous, soft, purely elastic and linear medium, the shear wave speed V_s only depends on the shear modulus G:

$$G = \rho V_s^2 \quad (1)$$

where ρ is the mass density. In soft tissues, ρ is assumed to be 1.0 g/cm^3.



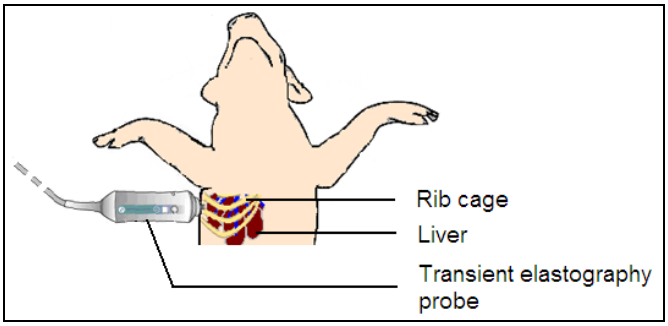
a) TE Device **b) Shear Wave Front**

Fig. 2 - Schematic Diagram of Transient Elastography (TE) Setup and Spatial-Temporal Strain Map.

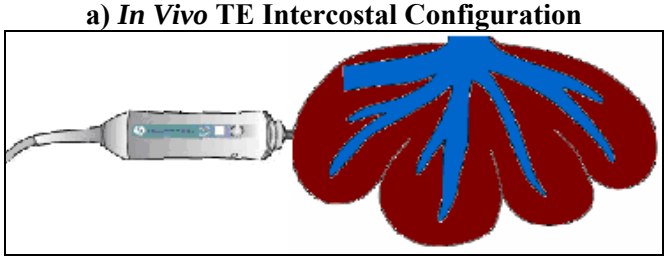
For the *in vivo* TE tests, the anesthetised and intubated animal was placed on the operating table in the supine position, upper and lower limbs attached to the table. Before starting these tests, a prior localisation of the animal liver was necessary and was performed using a portable ultrasound (Echo Blaster, TELEMED Ltd, Lithuania). For these *in vivo* measurements, the probe of the TE device was at the surface of the skin of the animal, placed between its ribs (intercostal position) next to the right lobe of the animal liver as shown in Fig. 3a. This position is closer to the use of the TE device in clinical routine (Sandrin, 2003). The rib cage protected the liver from a compression that might have an influence on liver elastic properties.

For the *in vitro* TE investigations, the animal underwent a hepatectomy. The hepatectomy generates a haemorrhage which entails rapidly the death of the animal. Tests were performed on the entire clamped liver for the TE tests just after the death of the animal. Therefore, the liver temperature remains the same as the internal temperature of the animal namely 34°C. The shear modulus was measured with the probe directly in contact with the clamped liver as illustrated in Fig. 3b. The pressure of the probe on the organ was kept as low as possible.

In each case, the calculated shear modulus was the median of ten valid measurements. The interquartile range *IQR* was also computed. It gives information on data dispersion around the median. It should be noted that TE tests were without effect on the liver, the TE device being non-invasive. TE results can be thus compared to the *in vitro* results obtained by DMA on the same animal.



a) *In Vivo* TE Intercostal Configuration



b) *In Vitro* TE Configuration

Fig. 3 - Position of the Probe for *In Vivo* and *In Vitro* Transient Elastography Measurements

DYNAMIC MECHANICAL ANALYSIS TESTS

To quantify the dynamic behaviour of the porcine liver, oscillation tests were performed. A sinusoidal strain is imposed to the medium; a sinusoidal stress is induced at the same frequency but is phase-shifted ahead by an angle δ. The complex viscoelastic modulus G^* is then defined as the ratio of stress and strain amplitude. The stress-strain relation may be represented as having a component in phase with the strain (representing the elasticity) and a component which is 90° out of phase with the strain (representing the viscosity), thus giving rise to storage and loss moduli G' and G'', respectively. Besides, the viscous damping $\tan \delta$, defined as the ratio of the loss and storage modulus, is often used as an indicator of the amount of strain energy lost relative to the energy stored per cycle.

