INJURY BIOMECHANICS RESEARCH
Proceedings of the Thirty-Seventh International Workshop

A Modified Split Hopkinson Pressure Bar Technique for Characterizing Soft Materials and Biological Tissue under Shear Loading

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ABSTRACT

Both modeling human body response to dynamic loading events and developing biofidelic human surrogate systems require accurate material property characterization over a range of loading rates for the various tissues that constitute the human body. This work describes a technique for measuring the shear properties of soft biomaterials at high rates of strain (100 to 1000 s^{-1}) using a modified split-Hopkinson pressure bar (SHPB) technique. Establishing a uniform state of stress in the sample is critical for high rate testing of this type. The technique presented includes direct experimental verification of the uniformity of the stress in the sample using piezoelectric quartz force gages on both the input and output sides of the shear specimen. Maintaining a uniform stress within the specimen from the start of the test requires input pulse shaping to ramp the incident loading pulse. Soft foam pulse shapers are used to control the loading when testing the response of compliant biomaterials that are subjected to peak applied loads of only ~3 N. Preliminary results from fresh and frozen porcine brain tissues and silicone-based biosimulant materials are presented. Future experiments will utilize the technique developed here to characterize both human biological tissue and candidate biosimulant materials and compare their dynamic mechanical properties.

INTRODUCTION

An understanding of the dynamic mechanical properties of biological tissues is required to study human body response to high-rate events including automotive crashes, ballistic impact (e.g. behind armor blunt trauma) and blast loads. Computational and experimental models of the human anatomy have been constructed to predict the risk and severity of injury to the human body due to these events. It is critical that these models properly simulate the response of human tissue to various load conditions. Thus, the material

properties of biological tissues and biological simulant (biosimulant) materials must be determined at high strain-rates. The tissue properties are employed in computational models whereas the biosimulant properties establish the fidelity of experimental surrogate systems with regard to human tissue. Organ tissues and biological simulants are viscoelastic materials with strain rate dependent properties. Many constitutive models describing viscoelastic materials require input of a material's bulk and shear moduli over a range of strain rates.

High strain rate materials characterization can be performed using a variety of techniques. Ultrasound allows for *in vivo* measurement, but is limited to very small strains in a very high frequency regime (MHz). Dynamic Mechanical Analysis (DMA) allows for testing over a range of frequencies, but is normally limited to approximately 200 Hz. Furthermore, the oscillatory nature of DMA experimentation may not be representative of high-rate impact scenarios. The Split Hopkinson Pressure Bar (SHPB) is not as suitable for obtaining small strain data, but it is capable of imparting large shear strains (up to ~1) and high strain rates (10^2-10^4 s^{-1}) with a single pulse. For these reasons, this method has been chosen for characterizing dynamic mechanical properties of biological tissues and simulants.

Although the SHPB has several advantages for dynamic characterization of materials, testing soft materials presents some challenges. Uniform stress distribution in the specimen is a fundamental requirement for conducting valid SHPB experiments. Due to the low wave speeds in soft materials, it is difficult to achieve equilibrium within the timescale of a dynamic test. Modifications that have been made to SHPB systems to accommodate soft materials include the use of low impedance polymeric bars (Zhao et al., 1997, Van Sligtenhorst et al., 2006) and quartz force gauges (Chen et al., 2000, Saraf et al., 2007, Saraf et al., 2007), pulse shaping (Chen et al., 2002, Song et al., 2007), and changes to specimen geometry, including reduced thickness (Dioh et al., 1993, Chen et al., 2002). Without modification to the traditional SHPB system, an early, rapid rise in the initial stress measurements of the specimen are often observed when testing soft materials (Chen et al., 2002, Song et al., 2007). In a traditional compression test, this spike is due to added stress from the specimen's radial inertia during the acceleration stage of axial deformation. The amplitude of the spike is related to acceleration (strain differential), specimen diameter and density. This spike, and the inertial effects it is indicative of, can be minimized via use of pulse shaping (Chen et al., 2002, Song et al., 2007) to slowly ramp the incident loading pulse. This leads to reduced strain acceleration during initial dynamic loading and generation of a nearly constant strain rate. Additionally, this causes an increase in rise time such that dynamic equilibrium can be achieved since the extra time allows for multiple reflections within the specimen during the onset of loading. As a result, significant deformation occurs after dynamic equilibrium is established. Identification of a suitable pulse shaper is critical for testing of soft human tissues and biosimulants. However, pulse shaping alone will not guarantee stress uniformity. Due to their low wave speeds, soft materials must be tested using thin specimens (Dioh et al., 1993). The principles of both pulse shaping and thin samples apply not only to traditional compression testing on the SHPB, but also to the shear configuration. It is crucial that stress uniformity be verified for each experiment when testing soft materials.

The objective of this study was to develop a modified SHPB technique for assessing high rate response of biological tissue and simulants under dynamic shear loading conditions. Specifically, the use of pulse shaping was explored and stress uniformity in the tissue and biological simulants was verified.

