2

INJURY BIOMECHANICS RESEARCH
Proceedings of the Thirty-second International Workshop

An Intravascular Pressure Measurement Technique Applied to Impact Testing of Unembalmed Human Liver Specimens

J. L. Sparks, J. Stammen, R. Herriott, and K. H. Jones

ABSTRACT

The anatomic complexity and heterogeneity of the human abdomen make it difficult to develop a biofidelic abdominal component in an anthropomorphic test device that can predict abdominal injuries in crash testing. Several previous studies have attempted to relate external physical parameters such as energy, external pressure, impact force, velocity, and compression to abdominal injury. While many of these approaches have been useful in developing response corridors for the abdomen as a whole, the interaction between different abdominal organs and the responses of individual organs are not well understood. This paper presents a method for measuring intravascular pressure gradients within isolated, physiologically pressurized human livers in response to blunt impact. Intravascular pressure data is presented from testing of a rigidly constrained, unembalmed liver specimen. Future work will include analysis of intra-liver pressure changes and injury outcomes in impact tests of intact post-mortem human subjects (PMHS). By matching the pressure gradients in the isolated liver testing to those observed in the intact PMHS tests, the boundary conditions describing the liver's environment within the abdominal cavity could potentially be modeled.

INTRODUCTION

Studies of field accident data have found that blunt trauma to the abdomen is a common result of motor vehicle accidents. An analysis of the National Crash Severity Study (NCSS) found that injury to the abdomen accounted for 29.9% of all severe (AIS ≥ 4) and 30.7% of all critical (AIS ≥ 5) injuries (Ricci, 1980). A more recent study of the National Automotive Sampling System (NASS) database found that injury to the abdomen constituted 16.5% of all AIS ≥ 4 injuries and 20.5% of all AIS ≥ 5 injuries (Elhagediab and Rouhana, 1998).

Field accident data also indicates that the liver is among the most frequently injured of all the abdominal organs, in both frontal and side impact collisions. Elhagediab and Rouhana analyzed the NASS database for abdominal injuries in non-rollover frontal impacts to non-ejected drivers and right front passengers. They found that the liver was the most frequently injured abdominal organ (38%), followed by the spleen (23%) and the digestive system (17%). In a study of the NCSS database for side impact collisions, Rouhana and Foster (1985) found that for passengers in right-side impacts, injury to the liver was the most common serious abdominal injury. For drivers in left-side impacts, liver injury was the second most common serious abdominal injury, second only to kidney injury. Overall, these studies are consistent with the general finding that solid abdominal organs, such as the liver and spleen, tend to be injured more frequently than hollow organs such as the stomach and intestines (Rouhana, 2001).

Previous studies have attempted to relate several external physical parameters to abdominal injury. Early research by Mays (1966) investigated the amount of energy input needed to reproduce clinically relevant injuries in isolated cadaver livers. His methodology involved dropping liver specimens onto a concrete surface from varying heights. He found that superficial lacerations of the liver capsule corresponded to an energy input range of 36 to 46 J, deep lacerations without vascular involvement corresponded to a range of 144 to 182 J, and extensive maceration of the parenchyma with severe disruption of the vasculature and bile ducts corresponded to a range of 386 to 488 J. He also observed that in order to replicate clinically relevant injury patterns it was necessary to re-pressurize the liver specimens by injecting them with saline and barium prior to testing.

A study by Walfisch et al. (1980) investigated external pressure (pressure applied to the organ surface or abdomen surface) as a predictor of injury severity. In this study, cadaver subjects were dropped on their right sides onto simulated armrests from varying heights. The study found that an applied pressure of 260 kPa was associated with AIS $\geq$ 3 liver injury. The investigators also concluded that the average pressure on the armrest was a good predictor of injury severity, with $r = 0.93$.

Viano et al. (1989) investigated the correlation between impact force and abdominal injury by conducting a series of lateral impact tests of unembalmed cadaver subjects using a rigid pendulum impactor. Of all the biomechanical parameters compared in this study, including viscous response, maximum compression, and acceleration, the investigators found that peak force was the best correlated parameter with AIS $\geq$ 4 abdominal injury. A study by Cavanaugh et al. (1993) also analyzed lateral impact tests of human cadaver subjects. These tests utilized a side-impact sled design. The authors reported that the peak impact force was not a reliable indicator of abdominal injury severity.

In a study of 117 abdominal impacts to anesthetized rabbits, Rouhana et al. (1985) found that the product of maximum pre-impact velocity (V) and maximum abdominal compression (C) was well correlated with the severity of abdominal injury. A study by Stalnaker and Ulman (1985) reanalyzed data from previous abdominal impact research to investigate the relationship between V*C and abdominal injury severity in both primates and human cadaver subjects. They found that V*C was very well correlated with AIS $\geq$ 3 abdominal injury ($r = .92$ to $.99$), if each abdominal region was considered separately and if side impacts were considered separately.

In much of the previous work on blunt abdominal injury, the focus of attention has been on physical parameters that are measured externally, either outside the abdominal wall or at the surface of the abdominal organ of interest. While many of these approaches have been useful in developing response corridors for the abdomen as a whole, the interaction between different abdominal organs and the responses of individual organs are not well understood. The goal of this study is to develop a new approach for analyzing the impact response of the liver by placing instrumentation inside the liver itself and measuring pressure changes inside the blood vessels of the liver during impact. This paper presents a methodology for measuring intravascular pressure gradients within isolated, physiologically pressurized human livers in response to blunt impact.

