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Journal of Biomechanics 40 (2007) 1960–1967

JOURNAL OF BIOMECHANICS

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Mechanical properties of soft human tissues under dynamic loading

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Accepted 15 September 2006

Abstract

The dynamic response of soft human tissues in hydrostatic compression and simple shear is studied using the Kolsky bar technique. We have made modifications to the technique that allow loading of a soft tissue specimen in hydrostatic compression or simple shear. The dynamic response of human tissues (from stomach, heart, liver, and lung of cadavers) is obtained, and analyzed to provide measures of dynamic bulk modulus and shear response for each tissue type. The dynamic bulk response of these tissues is easily described by a linear fit for the bulk modulus in this pressure range, whereas the dynamic shearing response of these tissues is strongly non-linear, showing a near exponential growth of the shear stress.

Keywords: Mechanical properties; Soft tissues; Bulk modulus; Shear response; Kolsky bar; High strain rate

1. Introduction and background

There are a number of situations in which the human body is subjected to impact loading, including automobile crashes, falls, blast effects, and gunshots. Consider, for example, a law enforcement officer wearing a bulletproof vest: even with the vest on, the result of a gunshot to the torso is the propagation of stress waves through various organs, bones and interconnecting tissues, and subsequent damage. Minimization of injury or more effective design of the vest requires a quantitative model of the response of the human torso to impact loading (e.g., Grimal et al., 2005). However, such a model requires knowledge of the dynamic mechanical properties of the soft tissues, and such properties are scarce in the literature. Similar issues arise for estimates of injury or the design of effective interventions in other problems involving the human body subjected to impact.

The majority of the experiments conducted on human tissues so far have been limited to "quasistatic" load conditions (Yamada, 1970; Carter et al., 2001; Egorov et al., 2002). Most of the existing dynamic data in the literature has been obtained on porcine tissues (Snedeker et al., 2005b), and some on bovine tissues (Mavrilas et al., 2005). Even when dynamic data are provided in the literature, this is typically limited to strain rates on the order of $10\,s^{-1}$, significantly lower than the strain rates that are estimated by models of impact on the body (Roberts et al., 2006). Miller and Chinzei (1997) performed unconfined uniaxial compression tests on swine brain tissues over a wide range of strain rates, with a maximum strain rate $\sim 1\,s^{-1}$. In recent work, Snedeker et al. (2005a) make a comparison between human and porcine tissues, but only the porcine tissues were tested in impact loading; the human tissues were tested in quasistatic conditions.

The lack of data on the dynamic behavior of human tissues is primarily because of the difficulty involved in obtaining human tissues in sufficient quantities and secondarily because there are no standard testing techniques for dynamic loading of soft tissues. Arbogast et al. (1997) suggested an oscillatory testing device for testing soft biological tissues in shear, but their device is limited to strains of up to 20% and frequencies up to 200 Hz, which is not sufficient for studying ballistic impacts. Others have used oscillatory shear to explore larger frequencies, but typically compromise by examining only small strains (the linear approximation). Further, techniques that involve

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0021-9290/$ - see front matter © 2006 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jbiomech.2006.09.021

oscillatory loading may examine different mechanisms than are evident in the single high-rate loading pulse typically generated during an impact, because of lower strains, the lack of development of fiber alignment, and lower impulse. Much of the existing data on the mechanical properties of human tissues is for tissues from the brain (Nicolle et al., 2004), kidney stomach (Egorov et al., 2002), and ligaments (Weiss et al., 2002), although most of these are for relatively low strain rates.

The stress state developed is also of interest. Most experiments on soft tissues have been conducted either in uniaxial tension or uniaxial compression (Miller, 2005) or drained and "undrained" compression (Kwan et al., 1990). However, as far as we know there are no data on the confined compression of human tissues except at low stresses and low rates or at high frequencies (ultrasonics). It is not a priori reasonable to assume that the bulk modulus of tissues will be independent of the rate of loading or applied pressure. The distortional or shear response is just as important, and is often addressed by performing the oscillatory shear tests previously discussed. We address both of these major modes of deformation in this paper for dynamic loading.