METHODS

Split Hopkinson Pressure Bar

A SHPB is composed of three bars: a projectile, an incident bar, and a transmitted bar, and a specimen is sandwiched between the incident and transmitted bars, as shown in the schematic in Figure 1. During a SHPB experiment, air pressure from a breech is used to drive a projectile to impact an incident bar, producing in it an elastic stress wave with a pulse duration that is long with respect to the sample (Meyers, 1994). The elastic wave travels through the incident bar and reaches the specimen, which is sandwiched between the incident and transmitted bars, and the amplitude of the wave induces deformation in the specimen. The level to which the breech is pressurized dictates the impact velocity, and subsequently the

stress imparted on the specimen. This technique relies on the assumptions that the bars are stressed within their elastic limits and uniform stresses act throughout the entire specimen.

Several modifications, including: double lap shear configuration, low impedance bars, long loading pulse duration, line laser direct strain measurement, quartz force gauges, and pulse shaping, were made to the traditional SHPB technique to allow for shear testing of soft materials. A schematic of the modified setup can be seen in Figure 1, and the modifications are described here. A double lap shear specimen configuration was utilized (Saraf et al., 2007, Saraf et al., 2007). This specimen configuration is commonly used for DMA shear testing (Cognard et al., 2005, Yang et al., 2007) and resists changes in specimen height, creating a state of simple, uniform shear. Specimens used in this work were 9 mm x 18 mm and ~1 mm thick, and were arranged in a layered structure between aluminum shims, with the center aluminum plate having a length ~5 mm longer than the opposing layers. Before the experiment begins, the input bar is moved to touch the center plate of the double lap shear specimen configuration. With this setup, axial impact by the incident bar causes the center plate to translate, which in turn translates only the specimen surfaces adhered to the center plate, thus imparting shear stress and strain on the specimen – there is no other path through which the force can travel from the incident bar to the transmitted bar. Schematics of the double lap shear configuration prior to and during testing can be seen in Figure 2(a) and (b), respectively, and a photograph of a silicone gel biosimulant in double lap shear configuration can be seen in Figure 2(c). A long loading pulse duration was achieved by using a projectile to incident bar length ratio > 1. Direct strain measurement was performed via a collimated line laser and photodetector (Ramesh and Kelkar, 1995) rather than integration of strain rate, as in traditional SHPB. Displacement was measured by tracking the length of the opening through which light was transmitted, which decreased as the specimen deformed. A schematic of the line laser setup can be seen in Figure 3. Piezoelectric quartz crystal gauges (Chen et al., 2002) were used to measure force. These are very sensitive to small forces (~ 5 mN), whereas traditional strain gauges are not sensitive enough to discern the low forces transmitted during tests of soft materials. One gauge was placed at the interface between the incident bar and specimen, and two gauges were placed in the transmitted bar to verify stress uniformity in the specimen, and for redundancy (Figure 1). Aluminum (7075-T651) bars were used to avoid reflections from the quartz crystal gauges due to acoustic impedance mismatch. The difference in the uniaxial stress acoustic impedance between X-cut quartz and 7075-T651 is only 1.5%. For comparison, the impedance of quartz is 62% less than that of maraging steel, which is a traditional SHPB material. The lower impedance of the aluminum also decreases the proportion of the incident wave that is reflected due to the impedance mismatch at the interface with the specimen (Song et al., 2007). Although most of the wave is still reflected when using aluminum bars, a portion is transmitted which is sufficient to induce shear deformation in the specimens. Pulse shaping (Chen et al., 2002, Song et al., 2007) was employed to increase rise time and achieve stress uniformity. Pulse shapers investigated included Styrofoam (2.45 mm thick, durometer 67 OOO), neoprene (2 mm thick, durometer 80 OOO) and silicone (~13 mm thick, durometer 26 OOO) foams, as well as several other foams and rubber materials not discussed herein. A photograph showing the setup of the specimen with the laser, photodetector, and quartz gauges can be seen in Figure 4.

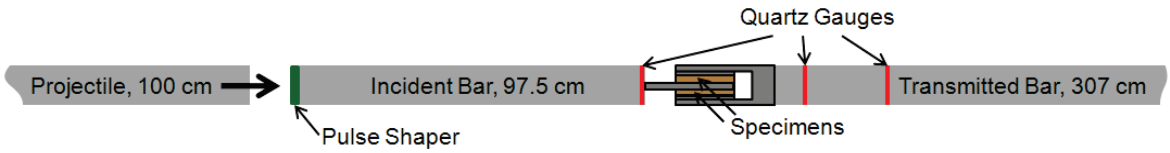

Figure 1: Schematic of the modified SHPB setup showing the aluminum bars, specimen, pulse shaper and instrumentation.