METHODS

Anatomy of the Liver

The liver, the largest gland in the body, is situated predominantly in the upper right quadrant of the abdomen. It is held in position in the body through two means: (1) it is supported from below by the presence of several other abdominal organs, such as the stomach and large intestine, and (2) it is attached to the diaphragm by five ligaments, including the falciform, coronary, round, and left and right triangular ligaments. The three major components of the liver vascular system are the hepatic artery, the inferior vena cava, and the hepatic portal vein. The hepatic artery carries oxygenated blood into the liver at a rate of about 400 mL/minute (Guyton, 1976). The pressures in the hepatic artery range in a pulsatile manner from roughly 80-120 mmHg. The inferior vena cava (IVC), which carries venous blood from the liver and from the rest of the lower part of the body back toward the heart, generally exhibits low physiologic pressures in the range of 0-1 mmHg. The hepatic portal vein carries 1000 mL/minute of venous blood into the liver, and the pressure in the portal vein ranges from 7-10 mmHg (Guyton, 1976). The physiologic conditions in each part of the liver vascular system provided the basis for much of the experimental apparatus that was developed in this study for testing isolated liver specimens.

Selection of Instrumentation

Several criteria were used to guide the selection of appropriate instrumentation for this study. The device had to be small enough to be placed within the hepatic veins through the opening of the inferior vena cava on the superior surface of the isolated liver specimens. The size of the hepatic veins normally ranges from approximately 2-6 mm in diameter at the largest point, gradually decreasing as they approach the level of the hepatic sinusoids. The device also needed to function under specific environmental conditions, including submersion in saline solution at room temperature. Finally, the device had to be capable of measuring pressures in the range of 300-400 kPa (44-58 psi), with a response time sufficient for high velocity impacts.

Based on these considerations, fluid-filled angiographic catheters were selected to measure pressure gradients inside the blood vessels of the liver. The catheters have an outer diameter of 1.7 mm (size 5 French) and are designed for use in the human vascular system (Cordis Corp., model MPA2). Angiographic catheters are used commonly in the clinical setting for a variety of purposes such as injecting dye into the coronary arteries or measuring blood pressures inside the chambers of the heart. In order to use the catheters to measure intravascular pressure gradients inside the liver, the open tip or internal end of the catheter was positioned inside the lumen of a hepatic vein, and the remote (external) end of the catheter was connected to a pressure transducer (PSI-Tronix). The pressure transducers have a range of 0-689 kPa (0-100 psi). In this instrumentation technique, a continuous column of fluid exists from the vein through the catheter to the sensing element of the pressure transducer, so that pressure changes recorded by the transducer are related to pressure changes occurring locally at the open tip of the catheter inside the blood vessel.

Mathematical Correction of Catheter-based Pressure Measurements

In order to determine the actual local pressure changes occurring inside the hepatic veins in response to impact, it was necessary to develop a mathematical relationship between the pressure at the internal end of the catheter (P_{int}) and the pressure at the remote end of the catheter (P_{rem}). Previous studies have found that a fluid-filled catheter can be described as an underdamped second order linear system (Glantz and Tyberg, 1979; Lambermont et al., 1998):

$$\text{(1)} \qquad \ddot{P}_{\text{rem}} + 2\omega_{\text{n}}\zeta\dot{P}_{\text{rem}} + \omega_{\text{n}}{}^{2}P_{\text{rem}} = CP_{\text{int}}$$

In this equation, ω_n is the undamped natural frequency, ζ is the damping ratio, and C is a proportionality constant related to the physical properties of the pressure transducer. The dynamic response of the system can be described by specifying the two parameters ω_n and ζ.

A "fast-flush" transient step-response test was used to determine the undamped natural frequency and damping ratio of the catheter-transducer system. This technique has been described and tested in several previous studies (Glantz and Tyberg, 1979; Sheahan et al., 1991; Kleinman et al., 1992; Brennan and O'Hare, 1998). The experimental set-up for the fast-flush test is depicted in Figure 1. The catheter-transducer system was subjected to a step increase in pressure by opening and quickly closing the flush valve, initiating a step-like pulse input to the system. Pressure transducers were positioned at both ends of the catheter to record pressure changes corresponding to P_{int} and P_{rem}. The pressure waveform from each transducer showed an initial spike followed by transient oscillations, as illustrated in Figure 2.

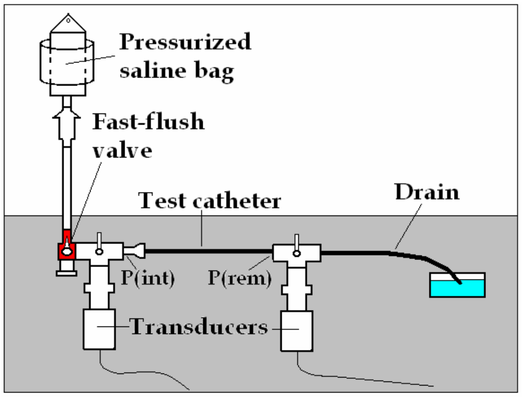

Figure 1: Experimental set-up for fast-flush test to determine catheter frequency response characteristics.