Our approach draws on a wealth of experience in high-strain-rate testing that has been accumulated in the literature on non-biological materials. Variations of the Kolsky bar approach have been used to investigate the dynamic response of comparatively stiff materials in uniaxial compression (Jia and Ramesh, 2004), tension (Li et al., 2004a, b), and torsion (Li et al., 2004a, b). In this paper, we use modifications of this technique that allow us to measure the bulk modulus and the shearing response of soft human tissues under dynamic loading.

2. Tissue sample preparation and handling

Human tissue was obtained from the Brain and Tissue Bank for developmental disorders at the University of Maryland, Baltimore, Maryland. The post-mortem human subjects (PMHS) were all between the ages of 18 and 30, with causes of death that did not affect the organs targeted for this study. Fresh frozen specimens were collected from the heart, liver, lung, and stomach.

The heart specimens were harvested from the left ventricle. The liver specimens were taken from the left and right lobe. The lung sections were removed from the upper lobe and the stomach specimens were taken from the stomach body just below the cardiac notch. No information on tissue orientation was retained, although tissues in general and heart tissues in particular may have anisotropic characteristics. All tissue samples were harvested at the Brain and Tissue Bank for Developmental Disorders and packed in dry ice ($-78.5\,^{\circ}C$) for immediate shipment to JHU/APL. The specimens were then placed and held in a freezer ($-80\,^{\circ}C$) until test preparation.

Bulk modulus samples were prepared by using a cork-boring tool with an inner diameter of 12.7 mm to cut cylinders of tissue. Then, a scalpel was used to slice sheets of tissue 1–2 mm thick while the specimen was still partially frozen. The dimensions of the disks of tissue were measured and then they were individually wrapped in cellophane and labeled for tissue type and PMHS identifier. The wrapped samples were then kept in freezer storage until the Kolsky bar tests were to be performed.

The shear samples were prepared by first cutting rectangular sections of 9×20 mm cross-section from the tissue specimen. Then, the thin (1–2 mm) samples were sliced from the ends of these rectangular sections. Two tissue rectangles of matching thickness were then glued to aluminum plates using moisture-curing cyanoacrylate glue. The assembled test samples were individually wrapped in cellophane, marked for tissue type and PMHS identifier, and frozen until the day the dynamic tests were performed. The samples remained in the cellophane for as long as possible to minimize dehydration. Prior to testing, the specimens were transported to the laboratory encased in dry ice. Once the test preparations were complete, each specimen was removed from the ice and placed in the Kolsky bar test fixture, and allowed to thaw prior to testing. Typical elapsed time between thawing of the specimen and testing was 2 min, with no specimen having been thawed for more than 5 min prior to testing. Due to their small size, it can reasonably be assumed that the tissue samples were at room temperature ($\sim 20\,^{\circ}C$) at the time of testing. Sample dimensions were measured just prior to testing for each sample.

3. Experimental techniques

Our interest is in measuring the dynamic bulk and shear response of soft human tissues. One of the authors has previously developed a technique for the measurement of the volumetric response of liquids and other soft materials at high loading rates (Ramesh, 1991), and we incorporate this technique to test human tissues using the Kolsky bar. In addition, a new modification to the Kolsky bar experimental method has been developed which allows loading of tissue specimens in simple shear and provides the shear stress versus shear strain response of the tissues. We describe each of these techniques very briefly in this section: the detailed description will be presented in another paper.