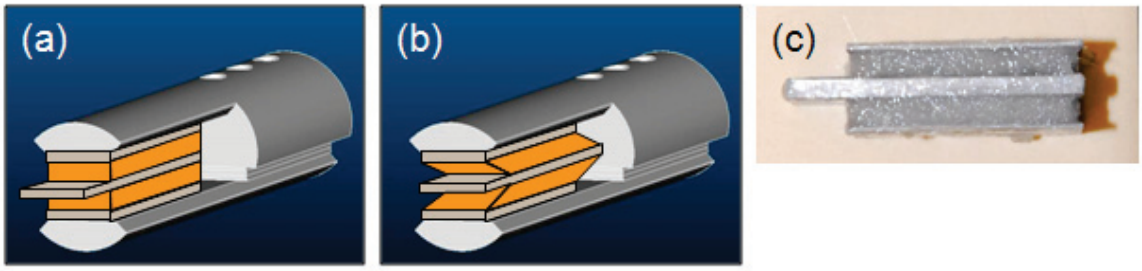

Figure 2: Schematic of specimens in double lab shear configuration (a) unstressed and (b) during impact. Fixtures were adjustable for individual specimen height. (c) A photograph of a silicone gel biological simulant in double lap shear configuration.

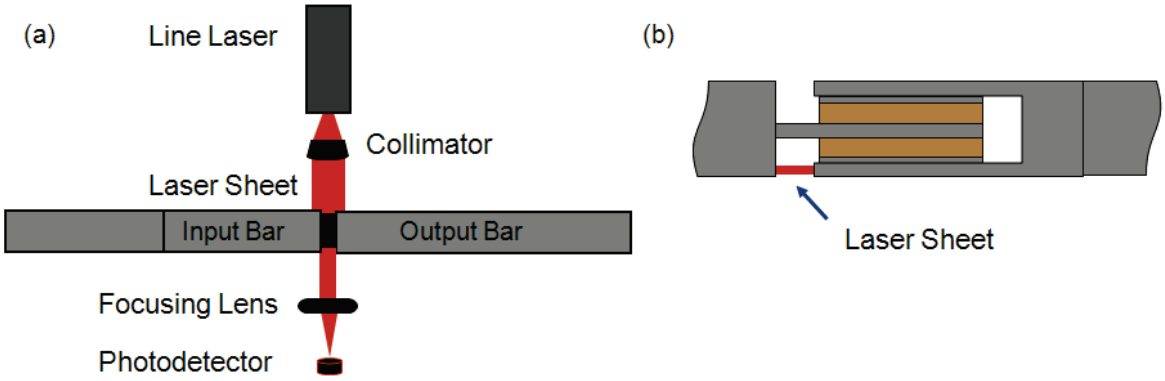

Figure 3: (a) Top view and (b) side view schematics of the collimated line laser and photodetector setup.

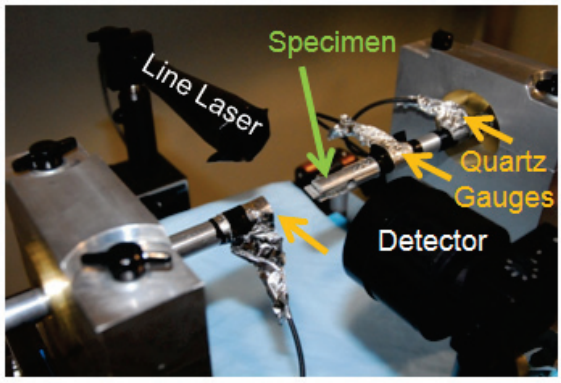

Figure 4: Photograph of the experimental setup showing a double lap shear specimen, placement of one input and two output quartz force gauges, line laser and photodetector.

Shear force and displacement are used to calculate shear stress and shear strain according to equations (1) and (2), respectively.

$$\tau(t) = \frac{F(t)}{2A} \quad (1)$$

$$\gamma(t) = tan^{-1}\left(\frac{\delta(t)}{h}\right) \quad (2)$$

where τ(t) is shear stress, F is force, A is area of specimen in contact with top or bottom aluminum plate, γ(t) is shear strain, δ is axial displacement, and h is sample thickness (Saraf et al., 2007, Saraf et al., 2007). Using (2), shear strain rate is calculated as

$$\frac{d\gamma}{dt} = \dot{\gamma} = \frac{1}{\left[1+\left(\frac{\delta}{h}\right)^2\right]}\frac{1}{h}\frac{d\delta}{dt}. \tag{3}$$

Shear stress as a function of shear strain data are plotted and shear modulus is then calculated by fitting a line to the data. For nonlinear data, a tangent modulus is calculated and reported for strain ranges.

Biological Simulants

Aluminum top/bottom end plates and center plate were machined and treated with silicone primer and adhesive. To make a single double lap shear specimen package, an aluminum end plate was placed into a custom mold, then silicone biosimulant was poured into the mold (enough to make a ~1 mm thick specimen), followed by an aluminum center plate, more biosimulant and then the second aluminum shim. Specimens were cured and then removed from the mold. A photograph of a typical biosimulant gel specimen package was shown in Figure 2(c).

Porcine Tissue

Porcine brain tissues were procured from a local slaughterhouse, prepared and tested fresh (within 24 hrs of slaughter) and frozen. The tissue specimen preparation process is shown in Figure 5. Samples for SHPB shear testing were prepared by embedding the tissue in agarose and then slicing thin brain sections, ranging from 1-2 mm thick, using a thin-kerf rotating blade. A custom punch was used to create 9 mm x 18 mm rectangular specimens from the thin sections. The specimens were subsequently bonded to aluminum shims in a double lap shear configuration using a cyanoacrylate tissue adhesive. Prior to testing, tissue samples remained hydrated by wrapping them in gauze and immersing the tissue in Hank's Buffer Solution. These experiments considered brain as a composite of white and gray matter, and orientation differences were not controlled.