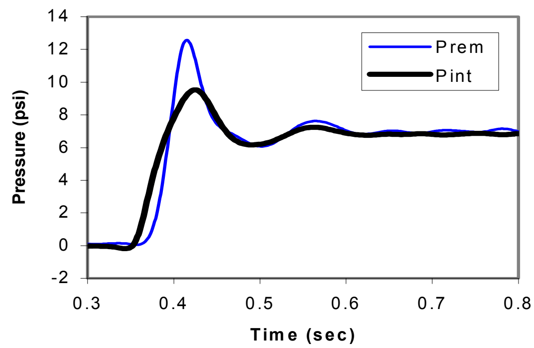

Figure 2: Results of fast-flush test. Readings from both transducers show an initial spike followed by transient oscillations.

According to the methodology published by Glantz and Tyberg (1979), the undamped natural frequency and damping ratio of the system can be calculated based on the transient oscillations recorded by the downstream transducer (P_{rem}) using a logarithmic decrement approach. The amplitudes of four consecutive cycles were recorded, the natural logarithms of these values were plotted versus time, and the slope (α) of the resulting best-fit line was calculated. The average period (T) also was calculated for the four consecutive peaks. Once T and α were known, these values were used in the following equations to calculate the undamped natural frequency (ω_n) and damping ratio (ζ) of the system:

(2) $$\omega_n = \left[(1/T^2) + (\alpha^2/4\pi^2)\right]^{1/2}$$, with T in sec, α in sec^{-1}, and ω_n in Hz

(3) $$\zeta = \alpha/(2\pi\omega_n)$$

The calculated value of ω_n was 19.3 Hz (121.27 rad/sec) and the calculated value of ζ was 0.105. These results were then substituted into Equation (1) to provide a mathematical approach for determining P_{int} when P_{rem} is known:

(4) $$\ddot{P}_{\text{rem}} + (25.5)\dot{P}_{\text{rem}} + (14706)P_{\text{rem}} = CP_{\text{int}}$$

For the catheter-transducer system in this study, Equation (4) was proposed as a means for estimating the local intravascular pressure changes based on the pressure changes recorded by the remote pressure transducer at the external end of the catheter.

Verification of Local Pressures at Internal Catheter Tip

A verification experiment was conducted in order to evaluate how well the mathematically corrected pressure changes ($P_{int(pred)}$) corresponded to actual pressure changes occurring at the internal end of the catheter ($P_{int(meas)}$). The experimental set-up for this series of tests is shown in Figure 3. In addition to comparing measured and predicted internal pressure measurements, this set-up also was used to examine the effects of catheter orientation, catheter diameter, distance from impact site, and impact force on the resulting pressure readings. The test chamber was designed and built to allow controlled volume changes due to blunt impact. The chamber is a six inch diameter cylinder made of transparent cast acrylic. High-strength ¼ inch thick latex diaphragms were fit securely to the ends of the chamber in line with the impact direction. A removable acrylic platform with a grid arrangement of numerous small holes was constructed to facilitate attachment of transducers and catheters at several locations within the chamber. The chamber was attached to an aluminum bracket fixture that could be secured temporarily to a solid immovable base using clamps.

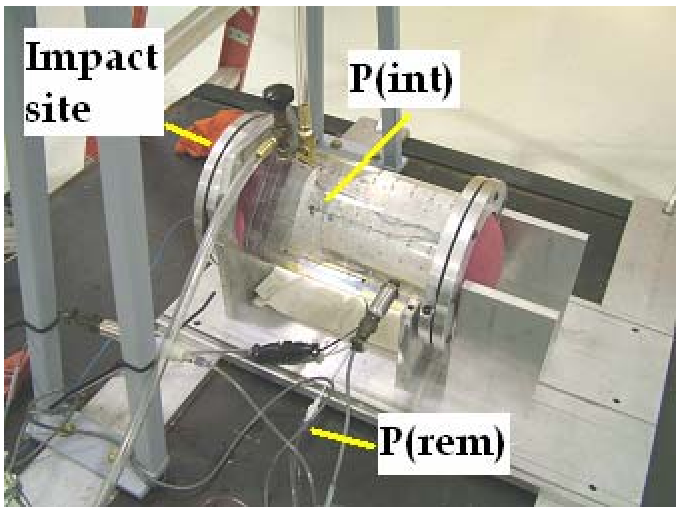

Figure 3: Test chamber used for experiments to verify the local pressure at the internal catheter tip for four impact velocities.

The chamber design included four 3/8 inch diameter holes or ports. The port on the lower surface of the chamber allowed the insertion of catheters and transducers inside the chamber. This port was sealed with waterproof RTV prior to filling the chamber with fluid. The two threaded ports on the top surface permitted the connection of a filling tube and a bleed-off tube. The threaded port on the side of the chamber allowed the direct connection of an additional pressure transducer to record static and dynamic pressure changes near the chamber wall. In addition to the four 3/8 inch ports, four 1/8 inch threaded holes were included in the chamber design to accommodate setscrews used to hold the acrylic platform in position.