3.1. Dynamic compressibility experiment

The apparatus (Feng and Ramesh, 1993) consists of a projectile, an "input" bar, and an "output" bar, all made of an aluminum alloy, and a specimen confinement cell that is held between the two bars (Fig. 1a). The input and the output bars also serve as pistons that move with a close sliding fit within the confinement cell. The projectile is fired at the input bar with a high velocity (1–50 m/s), developed

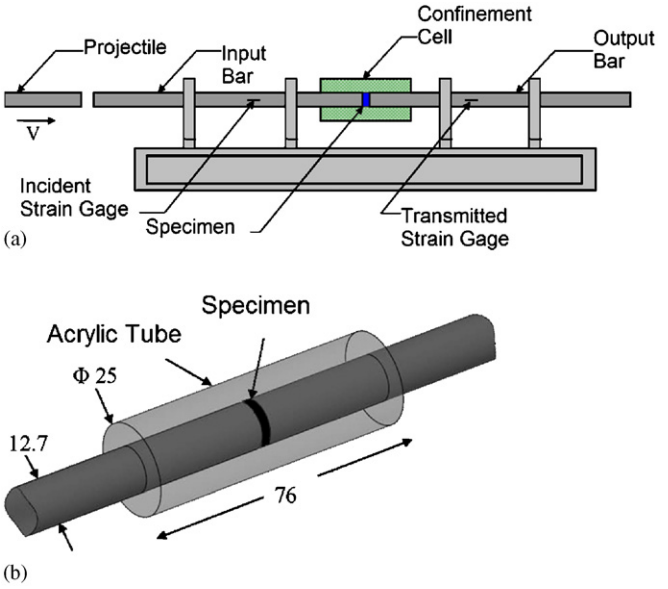

Fig. 1. (a) Schematic of the setup for measuring the bulk response of the tissues. (b) Enlarged view of the confinement cell for dynamic confined compression. All dimensions in mm.

using a gas gun arrangement (not shown). The resulting compressive pulse travels through the input bar and into the specimen. Part of this ''incident'' pulse is reflected from the specimen–bar interface and the rest is transmitted through the specimen and into the output bar. The incident, reflected, and transmitted pulses are recorded by strain gages on the input and output bars. The measured pulses can be used to calculate the axial stress in the sample as well as the velocity histories of the boundaries of the sample (Feng and Ramesh, 1993). The critical feature of this experimental technique is that the confinement cell prevents radial expansion of the tissue, resulting in the development of a uniaxial strain state rather than a uniaxial stress state. Thus, the only deformations are along the axis of the specimen, and measurement of these deformations provides a measure of the volume change in the sample.

The stress state in the sample builds up to a hydrostatic stress state approximating the measured axial stress, because the finite (non-zero) compressibility of the sample material will ensure that the volumetric strain cannot increase indefinitely. Each test therefore provides one data point: a measured volumetric strain corresponding to a measured pressure, see Ramesh (1991).

By performing such experiments over a range of impact velocities (thus varying the pressure) and measuring the final pressures and volume changes, one can construct a plot of the pressure–volume behavior of the material. In the experiments performed here, the stress level rises in the sample and reaches the final pressure over approximately 100 μs, and what is measured is a dynamic bulk response at an effective nominal strain rate (defined by final strain over risetime) of $\sim 10^3\,s^{-1}$.

3.2. Dynamic shear response measurement

For the shear measurements, we use a double lap shear configuration, implemented within a compression Kolsky bar setup as in Fig. 2a, which allows us to load tissue specimens in simple shear. In place of the confinement cell described above, we now have the special double lap shear fixture shown in Fig. 2b. Two specimens are now required for each test for symmetry, and the double lap configuration cancels bending stresses. The two pieces of tissue have one surface of each glued to a common aluminum plate and the other surface glued to a thinner aluminum plate, which is in turn glued to the fixture. The gap height of the fixture is used to determine the average height of the samples. The central plate is translated along the axis as a result of the compressive incident stress pulse, and this develops symmetric shear in the two specimens. The force transmitted into the output bar is now directly the shear force on the two specimens. We obtain an accurate measurement of the small transmitted signal using a piezoelectric crystal (quartz) mounted between a short intermediate bar and the output bar (Cohen et al., 1999). This quartz gage, when coupled to a charge amplifier, generates a voltage signal that can be used to determine the transmitted load F_t and thus the shear stress $\tau_s(t)$ in each specimen. The shearing deformations are measured by direct measurement of the shear displacement of the specimen top and bottom surfaces using the non-contact optical technique first developed by Ramesh and Kelkar (1995). This shear displacement history $\delta(t)$ is used to compute the shear strain history $\gamma(t) = \tan^{-1}(\delta(t)/a)$ where a is the sample thickness. The shear strain rate is the rate of change of $\gamma(t)$—typical shear strain rates that we achieve are of the order of $1 \times 10^3\,s^{-1}$. Correlating the shear stress and shear strain histories, we obtain the shear stress as a function of the shear strain.