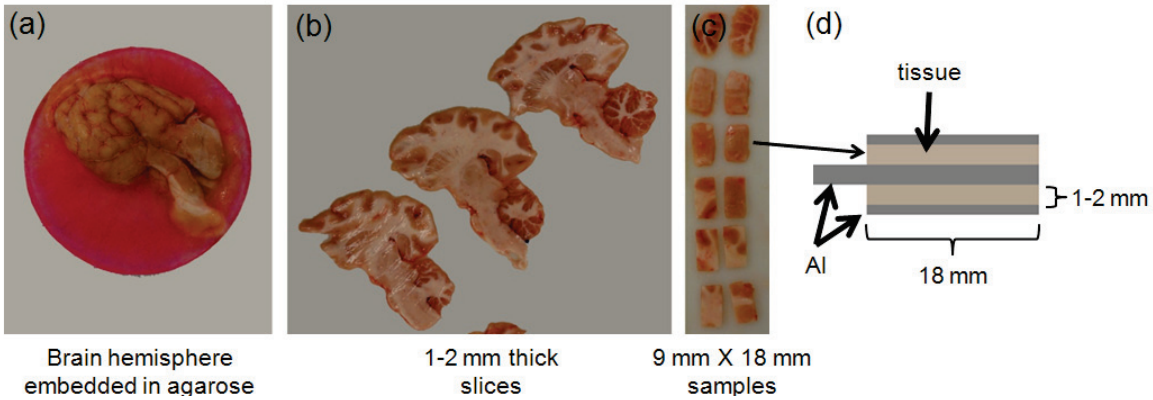

Figure 5: Photographs of the porcine brain tissue preparation process. (a) Brain was embedded in agarose and then (b) sliced into 1-2 mm thick sections using a thin-kerf rotating blade. (c) A punch was used to create 9 mm x 18 mm rectangular specimens, which were subsequently (d) bonded with cyanoacrylate to Al shims in the double lap shear configuration.

RESULTS

SHPB impact experiments were performed on silicone gel biological simulants and porcine brain tissue in the double lap shear configuration. A variety of pulse shaper materials were investigated and their effects on loading and stress uniformity in the specimens were analyzed. Figure 6 shows still images taken from a video captured using a Phantom V10 (Vision Research) high-speed digital video camera during a typical shear test on silicone biosimulant. Images were captured at a rate of 6400 frames per second. These still images show the specimen just prior to initial axial displacement of the center plate (t=0), during axial displacement (t=2.8 ms), and near the extent of complete axial displacement (t=4.4 ms). Axial displacement of the center plate is apparent as the input bar travels from left to right. This axial displacement creates shear deformation in the silicone gel specimens since the specimen surfaces adhered to the center plate are translating whereas the other surfaces are stationary. It can be seen in the image captured 4.4 ms after impact that the center plate ultimately impacts the back of the double lap shear fixture after it has translated approximately 4-5 mm. The experiment is effectively complete once this occurs and only data prior to this event is used for analysis.

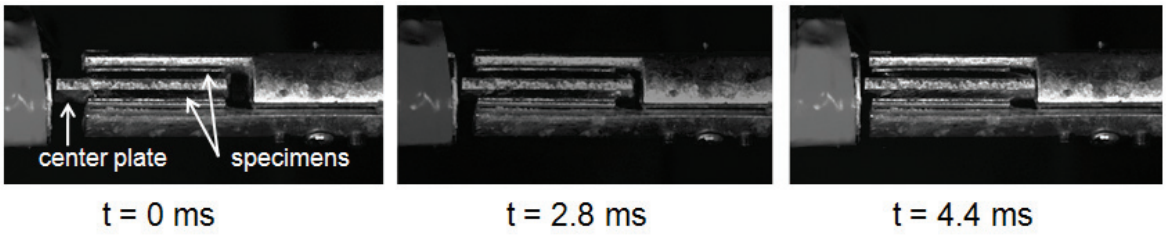

Figure 6: High-speed photographs captured during a shear SHPB experiment on silicone gel biosimulant. As time progresses, the input bar causes the center plate to displace, which in turn shears the silicone gel specimens. The test is complete when the center plate impacts the back of the fixture at ~4.4 ms.

To investigate the influence of candidate pulse shaper materials, several experiments were performed on biosimulants using a constant projectile firing pressure (25 psi), but with pulse shapers of various materials. Characteristic plots of shear strain and shear strain rate as a function of time are shown in Figure 7(a) and (b), respectively. These data reveal the significant differences in loading rate and sample deformation that are possible simply by changing the pulse shaper material. Using the same shot pressure, strain rates ranging from 1400 s^{-1} with no pulse shaper to 100 s^{-1} with a soft and thick pulse shaper are possible. However, it is important to verify whether stress uniformity is achieved under these conditions, as described below.