To determine the limit of linear viscoelasticity of porcine liver, strain sweep oscillation tests were performed on stress-controlled rheometer using cylindric shaped samples. Tests were conducted at constant frequency with varying strain amplitude. In the region of linear viscoelasticity, the values of dynamic moduli G' and G'' as a function of the applied strain remain steady. Beyond a certain strain, they start changing significantly. This strain specifies the limit of linear viscoelasticity, beyond which the behaviour of the material is non-linear.

The *in vitro* DMA tests were carried out on liver cylindric shaped samples directly extracted from the organs previously tested *in vivo*. Just after the animal death, the entire liver was placed in an insulated container at 6°C surrounded by ice and was brought to the laboratory where the DMA tests were performed. At the laboratory, liver samples were taken from the entire organ to be tested within 6h after the death of the animal. The samples were placed in a beaker whose base was cooled by ice to prevent *post mortem* decay. Tests were carried out on samples of 20 mm diameter and 4 ± 1 mm thickness. Samples of less than 3 mm thick would be subjected to dehydration by absorption of blood by the sandpaper that covered the surface of the plates of the rheometer and more than 5 mm to wave propagation effects. The samples were cut parallel to the Glisson's capsule that surrounds the liver in order to have uniform samples texture as shown in Fig. 4. Samples were taken from the left and right side lobes as presented in Fig. 1. The left and right median lobes were not appropriate due to the presence of numerous major veins. Indeed, the presence of structures like blood vessels or other veins generated inhomogeneities in the sample. To study the isotropic and homogeneous properties of hepatic tissue, samples were taken from the different lobes and perpendicular to the Glisson's capsule (Fig. 4).

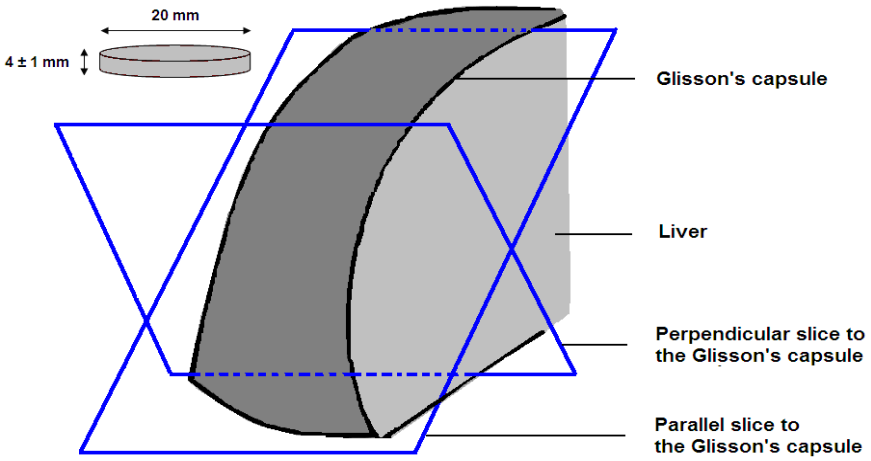


Fig. 4 – Liver Sample Size and Cutting Directions.

The experiments of oscillation in small deformations (linear domain) have been carried out at room temperature (22°C) using a stress-controlled rheometer (AR2000, TA-Instruments, New Castle, DE) in its plane configuration. The diameter of the plain geometry was 20 mm.

The top plate of the rheometer was lowered until it contacted the upper surface of the specimen. The liver tissue samples were placed between the parallel plates of the rheometer, to which sandpaper was glued to avoid slippage between the sample and the plates and to allow better gripping of the sample. While rotating the lower plate, the torque and normal force on the upper plate were recorded and linked to the dynamic moduli (Macosko, 1994). The influence of sandpaper and the glue has been the subject of an independent study. The results showed no influence, according to Hrapko et al. (Hrapko et al., 2007; 2008). To minimize the effects of inertia (Baravian, 1998a; 1998b) and the application of a non-uniform shear strain, a constant sample thickness and a low frequency range (0.1-4 Hz) were used. The samples had the same diameter as the plates (20 mm). A precompression of 5 mN was also imposed on the sample to ensure a perfect contact between the sample and the plates. Then, it relaxes due to the viscoelastic properties of the material. Samples were not preconditioned as it was done in other studies (Liu and Bilston, 2000; Nasseri et al., 2002). Preconditioning has the effect of increasing the repeatability of *in vitro* measurements. It establishes an initial "standard" condition for the tissue and mobile fluid is redistributed through the tissue. However, these two aspects do not reflect the reality during *in vivo* measurements to the authors' opinion.