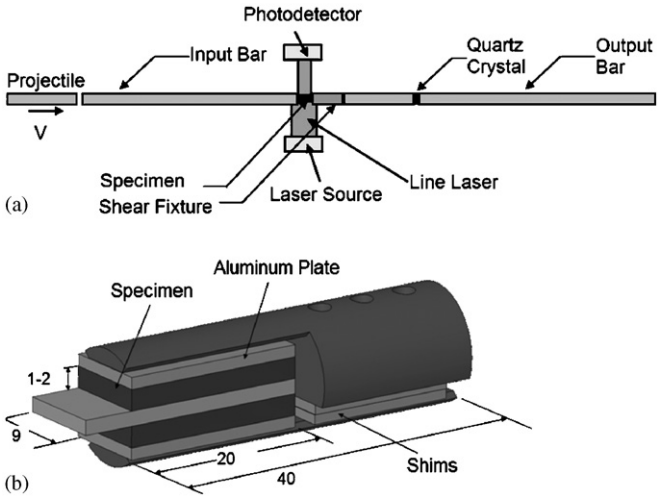

Fig. 2. (a). Schematic of the setup for measuring the shear response of the tissues. (b) Enlarged view of the double lap shear fixture with the specimen. All dimensions in mm.

3.3. Equilibration of the stress under dynamic loading

The experimental techniques described above both assume that the stress state within the sample will be uniform (often referred to as the equilibration of the stress) after an initial risetime, and it is important to ensure that this is the case in order to have confidence in the data. This issue is a well-recognized one in the literature on the dynamic behavior of materials (e.g., Song and Chen, 2004). We address the equilibration question for these experiments in the online supporting material.

4. Dynamic bulk response of human tissues

Thin discs (thickness ~1–2 mm) of tissues from the human heart, liver, lung, and stomach were tested in confined compression at nominal strain rates ranging from 300 to 5000 s^{-1}. The duration of the compressive pulse was either 100 or 140 μs, depending on the length of the projectile used. The measured data are presented in Fig. 3 in terms of the pressure–volumetric strain behavior of each tissue. All of the fully equilibrated tests are presented in the data as solid squares, while the results of tests in which the stress equilibrates but the strain never saturates (that is, it continues to rise) are presented as hollow triangles. The slope of the linear fit (which is fit only to those tests where saturation of the strain was also achieved) gives us the bulk modulus of the tissues at these strain rates. Note that in the case of the heart tissue, there was only one such test—it appears that the tissues from the heart generally develop either a damage or extrusion mechanism. Aside from the heart tissue, we see that the bulk modulus estimates would change very little if we included the unsaturated strain data