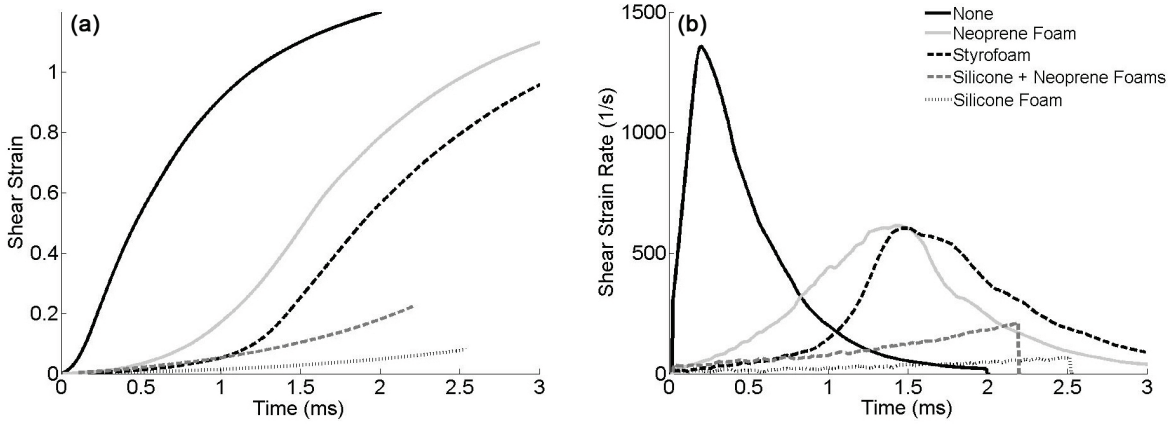

Figure 7: Comparison of (a) shear strain and (b) shear strain rate as a function of time for 1 mm thick silicone gel biosimulant tested with different pulse shapers. Data reveal that as pulse shaper stiffness increases, strain rate and strain in the sample increase for identical loading conditions.

Typical data collected from an impact without a pulse shaper are shown in Figure 8. Figure 8(a) shows the raw voltage data collected from the three quartz gauges as well as the line laser. It can be seen that although the two output quartz gauge signals agree well, the voltage measured by the input quartz gauge shows the spike feature caused by inertial effects and lack of stress uniformity, as described previously. This effect was expected given the soft specimens and the rapid loading produced without use of pulse shaping. Figure 8(b) shows the data after conversion to shear stress and shear strain. The same strain measurement was used for each curve, but stress from each quartz gauge was calculated separately. Again, it can be seen that the stress-strain curves resulting from the two output gauges agree well, but the stress data from the input gauge is significantly greater (~5 times). If stress uniformity had been achieved in this experiment, it would be expected that these three curves would be identical.

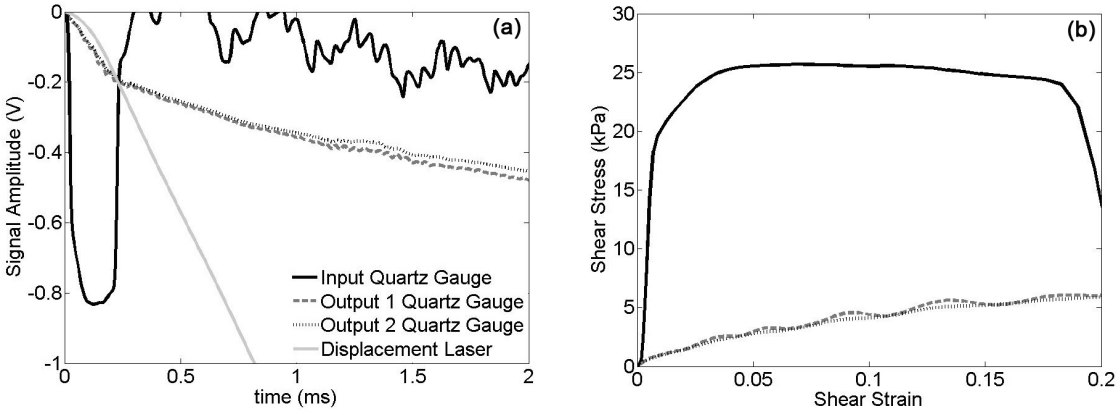

Figure 8: (a) Raw data collected from the input and two output quartz gauges as well as from the laser, and (b) stress-strain plots for shear loading of 1 mm thick silicone gel biosimulant samples without pulse shaping. Data show a large spike in input signal due to specimen inertia. This results in a large discrepancy between input and output stress measurements, indicative of non uniform stress distribution within the sample.