First strain sweep oscillation tests were performed in order to define the material linear behaviour domain. The samples were subjected to a sinusoidal deformation at a fixed frequency of 1 Hz. The strain amplitude was increased incrementally from 0.01 to 20%.

Then, frequency sweep oscillations were investigated. Dynamic moduli were measured as a function of frequency. Frequency was increased from 0.1 to 4 Hz. The strain was fixed at 0.1% which is within the linear viscoelastic region.

Experiments for each test (strain and frequency sweep) were repeated 4 times for each animal and the results averaged to assess the error measurement. When data distributions allow it, a statistical analysis was done with a Kruskal-Wallis test (Kruskal and Wallis, 1952). The Kruskal-Wallis test compares the sample median of different data groups and gives a p-value which defines the probability that the compared groups may be different. A p-value near to zero suggests that at least one sample median is significantly different from the others. It is common to declare a result significant if the p-value is less than 0.05.

RESULTS

IN VITRO DMA TESTS RESULTS

Influences of sample sites, cutting direction and *post mortem* time on viscoelastic properties obtained by DMA are presented first as DMA results are often seen as reference values.

In order to study the liver homogeneity, samples were taken from the left and right lateral lobes. Shear modulus values, defined as the magnitude of the complex dynamic modulus, of about 1 kPa obtained with the different sample sites were in agreement. Indeed, the statistical analysis by Kraskal-Wallis gave a p-value equal to 0.9 for the shear modulus *G*. Thus, no significant difference between the shear modulus obtained with the left or right lateral lobes was underlined.

Samples were then cut parallel and perpendicular to the Glisson's capsule to investigate the liver anisotropy. The statistical analysis by Kraskal-Wallis gave a p-value equal to 0.1 and 0.2 for the storage and the loss modulus, respectively. No significant difference between the dynamic moduli measured in parallel or perpendicular liver slices to the capsule of Glisson was shown.

To study the *post mortem* time on tissue properties, two groups of samples were tested within 6h and between 18h and 24h after the death of the animal.

For the short-time-after-death tests, the linear behaviour limit was within 0.8 and 1.8% while for the long-time-after death tests, it was between 5.6 and 6.3%. For higher strains, the viscoelastic properties changed significantly. Results show that the linear viscoelastic limit is higher at long *post mortem* time than at short *post mortem* time.

The shear modulus *G* decreased with the *post mortem* time as shown in Fig. 5. A significant change of about 41% was observed for the mean shear modulus in the frequency range 0.1-4 Hz when the *post mortem* time varies from 6 to 18 hours.

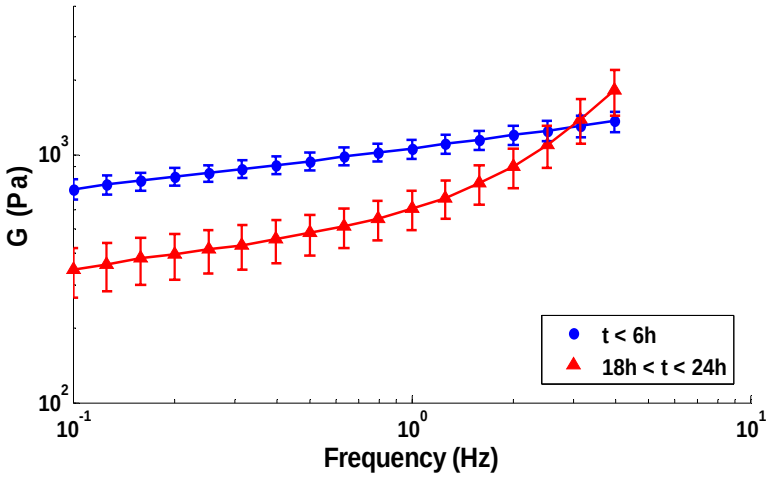


Fig. 5 - Effect of *post mortem* Time on Shear Modulus of Hepatic Tissue.