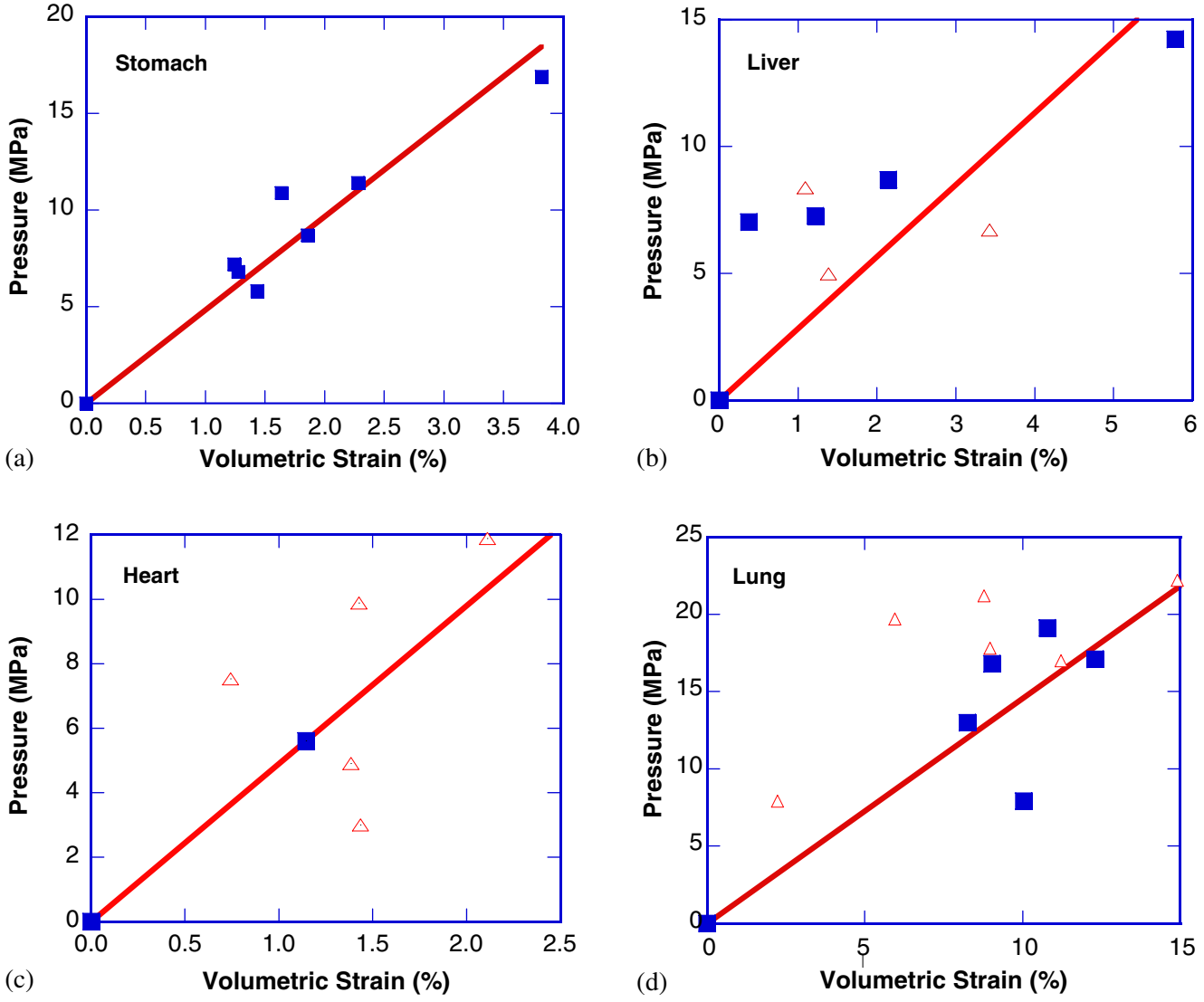

Fig. 3. Results from the dynamic compressibility experiment. Pressure–volumetric strain data for (a) stomach, (b) liver, (c) heart, and (d) lung tissues. The hollow triangles represent tests in which the stress equilibrates but the strain does not saturate; linear fits are only to the fully equilibrated and saturated data (the solid squares). The corresponding strain rates are shown in Table 1.