Water was selected as the fluid medium to use in these tests due to its similarity in viscosity to physiologic saline. After the chamber was filled with water, a pre-test static pressure of 100 mmHg was established inside the chamber by using the fill tube as a manometer with a fluid column 1.36 meters in height, according to the standard hydrostatic formula derived from Newton's second law:

(5) $$H = (P_{chamber} - P_{atm}) / \rho_{water}\, g$$, where H = fluid height, ρ = density of water, g = acceleration due to gravity

The pre-test pressure of 100 mmHg was chosen as an arithmetic average of typical systolic and diastolic pressures of 120 mmHg and 80 mmHg, respectively.

The instrumentation used in these tests included angiographic catheters, miniature pressure transducers, and standard pressure transducers. The catheters were size 5 French, with an outer diameter of 1.4 mm and an inner diameter of 1.3 mm, and were 65 cm in length (Cordis Corp.). The miniature pressure transducers were 1.3 mm in diameter and 13 mm in length, and had a measurement range of 0-689 kPa (0-100 psi, Entran EPS). The standard pressure transducers were ¾ inch in diameter, 2 inches in length, and had a measurement range of 0-689 kPa (0-100 psi, PSI-Tronix).

The test plan for this series of experiments involved measuring pressure changes at two locations, P_{int} and P_{rem}, for four different impact speeds ranging from 1.5 to 3.0 m/s. P_{int} was measured using a miniature pressure transducer positioned inside the chamber immediately adjacent to the open end of the fluid-filled catheter. P_{rem} was measured using a standard pressure transducer attached to the remote end of the fluid-filled catheter, positioned outside of the chamber. The objective of the tests was to use the measured values of P_{rem} to calculate a predicted internal pressure curve, $P_{int(pred)}$, according to the mathematical relationship described in Equation 4. This predicted internal pressure curve could then be compared to the measured internal pressure at the catheter tip inside the chamber, $P_{int(meas)}$, to evaluate the validity of the correction equation across a range of impact speeds.

In order to apply Equation (4) it was necessary to calculate the value of the proportionality constant *C*. The internal pressure curves, $P_{int(pred)}$ and $P_{int(meas)}$, can be considered as two time series that are linearly related. The cumulative variance of the two time series was calculated for each of the four impact velocities. Then the combined cumulative variance was calculated for the whole set of time series, over all four impact velocities. This variance was minimized on an amplification factor, *C*, which was found to be 16,670 sec^{-2}. This constant is believed to be a function of both the electronic sensitivity and diaphragm geometry of the pressure transducer. Prior to applying the transfer function, the remote pressure data was filtered using a Butterworth low-pass filter at 22 Hz and sub-sampled from 20,000 to 500 samples per second to minimize differentiation errors. A fast Fourier transform (FFT) showed that these filter/sampling frequencies did not interfere with the useful portion of data obtained in the test.

Impact Test of Isolated Liver Specimen

A preliminary impact test of an isolated liver specimen was conducted using the experimental technique described below. The major components of the experimental methodology include isolation and pressurization of the liver specimen, instrumentation, and impact testing. The goal of the preliminary test was to use this methodology to record pressure changes inside the blood vessels of an isolated, physiologically pressurized human liver in response to a low-energy blunt impact. The correction equation developed in the previous sections was then applied to the results to convert the remote, catheter-based pressure measurements to local intravascular pressure changes.

The liver specimen used in this study was obtained from the autopsy service at the Ohio State University in accordance with protocol approved by the Institutional Review Board. The specimen was removed by cutting the diaphragm in order to preserve the original ligamentous attachments between the liver and the diaphragm. The liver specimen was separated from its vascular attachments at two points: the porta hepatis and the inferior vena cava (IVC). The porta hepatis includes the hepatic portal vein, the hepatic artery proper, and the common bile duct. In order to allow sufficient working distance for the placement of instrumentation, these attachments were severed a small distance (4-5 cm) from their point of entry into the liver. The IVC was cut approximately 2 cm above and 4 cm below where it extends beyond the liver's superior and inferior surfaces. Prior to testing, the specimen was submerged and flushed with room temperature saline solution.

The experimental apparatus for testing isolated liver specimens was designed to replicate, to some degree, the physiologic pressures that are found in each of the three parts of the liver vascular system: the hepatic artery, the portal vein, and the inferior vena cava. The set-up is illustrated in both schematic and photographic form in Figure 4. The pressure sources for the hepatic artery and portal vein were designed to maintain a constant level of fluid at a predetermined height so that the appropriate constant pressure could be applied to each of these vessels. The hepatic artery pressure source was positioned 50 inches above a horizontal reference plane through the center of the liver test chamber. This height corresponds to a pressure of 93 mmHg in the hepatic artery, which represents an equal-area average of the physiologic range of 80 to 120 mmHg (diastolic to systolic) within the hepatic artery. The portal vein pressure source was positioned

4.5 inches above the liver reference plane, which corresponds to a pressure of 8.4 mmHg in the portal vein. This pressure is within the physiologic range of 7-10 mmHg inside the portal vein. A three-way T-valve was connected to the opening of the inferior vena cava at the superior surface of the liver. When this valve was in the open position, fluid could drain out of the liver through a length of tubing into a reservoir.