The viscoelastic behaviour of soft tissue can be modelled in the frequency range 0.1-4 Hz from DMA results. The shear behaviour was simulated by a generalized Maxwell model with two modes of relaxation from the *in vitro* experimental results obtained by DMA. The implemented law was expressed as a relaxation modulus:

$$G(t) = G_\infty + G_1 e^{-\beta_1 t} + G_2 e^{-\beta_2 t} \quad (2)$$

with G_∞ the equilibrium modulus, G_1 and G_2 the relaxation moduli, β_1 and β_2 the decay constants. The parameters of the model were determined by fitting the experimental data of the mean *in vitro* dynamic moduli G' and G'' with the model. G' and G'' are linked to the relaxation modulus by the general linear viscoelastic model. More details are given in Macosko (1994). Results are presented Fig. 6. Model parameters are given in Table 1. G_∞, G_1, and G_2 are given in Pa and β_1, and β_2 in s^{-1}.

Table 1. Values of the Mechanical Parameters Identified from the *In Vitro* DMA Tests.

G_∞ (Pa)	G_1(Pa)	β_1 (s^{-1})	G_2(Pa)	β_2(s^{-1})
337	169	1	290	10

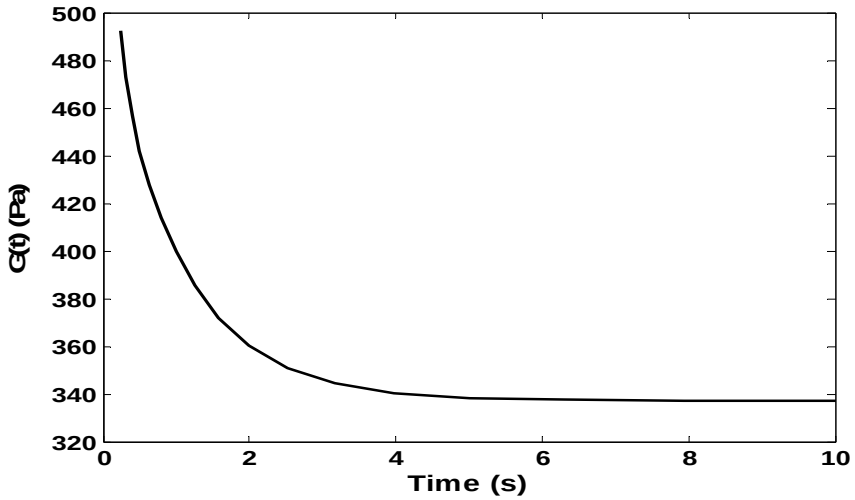


Fig. 6 - Reconstruction of the *in Vitro* Relaxation Modulus in the Frequency Range 0.1-4 Hz from a Two-Mode Relaxation Model.

COMPARISON OF *IN VITRO* TRANSIENT ELASTOGRAPHY AND DYNAMIC MECHANICAL ANALYSIS RESULTS

In vitro shear modulus results obtained by TE and by DMA were compared. For TE, the *in vitro* shear modulus was obtained at a frequency of 50 Hz and for DMA in the frequency range from 0.1 to 4 Hz. A direct and quantitative comparison can therefore not be made, the frequency ranges of both measurement techniques being different. However, as shown in Fig. 7, results obtained with both techniques suggest that they are of the same order of magnitude. As the *in vitro* shear modulus increased with the frequency, these results validate the TE-based technique.

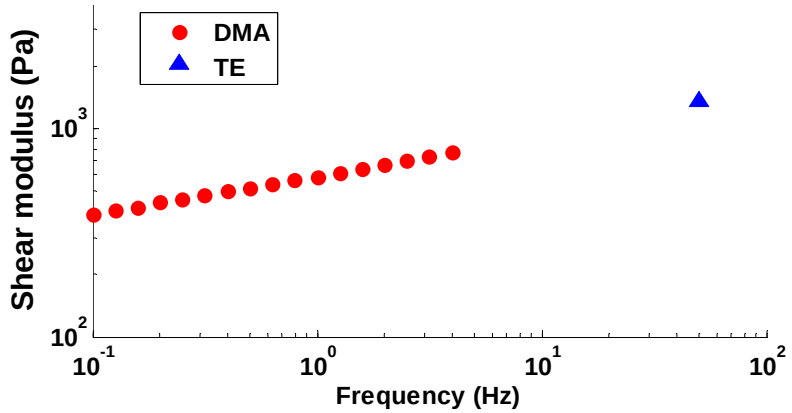


Fig. 7 – *In Vitro* Shear Modulus in the Frequency Range 0.1-50 Hz Obtained by Transient Elastography (TE) and by Dynamic Mechanical Analysis (DMA).