in the fit. For the heart, however, the estimated bulk modulus using the one fully saturated test is 0.38 GPa, while if we use all of the test results we would estimate a bulk modulus of 0.25 GPa. The smallest amount of scatter was obtained with the stomach tissues. Given the general range and scatter of the data, there is no reason to assume anything other than a linear bulk response and thus a simple (dynamic) bulk modulus over this range. Note, however, that the instantaneous bulk modulus at very low pressure obtained by ultrasound and acoustic microscopy techniques may well be different because of the lack of structural relaxation with such techniques. The results for all of the human tissues are summarized in Table 1 in terms of the effective dynamic bulk modulus given by the linear fit. The stomach tissues were the stiffest of the four tissues in terms of the bulk response, while the lung tissues were observed to be the softest. The dynamic bulk moduli vary from 0.15 to 0.5 GPa. It appears that incompressibility is not a good assumption for these human tissues at these very high pressures of the order of several MPa, which may be developed in a high-velocity impact. The effective bulk moduli that we estimate are significantly smaller than the instantaneous moduli reported in the ultrasound and acoustic microscopy literature (e.g., Masugata et al. (1999) obtain ~2.9 GPa for myocardial tissue), but that literature is explicitly dealing with very low pressures and very small deformations because of the techniques involved. It appears that at these higher pressures corresponding to impact events, the effective (or tangent) bulk modulus is about an order of magnitude lower. The deformation of the confining cylinder has been computed using a full finite-element analysis (e.g., see Online Fig. 1), and the data shown in Fig. 3 have been corrected for the small deformations of the cylinder, and so this latter effect is not the source of this discrepancy. One possible cause for the lower moduli that we measure is that (in contrast to ultrasonics and acoustic microscopy) our experimental timeframes at high pressure allow for true rearrangement of molecules and fibers during the experiment, resulting in more relaxed structures. In ultrasonic and acoustic measures, the high pressures are retained for extremely short times; in that sense, our data may be more representative of true material behavior under impact loading durations.

5. Dynamic shear response of human tissues

The dynamic shearing experiment gives us direct measures of the shear stresses and shearing displacements in the tissues using the quartz gage and the laser-based displacement sensor, respectively. Making the assumption of the uniformity of the stress state (as discussed in online material, this should be valid after the first 160 μs or so), we can make plots of the shear stress in the tissue versus the shear strain developed in the tissue. Such plots are presented in Fig. 4 for all four tissues over the range of strain rates evaluated. Note how large the shear strains are in these plots—such large shear strains can never be attained in typical oscillatory shear experiments.

All of the shear stress versus shear strain curves have the same shape: little resistance is offered to shearing at low shear strains, but at larger strain the material stiffens dramatically. Such behaviors have been observed often in low strain rate testing of human tissues (Fung, 1993; Yamada, 1970) that have been tested in tension. Such stiffening is sometimes attributed to progressive alignment of the fibers in the tissue with the loading direction, and is to be expected also in shear where the principal axes of stretch rotate during the deformation. The initial part of these curves is sometimes referred to as the toe region. The apparent increase in the length of the toe region with increasing strain rates is likely to be an artifact of this testing technique (since a finite time of the order of 160 μs is required for the test to become a valid test, and larger apparent strains will be accumulated during this time for higher strain rate tests). Note that it is not necessary that equilibration be established during the risetime of the incident pulse—unlike in the traditional compression Kolsky bar technique, we only use data obtained after sufficient time that equilibration is achieved (details of this process are presented in a forthcoming techniques publication). The very low shear stress data in the toe region are therefore not solely a result of material behavior. Some of the curves have flat portions at the ends, which are testing artifacts also discussed in the forthcoming techniques

Table 1
Summary of test results on high rate behavior of human tissues

Type of test		Tissue type			
		Stomach	Liver	Heart	Lung
Bulk tests	Applied strain rate (s^{-1})	350–2900	400–1000	300–1200	400–7700
	Bulk modulus (GPa)	0.48	0.28	0.49–0.25[a]	0.15
Shear tests	Applied strain rate (s^{-1})	300–2200	280–1800	200–2800	400–2300
	Tangent shear modulus (kPa)	8–45	37–340	60–148	10–54

[a]See text.