In a subsequent experiment on biosimulants, a Styrofoam pulse shaper was employed with the same loading conditions. Results from this experiment are shown in Figure 9(a) and (b). It can be seen in these figures that although the spike is still present in the input gauge data and stress uniformity has not been achieved, its magnitude is significantly less (~50%) than that in the case without pulse shaping. This indicates that the addition of the pulse shaper has decreased the inertial effects and the specimen is closer to a uniform stress state than was seen without pulse shaping. It appears that stress uniformity was maintained in the specimens for the first ~0.7 ms of the experiment, as indicated by agreement of all three gauge signals. At later times, however, it appears that the loading rate increases to the point that the stress state in the specimen becomes non-uniform. One possible explanation for this behavior is that near 0.7 ms the pores in the Stryofoam have completely closed, resulting in a rapidly increasing stiffness and, therefore, and rapid increase in the applied loading rate. The non-linearity of the response shown in Figure **9**(b) during input and output gage agreement (i.e. data prior 0.7 ms) may also be indicative of the rapidly increasing loading rate.

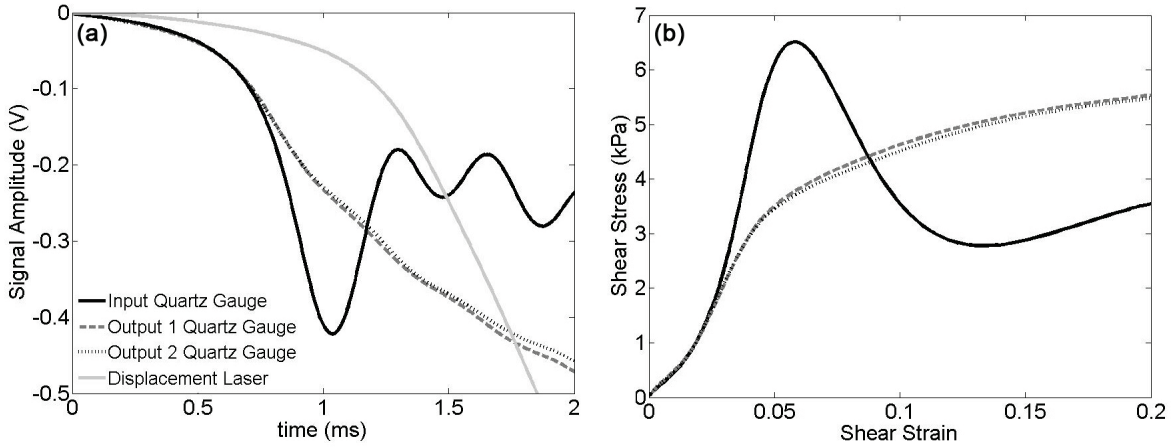

Figure 9: (a) Raw data collected from the input and two output quartz gauges as well as from the laser, and (b) stress-strain plots for shear loading of 1 mm thick silicone gel biosimulant samples with a Styrofoam pulse shaper. In this case the input signal initially matches the output signals, indicating a uniform state of stress for ~0.7 ms. Beyond that time, the input signal deviates from the output signal, indicating that the stress distribution becomes non-uniform.

Because specimen stress uniformity was not achieved when using the Styrofoam pulse shaper, a softer pulse shaper was needed. A silicone foam pulse shaper was utilized. This material was less stiff and also thicker when compared to the Styrofoam, so the loading rate was lower (as shown in Figure 7(a)), allowing for more time for stress to equilibrate. Results from impact of silicone biosimulant at 25 psi with a silicone foam pulse shaper are shown in Figure 10. It can be seen that all three quartz gauges record the same voltages during the experiments, which results in the same shear stresses regardless of which gauge is used for the calculations, as can be seen in Figure 10(b). The absence of the inertial spike and the agreement between all three gauges are indications that stress uniformity has been achieved in the specimens. This is a requirement for SHPB testing and is difficult to achieve in soft materials, but with proper pulse shaping techniques and thin specimens, high quality data is possible and can be used to determine the dynamic shear modulus of soft biological simulants and tissue.

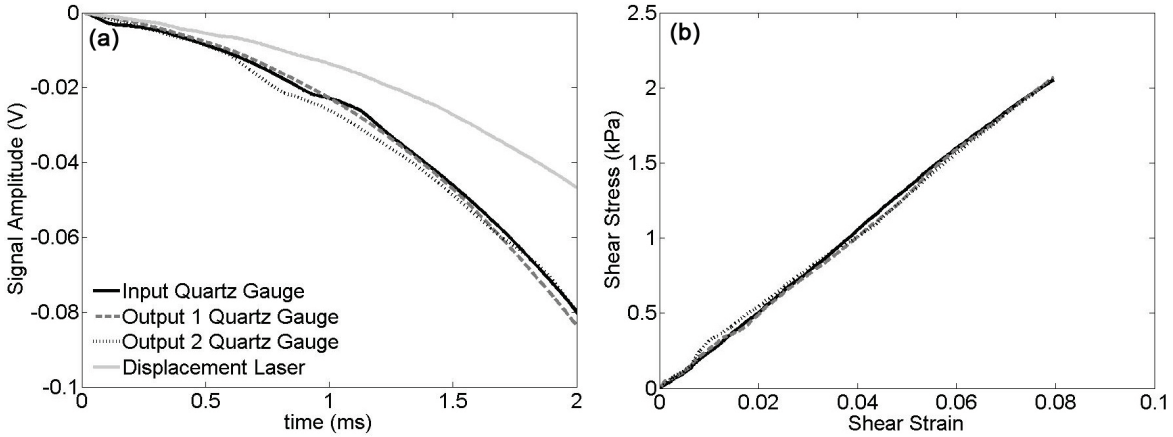

Figure 10: (a) Raw data collected from the input and two output quartz gauges as well as from the laser, and (b) stress-strain plots for shear loading of 1 mm thick silicone gel biosimulant samples with a silicone foam pulse shaper.