COMPARISON OF *IN VIVO* AND *IN VITRO* TRANSIENT ELASTOGRAPHY RESULTS

The *in vivo* shear modulus obtained in TE at a 50 Hz frequency ranged from 1.2 ± 0.8 (interquartile range) kPa to 2.3 ± 0.5 kPa for the 5 animals tested. An example of spatial-temporal strain map is given in Fig. 2b. The measured shear modulus was 1.7 kPa in that case.

The *in vitro* experiments have been carried out at long *post mortem* time after hepatectomy. The resulting *in vitro* shear modulus was found between 0.9 ± 0.1 kPa and 1.4 ± 0.3 kPa.

DISCUSSION

INFLUENCE OF TESTING CONDITIONS OF DYNAMIC MECHANICAL ANALYSIS

Knowledge of the linear behaviour of liver is needed to determine the level of deformation at which the material properties begin to depend on the applied deformation. Comparison of various results is based on this linear viscoelastic limit since below this limit the viscoelastic functions can be directly compared and beyond it the level of deformation has to be considered. In our study, this linear viscoelastic limit is equal to approximately 1% strain. This result is in agreement with low strain limits (of the order of 0.2%) reported in the study of Liu and Bilston (2000) for liver tissue.

The TE assumptions on which the shear modulus calculation (Eq. 1) is based can be questioned: linearity, pure elasticity, homogeneity, and isotropy. These hypotheses may appear as restrictive. However, the study of the sample sites and cutting direction influences in DMA show that they are negligible. Indeed, no significant difference between the dynamic moduli G' and G'' measured in the different sample sites or in different cutting directions was shown. Furthermore, due to the low values of the loss modulus G'' obtained by DMA, the shear modulus G is principally given by its real part G' (Appendix), the predominance of the elastic behaviour of the liver compared to its viscous one is confirmed. The mean viscous damping $\tan \delta$ was equal to 0.2.

Finally it is essential to verify the *in vitro/in vivo* validity of the shear modulus before focusing on viscous components of the tissue.

COMPARISON OF *IN VITRO* SHEAR MODULUS OBTAINED BY BOTH TRANSIENT ELASTOGRAPHY AND DYNAMIC MECHANICAL ANALYSIS

DMA technique is limited to the *in vitro* characterization of soft tissues. However, few *in vivo* techniques can measure the viscous behaviour of tissues independently. Thus, DMA method is still widely used for determining the viscoelastic properties of soft tissues. Furthermore, results show that *in vitro* shear modulus measured by TE and DMA are of the same order of magnitude. Therefore, data obtained by DMA can be considered as a first approach and be integrated in a first approximation for future experiments and models for the simulation of the shear response of soft tissues. These results validate the *in vitro* TE-based method.

For impact biomechanics, it would have been interesting to study a larger frequency range (up to tenth of kPa). However, in DMA, it was not possible to carry out tests at higher frequency due to inertia effects imposed by the rheometer used in the study. A maximum frequency of 4 Hz was used for DMA experiments. In TE, the device used in the study was not adapted to carry out experiment at higher frequency (> 50 Hz). Furthermore, the maximum frequency that can be studied depends on the viscoelastic properties of the liver (attenuation, elasticity, etc). In TE, some work is in progress to perform the experiments at higher frequency.

COMPARISON OF *IN VIVO*/*IN VITRO* SHEAR MODULUS BY TRANSIENT ELASTOGRAPHY

The *in vitro* shear modulus values obtained by TE were lower than those found *in vivo* (the mean difference was equal to 66%). The *post mortem* time effect is probably one of the reasons of the observed changes in mechanical properties. It is partly due to tissue degradation. The fact that the tissue is no longer in its natural environment may also explain this difference in elasticity. Indeed, by

taking the organ from its environment, the boundary conditions are changed (precompression of the liver due to other organs or blood pressure). In addition, the liver is no longer perfused. This last point may be correlated to the work done by Kerdok et al. in 2005 who studied the effect of blood on the viscoelastic properties of the liver and showed that properties changed with the perfusion of the liver. However, this study was conducted *in vitro* on removed liver. DMA results are also in agreement with this observation. G_{DMA} is lower at long *post mortem* time than at short *post mortem* time.