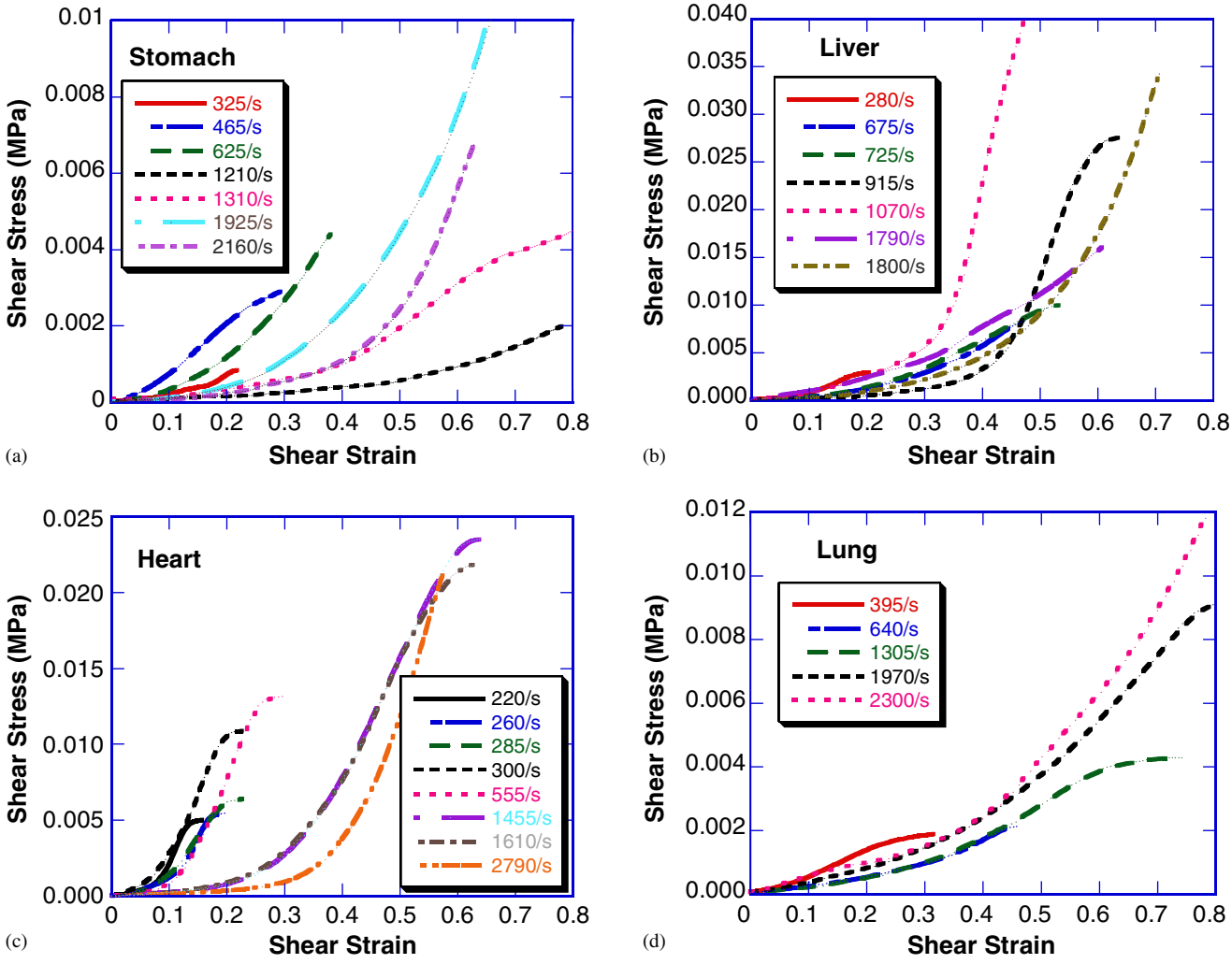

Fig. 4. High strain rate shear stress versus shear strain curves obtained on (a) stomach, (b) liver, (c) heart and (d) lung tissues. Note the rapid increase in the shear stress at large shear strains.

paper. Uniformity of deformation is being addressed through high-speed photography in current work.