Figure 11 shows data for a similar pulse shaping exercise performed on porcine brain tissue specimens. It can be seen that the same conditions apply and a soft pulse shaper is required to achieve stress uniformity in the tissue. Figure 11(b) shows a case extremely close to stress uniformity in which the input stress values oscillate around those from the output gauges; the spike is still present, however. The silicone

foam pulse shaper was found to be adequate for achieving stress uniformity in the porcine brain tissue, as evidenced by the agreement of all three quartz gauges.

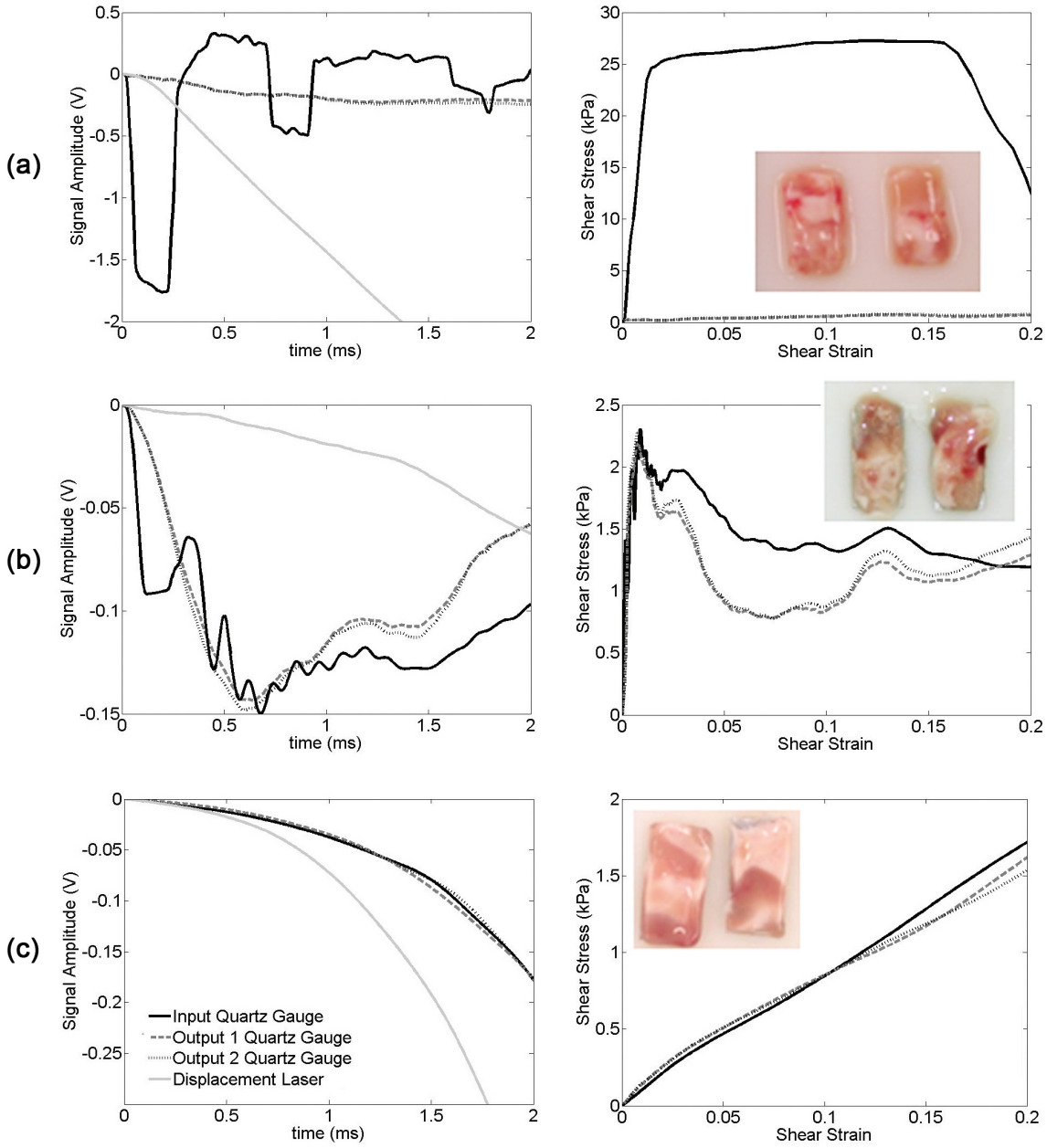

Figure 11: Raw data and shear stress-shear strain data for frozen porcine brain samples comparing pulse shapers. (a) Data collected without use of pulse shaper (shot pressure 30 psi) – stress uniformity was not achieved, (b) data collected with pulse shaping (shot pressure 24 psi), but still not complete stress uniformity, and (c) data collected with acceptable pulse shaping (shot pressure 32 psi), determined by agreement between the input and both output quartz gauges. Inset photographs show porcine brain samples.