IN VIVO SHEAR MODULUS BY TRANSIENT ELASTOGRAPHY

The average *in vivo* shear modulus obtained by TE (G_{TE} = 2.0 ± 0.5 kPa) was consistent with other results from the literature. Indeed, Huwart et al. (2006) determined a shear modulus of 2.3 kPa with magnetic resonance elastography experiments on human liver. It is also comparable to the value of 1.8 ± 0.5 kPa found by Roulot et al. (2008) with its TE results on human liver. In MRE, Klatt et al. (2006) found a shear modulus of 2.3 ± 0.4 kPa on human healthy liver at 51 Hz. Therefore, the elasticity of porcine and human liver is comparable. This is not surprising because apart from the geometry (number of lobes and dimensions), structure and functions of the liver are the same.

Results confirm also that transient elastography technique can be used to measure *in vivo* liver elastic properties.

CONCLUSION

In this study, tests were performed on the liver of 5 female pigs. Each organ has been tested successively *in vivo* by transient elastography, *in vitro* by transient elastography and at least *in vitro* by dynamic mechanical analysis.

The preliminary results obtained in this study are promising. In addition to the measurement of *in vivo* liver tissue mechanical properties, this study demonstrates the efficiency of the TE-based method to determine *in vivo* liver properties. In addition to the validation of the *in vitro* use of the TE experimental device, this study shows a significant decrease of the liver tissue elasticity obtained by TE between *in vivo* and *in vitro* configurations. Finally, at 50 Hz, the mean liver shear modulus is about 2.0 ± 0.5 kPa *in vivo* and 1.2 ± 0.4 kPa *in vitro*. This study has also confirmed the homogeneity and the isotropy as well as the *post-mortem* time dependence of the liver tissue.

Future work will extend this study to a more significant numbers of animals. Besides, TE does not measure the viscosity independently. Measuring the viscous property of a material in TE is of a great interest. However, this study has quantitatively demonstrated the validity of the elastic assumption in TE in a first approximation. Furthermore, this work presents the possibility to use a clinical device, dedicated for the diagnosis of liver fibrosis, as an efficient biomechanics tool not only for medicine purposes but also for impact biomechanics with more realistic numerical crash investigations.

ACKNOWLEDGEMENTS

The authors would like to acknowledge the surgical staff of the IRCAD for their contributions and availability. This work was supported in part by the European project PASSPORT.

APPENDIX: Dynamic mechanical analysis results

Results show the range of measured values in the studied DMA frequency interval 0.1 – 4 Hz. G', G'', G correspond to the storage, loss and shear moduli in Pa. tan δ is the viscous damping. The value between parentheses (SD) represents the standard deviation. γ_{lim} is the strain limit of the linear viscoelastic domain in percentage.

Hypothesis	Quantities	Values
Liver elasticity	**G' (SD)**	317,0 - 1302,2 (95,3 - 296,5)
	G'' (SD)	74,1 - 190,7 (27,8 - 59,4)
	G (SD)	325,7 - 1323,7 (99,6 - 306,7)
	tan δ	0.23 – 0.24 (0.27 – 0.31)
Liver homogeneity	**Right lobe: G (SD)**	337,1- 690,6 (72,0 - 171,8)
	Left lobe: G (SD)	371,3 - 677,0 (94,2 - 175,0)
Liver anisotropy	**Parallel slice: G' (SD)**	319,1 -585,1 (33,8 - 91,5)
	Parallel slice: G'' (SD)	67,8 - 157,5 (8,4 - 25,1)
	Perpendicular slice: G' (SD)	345,4 – 653,0 (118,8 - 213,8)
	Perpendicular slice: G'' (SD)	73,3 - 177,5 (17,8 - 47,3)
***Post mortem* time effect on shear modulus**	**Short time: G (SD)**	728.4 - 1367.2 (68,1 - 131,9)
	Long time: G (SD)	343,2 - 1822,0 (79,5 - 384,8)
***Post mortem* time effect on the linear viscoelastic strain limit**	**Short time: γ_{lim}**	0.8 ± 0.3 - 1.8 ± 0.2
	Long time: γ_{lim}	5.6 ± 2.4 - 6.3 ± 0.0

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