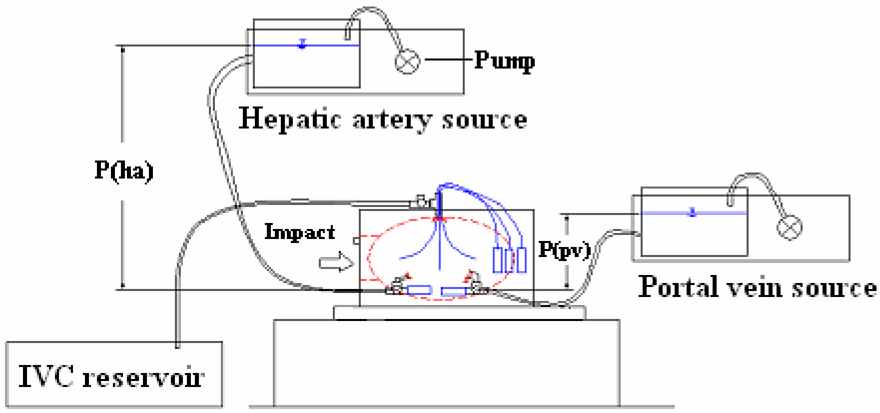

Figure 4A: Schematic of experimental apparatus for isolated liver impact tests. The red dotted line represents the liver inside the test chamber. The blue lines represent catheters positioned in three different hepatic veins inside the liver.

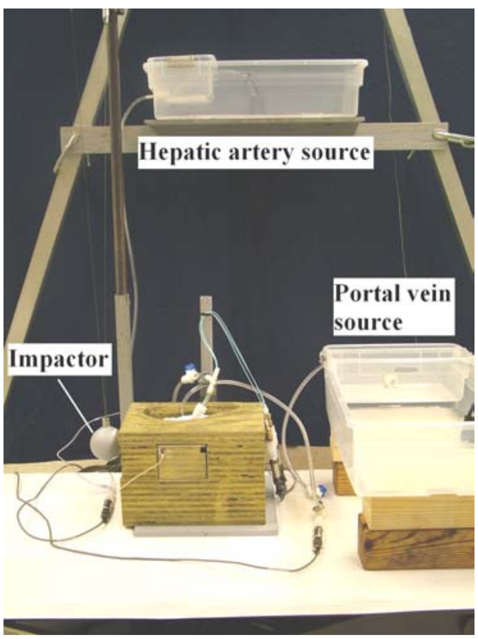

Figure 4C: Components of test chamber.

Figure 4B: Photo of experimental test set-up.

The instrumentation of the liver involved three catheters (Cordis Corp) and five pressure transducers (PSI-Tronix). The three catheters were routed through the T-valve and through the superior opening of the IVC into three different hepatic veins, prior to positioning the liver in the test chamber. The external end of each catheter was connected to a pressure transducer, and these transducers were positioned approximately level with the catheter tips within the test chamber to minimize hydrostatic error. Readings from these three transducers were recorded prior to impact and during the impact event. Additional pressure transducers were connected to the tubing attached to the hepatic artery and portal vein. These two transducers were used to monitor the applied static pressure in the hepatic artery and portal vein prior to impact and could also be used to measure pressures dynamically.

Impact energy was applied to the specimen using a 1.6 kg pendulum impactor. The design of the test chamber permits the impactor to strike a movable face at either a front or side location on the chamber (see Figure 4C). This movable face interacts with the liver inside the test chamber, and an accelerometer is mounted to the movable face. In the preliminary test, the impactor struck the movable face at the side position so that the right side of the liver was impacted. The impact velocity was 2.9 m/s.

RESULTS

Fast-Flush Test Results

The results of the fast-flush test are shown in Figure 5. P_{rem} indicates the pressure changes measured by the remote transducer, farther downstream from the pressurized saline bag (see Figure 1 for set-up). $P_{int(meas)}$ refers to the pressure changes recorded by the transducer closest to the pressurized saline bag. This transducer is analogous to the internal miniature pressure transducer in the chamber tests. $P_{int(pred)}$ indicates the pressure curve that was generated by applying the correction equation (Equation 4) and the *C* value of 16,670 sec^{-2} to the P_{rem} data points. A comparison of the maximum values of $P_{int(meas)}$ (9.54 psi) and $P_{int(pred)}$ (9.52 psi) indicated a 0.21% error.

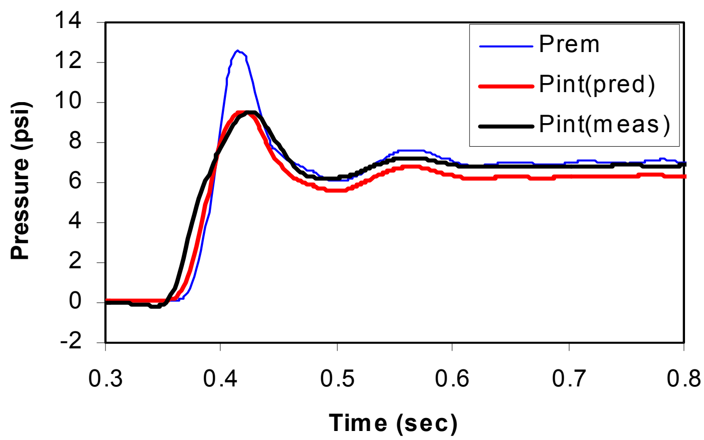

Figure 5: Results of fast-flush test to determine catheter frequency response characteristics. Predicted internal pressure curve also is shown.