Figure 12 illustrates the importance of verifying the stress uniformity in each experiment. When stress uniformity is not verified and data from all experiments is considered "good", the shear stress-shear strain responses can vary significantly, as can be seen in Figure 12. Shear moduli shown here range from 3.8

to 89.5 kPa, which generates a tremendous amount of scatter. When only the data from experiments with verified stress uniformity are considered, data are more consistent, as can be seen in Figure 13(a). Some amount of scatter is expected due to the inherent variance of biological tissue mechanical properties. However, comparison of Figure 12 with Figure 13 is meant to illustrate that artificial scatter can be introduced by bad quality data arising from neglecting to verify stress uniformity in the specimens. Data are only preliminary and are used simply to illustrate the importance of pulse shaping and verifying stress uniformity.

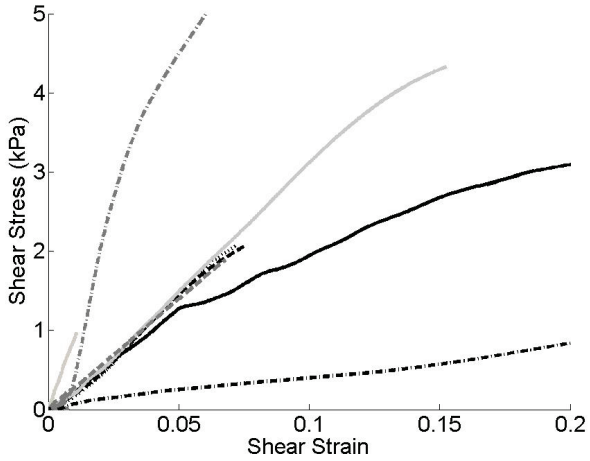

Figure 12: Shear stress-shear strain data for porcine brain. Data include tests both with and without stress uniformity in the specimens. Large scatter is apparent in data; shear moduli range from 3.8 to 89.5 kPa. Apparent scatter is due to tests lacking stress uniformity.

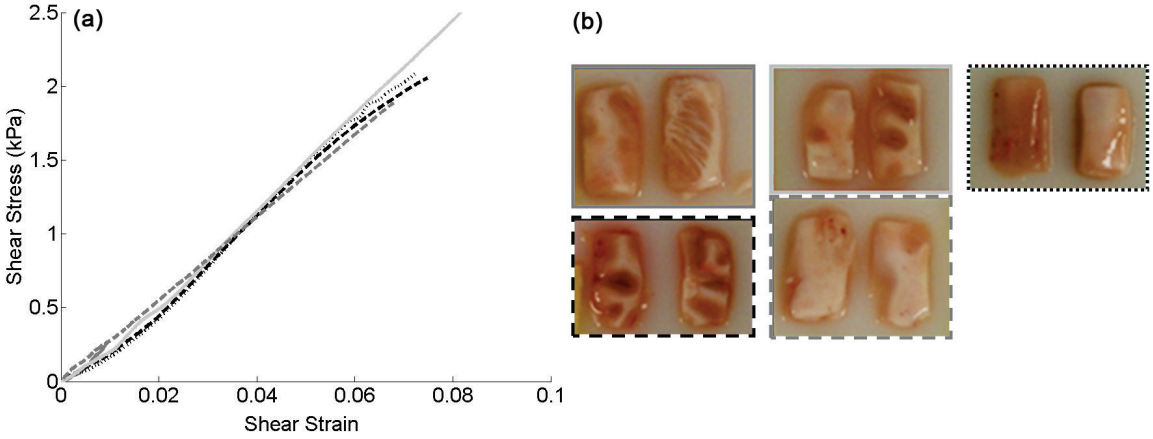

Figure 13: (a) Preliminary data from experiments on porcine brain in which uniform stress distribution was achieved and (b) corresponding photographs of specimens. Shot pressures ranged from 34-40 psi.

CONCLUSIONS

A modified SHPB technique has been demonstrated for testing soft biological simulants and tissue under dynamic shear loading. Stress uniformity criteria have been established and agreement between input and output quartz gauges is used to verify that the criteria are met. Suitable pulse shapers have been identified for achieving acceptable uniformity within the specimens.

In the future, this technique will be employed to characterize the dynamic shear response of human biological tissue and candidate biosimulant materials. This technique will also be extended to hydrostatic

pressure loading conditions for determination of bulk modulus. Upon full characterization of the selected organ tissues and biological simulant candidates, their dynamic mechanical properties will be compared and representative biosimulants will be selected for each organ for use in surrogates. The tissue properties will also be used in computational models of the human body.

ACKNOWLEDGEMENTS

The authors would like to acknowledge the funding support of The Office of Naval Research (Contract W911QY-09-C-0058) and The National Highway Traffic Safety Administration (Contract DTNH22-05-H-01021-03). They would also like to thank Jeff Paulson and Craig Leese for their expertise and contributions in the development of the biosimulant materials, and Jenna Graham for help with development of the Matlab code for data analysis.

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