The variation in behavior of these tissues from sample to sample may well be a result of the variations in fiber orientation within the set of samples. That said, there are some special features of each tissue. For example, the largest shear stresses are developed within the liver tissues, and the least within the lung tissues (not surprising, given the morphology of lung tissue). The results for the heart tissue are again unusual, in that there appears to be a clear grouping of the stress–strain curves into two groups corresponding to strain rates above and below about $10^3\,s^{-1}$. We have no explanation for this observation at this time.

Unlike the bulk response results, it is apparent from Fig. 4 that the behavior of these tissues in dynamic shearing is non-linear. However, if a tangent modulus were to be fit to the late-time near-linear stages of each stress–strain curve (ignoring the turnover at the end of the curves) by computing a best-fit straight line through a window of data points and extracting the maximum slope, tangent shearing moduli in the range of 8–340 kPa are obtained. The results for all of the human tissues are summarized in Table 1 in terms of the peak tangent shear modulus, obtained as described above. The liver tissue was found to be the stiffest whereas lung was the softest in terms of the shearing response. Exponential behaviors of the type suggested by Fung (1993) are perhaps best fit to these data, and it is possible to construct a full finite deformation model for the behavior utilizing strain energy functions and then fitting the shear and volumetric responses. We will pursue that effort in subsequent work.

6. Discussion and conclusions

No attempt was made to record the orientation of the final tissue samples with respect to the organs, although that would be a reasonable goal for a follow-on study. The primary goal of this study was to determine the average mechanical responses of the heart, liver, lung, and stomach

at high strain rate for input into a numerical model, and this numerical model is focused on the dynamic effects and so treats each organ as having homogeneous, isotropic mechanical properties. Thus while it is recognized that these tissues do have anisotropic characteristics, examining the effects of tissue orientation on the measured dynamic mechanical properties is beyond the scope of this investigation. It is likely that these factors contribute to the total scatter in the data. Quantifying the scatter due to specimen variations alone is not possible given our sample size and the inherent variability in tissue properties themselves due to location, orientation, and PMHS.

The modifications to the compression Kolsky bar system that we have presented allow for bulk modulus and shear response measurements on soft specimens at high rates of strain. The response of human organ tissues from the heart, stomach, liver, and lung was measured under dynamic confined compression and dynamic simple shear. To our knowledge, these are the first such results on the response of human tissues at impact strain rates. Explicit computational simulations of the shearing technique have been conducted to account for finite deformations and viscoelasticity. The key results are the following.

The behavior of these tissues under dynamic confined compression can be represented by an approximately linear relationship between the pressure and the volumetric strain. The compressibility of these tissues is considerable at these pressures (MPa) in terms of the volumetric strain observed, with the stomach being the stiffest tissue and the lung the least stiff. The effective bulk moduli that we estimate are about an order of magnitude smaller than those reported in the ultrasound and acoustic microscopy literature, but that literature is explicitly dealing with very low pressures and very small deformations.

In dynamic shearing, the behavior of these tissues is not linear elastic. The shear stress–shear strain curves for these tissues exhibit a toe region, followed by a rapid growth of the shear stress with moderate increases in shear strain. Exponential models such as those observed at lower strain rates on other tissues may continue to be appropriate. However, if a tangent modulus were to be fit to the final quasilinear stages of each stress–strain curve, tangent shearing moduli in the range of 8–340 kPa would be obtained, with liver being the stiffest tissue and lung the least stiff.

Acknowledgments

The authors express their gratitude to the Office of Naval Research for funding this work through Contract No. N00014-04-PD-2-0004(1). We thank Dr. Tonia Jiao for early development efforts and Dr. S.P. Joshi for the finite element calculations on the confining cylinders. The assistance of Tristan Flanzer in the laboratory is also gratefully acknowledged.

Appendix A. Supplementary materials

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.biomech.2006.09.021

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