Chamber Test Results

Figure 6 illustrates the results of the series of chamber tests for four impact velocities: 1.5, 2.0, 2.5, and 3.0 m/s. The pressure curves for each transducer at each impact velocity were averaged from two identical sets of tests. P_{rem} refers to the readings from the pressure transducer at the external end of the catheter, outside of the test chamber. $P_{int(meas)}$ represents the pressure changes recorded by the miniature pressure transducer adjacent to the open tip of the fluid-filled catheter inside the test chamber. $P_{int(pred)}$ indicates the curve that was generated by applying the correction equation to the output of P_{rem}, using C = 16,670 sec^{-2}. The maximum values for $P_{int(meas)}$ and $P_{int(pred)}$ are listed in Table 1 for each impact velocity.

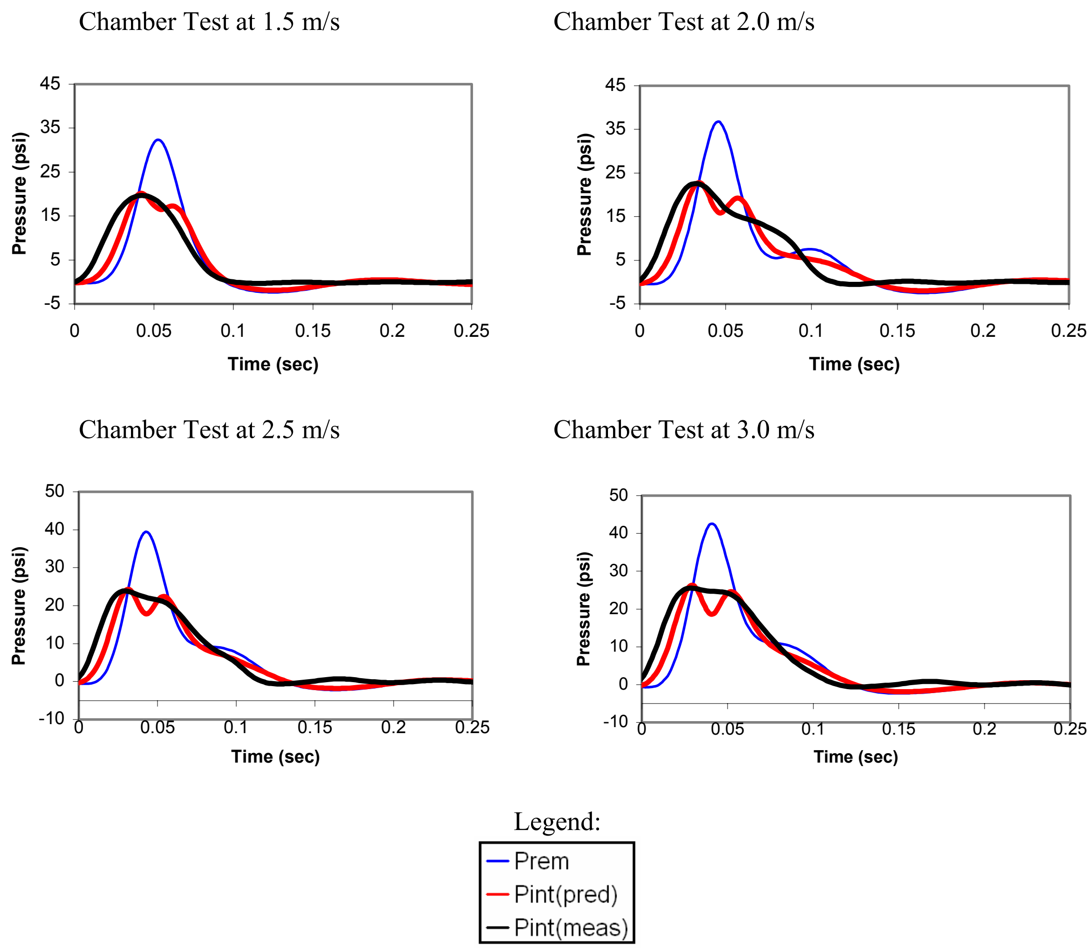

Figure 6: Results of chamber tests for four impact velocities.
Predicted internal pressure curves also are shown.

Table 1. Comparison of predicted and measured internal pressures at catheter tip for four impact velocities.

Impact Velocity (m/s)	**Maximum $P_{int(pred)}$** (psi)	**Maximum $P_{int(meas)}$** (psi)	**Percent Difference** (%)
1.5	20.2	19.7	2.5
2.0	22.8	22.6	0.9
2.5	24.3	23.9	1.7
3.0	26.3	25.5	3.1

Isolated Liver Impact Test Results

The results of the preliminary low-energy impact test of an isolated, physiologically pressurized human liver are shown in Figure 7. The intravascular pressure changes were recorded by pressure transducers connected to three catheters that were positioned in three different hepatic veins inside the liver. The correction equation was then applied to the data to generate the estimated changes in local pressure at each catheter tip inside the vascular system. Figure 8 illustrates the approximate orientation of the three catheters with respect to the impact site.

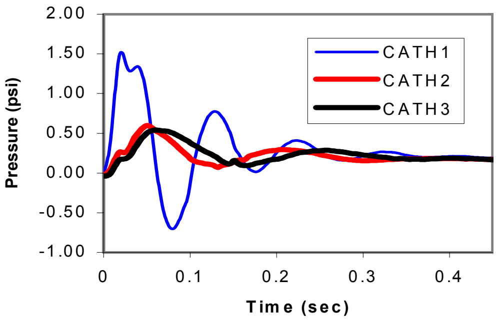

Figure 7: Local intravascular pressure gradients inside three hepatic veins, in response to low-energy blunt impact.

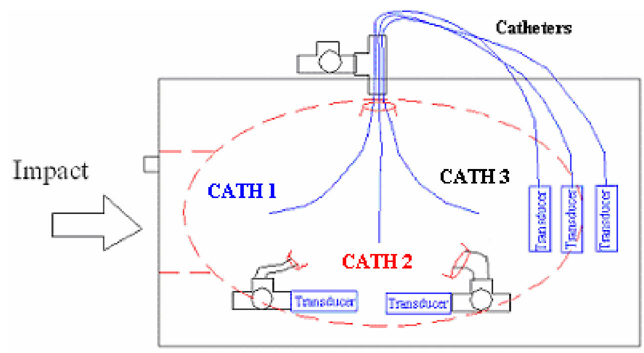

Figure 8: Approximate orientation of three catheters with respect to impact site.

CONCLUSIONS

The results of the fast-flush test indicated that this technique is a promising method for obtaining information about the frequency response characteristics of the fluid-filled catheter – transducer system used in this study. The results of the four chamber tests showed considerable similarity between the maximum predicted and measured internal pressures for each of the four impact velocities. This finding helps to support the robustness of the correction equation for predicting local pressures at the internal catheter tip under different experimental conditions. The most notable feature of the isolated liver impact test results is that the catheter positioned closest to the impact site read the highest pressures. Future work will include the use of radiologic imaging to determine the precise location of the catheters within a three-dimensional coordinate system. Also, post-test liver injuries will be assessed in higher energy impacts through gross inspection, microscopic analysis of tissue samples, and CT scanning of specimens. A long-term objective will be to investigate whether a relationship exists between measured intravascular pressure gradients and liver injury in post-mortem human subjects.

ACKNOWLEDGEMENTS

The authors would like to thank Dr. John Bolte of the Ohio State University Impact Biomechanics Research Laboratory, Dr. Bruce Donnelly of NHTSA Vehicle Research and Test Center, and Duane Hart of the Ohio State University Medical Center Clinical Engineering Department for their valuable contributions to this project.

REFERENCES

BRENNAN, E.G. and O'HARE, N.J. (1998). Calibration and assessment of a fluid-filled catheter—transducer system for the measurement of ventricular diastolic pressures. Physiol. Meas., 19, 405-412.

CAVANAUGH, J.M., HUANG, Y., ZHU, Y., and KING, A.I. (1993). Regional tolerance of the shoulder, thorax, abdomen, and pelvis to padding in side impact. SAE Technical Paper No. 930435.

ELHAGEDIAB, A.M. and ROUHANA, S.W. (1998). Patterns of abdominal injury in frontal automotive crashes. In: 16^{th} International ESV Conference Proceedings, 327-337.

GLANTZ, S.A. and TYBERG, J.V. (1979). Determination of frequency response from step response: application to fluid-filled catheters. Am. J. Physiol. 5, H376-H378.

GUYTON, A.G. (1976). *Textbook of medical physiology*. W.B. Saunders: Philadelphia.

KLEINMAN, B., POWELL, S., KUMAR, P., and GARDNER, R.M. (1992). The fast flush test measures the dynamic response of the entire blood pressure monitoring system. Anesthesiology, 77, 1215-1220.

LAMBERMONT, B., GERARD, P., DETRY, O., KOLH, P., POTTY, P., D'ORIO, V., and MARCELLE, R. (1998). Correction of pressure waveforms recorded by fluid-filled catheter recording systems: A new method using a transfer equation. Acta Anaesthesiol. Scand., 42, 717-720.

MAYS, E.T. (1966). Bursting injuries of the liver. Arch. Surg., 93, 92-106.

RICCI, L. (1980). NCSS statistics: passenger cars. UM-HSRI report 80-36.

ROUHANA, S.W. (2001). Biomechanics of abdominal trauma. In: Accidental injury: biomechanics and prevention, 2nd edition. A.M. Nahum and J.W. Melvin (eds.). Springer-Verlag: New York, pp. 405-453.

ROUHANA, S.W. and FOSTER, M.E. (1985). Lateral impact—an analysis of the statistics in the NCSS. Proc. 29th Stapp Car Crash Conference, SAE Technical Paper No. 851727, 79-98.

ROUHANA, S.W., LAU, I.V., and RIDELLA, S.A. (1985). Influence of velocity and forced compression on the severity of abdominal injury in blunt, nonpenetrating lateral impact. J. Trauma, 25, 490-500.

SHEAHAN, N.F., TUOHY, B., KIRKHAM, R., COAKELY, D.A., and MALONE, J.F. (1991). The fast-flush test: evaluation in radial artery catheter-manometer systems. Clin. Phys. Physiol. Meas., 12, 247-252.

STALNAKER, R.L. and ULMAN, M.S. (1985). Abdominal trauma—review, response, and criteria. Proc. 29th Stapp Car Crash Conference, SAE Technical Paper No. 851720, 1-16.

VIANO, D.C., LAU, I.V., ASBURY, C., KING, A.I., and BEGEMAN, P. (1989). Biomechanics of the human chest, abdomen, and pelvis in lateral impact. 33rd Annual Proc. of the Assoc. for the Advancement of Automotive Medicine.

WALFISCH, G., FAYON, A., TARRIERE, C., et al. (1980). Designing of a dummy's abdomen for detecting injuries in side impact collisions. 5th International IRCOBI Conf. Proc., 149-164.

DISCUSSION

PAPER: **An Intravascular Pressure Measurement Technique Applied to Impact Testing of Unembalmed Human Liver Specimens**

PRESENTER: ***Jessica Sparks, Ohio State University, NHTSA Vehicle Research Center and Transportation Research Center, Inc.***

QUESTION: *Guy Nusholtz, DaimlerChrysler*
Did you consider trying to control the temperature?

ANSWER: In the fluid that we're perfusing the organ with?

Q: … in the fluid, too, and the actual liver.

A: We did consider controlling the temperature in the fluid. Our plan is to use room-temperature saline, actually, in that entire pump system for our experimental set-up. And controlling the temperature of the liver, actually I haven't considered yet that that would be a good idea.

Q: That would probably have some aspect on the response of the liver to the impact type of phenomenon that you're going to see. Also, how did you compensate for the thickness of the catheters that went through the vein? As I saw, your equation looked like just 1-D approximation, but now you've got this fairly complicated phenomena because you've got a catheter that's coming through and it's going to produce interesting turbulence effects. Did you attempt to consider that?

A: The effective of many—Let me make sure I understand your question. So the presence of the catheter in the blood vessel would—

Q: Yeah, it's going to affect the pressure to some degree. So by measuring—The method of measuring changes the results that you could potentially get.

A: Yes, that's true. This is—We're actually in the early stages. If you have any thoughts on a good way to address that, I would be happy to hear them later. Basically, we used—One of the driving forces for using the catheter was just that they're small enough to fit into a lot of, into a variety of locations. We can position them pretty much wherever we want them in the liver, and they are designed for intravascular use. So, we were basically just doing the best we could with the instrumentation system that we have. I think any instrumentation system affects what it measures by just being there.

Q: If you're using the catheter and you have laminar flow, you may be able to approximate an effective diameter to take care of that particular problem, but that would require some experimental work to make sure that's correct.

A: Okay. Thanks.

Q: Thank you.

QUESTION: *Erik Takhounts, NHTSA*
I think I've missed the point of your research. Could you please repeat what the goal was? Once the information is there, how is it going to be used?

ANSWER: We're sort of developing a new technique to—What we want to do is have pressure readings inside the liver, inside the blood vessels of the liver as a result of impact. And if we use radiologic imaging, we can pinpoint the precise location of the catheter tip in how ever many different blood vessels that we have them in so that we can have, essentially, a 3-D map of a pressure wave traveling through an organ.

Q: And?

A: Yeah. And then, we want to—Ultimately, we want to see whether this, whether there's a relationship, a correlation between intravascular pressure and liver injury. So, we're going to be connecting the impact test.

Q: Okay. By the way, what kind of pressure do you measure? Do you measure static, dynamic or a combination of both?

A: It's gage pressure. It's dynamic. It's dynamic. Well actually, we're measuring both actually, if I—Let me show you the set-up. So prior to impact, this pressure transducer, which is connected to the hepatic artery opening…, those will be measuring essentially the static pressure before we've done any impacting. We've got these three trans—actually five could measure the dynamic pressure after the impact event.

Q: So you will be able to take into account there the actual velocity of the, of the fluid traveling through the, through those catheters. Is that right?

A: Do you say we will or we won't?

Q: The velocity effect, you know the dynamic effect that will change the pressure. Will you look into that in the future, too?

A: Yes. Definitely.

Q: Okay. Thank you.

QUESTION: *Jeff Crandall, University of Virginia*

I had a question. If we look in terms of injury as being caused by, say, stress or strain, certainly we could say that internal pressure would create higher; higher internal pressure might create higher stresses or strains. However, the injury will ultimately results from a combination of stresses due to inertial conditions, bowel recondition, and pressure will only play a role in that. So I'm wondering how you'll take the pressures you measure in, say, an isolated case and apply those in the body.

A: And apply them in the body. That's our second phase of this research once we have this technique developed into a satisfactory manner, we want to transition into testing impact in intact post-mortem human subjects. So, we can essentially tweak our design to better replicate what we're finding in the intact cadavers.