IMPACT RESPONSE OF THE HUMAN LOWER THORAX

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ABSTRACT

The objective of this study was to determine the dynamic force-deflection characteristics of the human lower thoracic region. Five unembalmed cadavers were pressurized and subjected to impact to the right lower anterior region of the thorax at a velocity of 4.3 m/s. A 230 N pendulum with 150 mm diameter face delivered the impact. The force-time history was obtained from the load cell attached to the pendulum impactor. Chest deflections at the impact site were obtained using the synchronized high-speed video camera operating at 2,250 frames per second. Peak forces ranged from 1.9 to 2.7 kN (mean: 2.29 kN ± 0.38). Peak deflections ranged from 76 to 125 mm (mean: 101 mm ± 17). Detailed autopsy revealed rib and liver trauma. The nonlinear force-deformation curves presented in this study provides fundamental data to design anthropomorphic test devices with improved biofidelity so that injury can be predicted in real-world crash environments.

INTRODUCTION

Anthropomorphic test devices (frequently called dummies or manikins) are routinely used to predict injury in simulated crash environments. One of the most important criterion for accurate injury assessment is the closeness of the mechanical response of the dummy to the living human, i.e., the biofidelity characteristics. The hybrid III manikin is the most widely used test device in vehicular crash simulations. This manikin was developed in the 1970's; it was based on the biomechanical data available to researchers and designers at that time [6]. Information from pendulum impact tests to the sternum of unembalmed and embalmed intact human cadavers formed the primary data set for the design of the thoracic region of the manikin [2]. Briefly, blunt impact tests were conducted using a 150 mm diameter rigid pendulum striker. The impact force was recorded using a load cell. Photographic analysis procedures facilitated the computation of chest deflections. Pathological observations included rib fractures of the thorax. The hybrid III manikin thorax is based on these force-deflection corridors.

This previous design was aimed at predicting injury to an unrestrained occupant in frontal impact. With the advent of restraint laws as well as design modifications in the restraint system (e.g., automatic two-point belts, airbags) the injury spectrum has changed. For example, one of the critical regions of the human body susceptible to serious/fatal injury in a two-point belt restraint system is the lower thorax-liver region. The loading of the diagonal shoulder harness to the lower thorax is a causal agent for these injuries which could be occult [1, 5]. Biomechanical studies on human volunteers and intact human cadavers have shown that the loading of the thorax is different between a regular three-point shoulder-lap harness combination and an airbag-knee bolster system [13, 14]. The response of the thorax of the hybrid III manikin is shown to be different compared to human volunteers [3, 4, 8]. Consequently, for this manikin to predict injury for the restraint systems currently in vogue, it needs refinement. To achieve this goal, appropriate dynamic force-deflection corridors applicable to the lower thoracic region must be developed using controlled laboratory experiments with human cadavers. The present study was designed to obtain these biomechanical data.

MATERIALS AND METHODS

A total of five unembalmed male human cadavers were used. The subjects ranged in age from 62 to 86 years, height 170 to 175 cm and weight 56 to 86 kg. All specimens underwent screening before procurement. They were tested for human immunodeficiency virus, and hepatitis A and B according to the standard guidelines [14]. Pretest chest radiographs were taken. The specimen was seated on a platform in the standard thoracic pendulum configuration. The torso was oriented fifteen degrees from the mid sagittal plane to receive dynamic loading from a pendulum. The lower extremities were fully stretched and the upper extremities were extended forward to allow proper positioning of the torso and document the kinematics during dynamic loading. The back of the torso was unsupported. Photo-targets were placed on the pendulum and the torso. High-speed video photography (Model 4250, Eastman Kodak, Rochester, NY) of the test taken at 2,250 frames/second documented the kinematics. The video camera was placed in the right lateral view to obtain the chest deflections. Dynamic loading was applied approximately at the level of the sixth rib anteriorly on the right side using a padded pendulum impactor at a velocity of 4.3 m/s. The impactor mass was 230 N.

The specimen was pressurized to approximate *in vivo* conditions. A balloon catheter was inserted into the femoral artery and inflated to prevent pressurization inferior to the abdominal region. A mixture of water and a thickening agent was pumped into the thoracic vasculature. In addition, the pulmonary system was pressurized. The vascular pressure for the test was recorded.

Each specimen was impacted once on the right lower thorax at an impact velocity of 4.3 m/s. Following the test, the specimen was palpated, radiographed and a detailed autopsy was conducted to evaluate the soft and hard tissue alterations. The uniaxial load cell (Model 1210 AO, Interface Inc, Scottsdale, AZ) attached in series with the impactor gathered the biomechanical force data during the test. The data was acquired according to SAE J211b specifications at a sampling rate of 12,500 Hz. The high-speed video photography unit and the digital data acquisition system (ODAS, DSP Technology, Inc., San Francisco, CA) were synchronized with a common trigger mechanism.

Data processing included digital filtering of the force-time information through a standard SAE Class 1000 filter. Chest deformation-time plots were obtained using the high-speed video frames. Every frame of the video was digitized and the deformation histories were plotted. A sixth or seventh order polynomial curve was fit to the deformation-time curve and this equation was used to obtain the deflection data corresponding to each data point on the force-time signal. In principle, eliminating the time variable (abscissa) between the two dependent variables results in the force-deflection response. This was accomplished using a force criterion as follows. On the force-time history plots, a nominal force level of 10 N was identified and a tangent was drawn to the curve at this magnitude of force. The intersection of this line with the x-axis (time -axis) was taken as the time at which the thorax begins to load. Using this time as the "t-zero" point, force-deflection curves for each specimen were obtained.

RESULTS

A review of high-speed video photographs indicated the following: Initially the face of the pendulum contacted the skin and the subcutaneous tissue at the impact site (right lower anterior rib cage). With further penetration of the impactor, deformations of the skeletal parts occurred resulting in rib fractures and/or liver lacerations with the concomitant action of the torso trying to move away from the impactor. However, the impacting face of the pendulum maintained its contact reaching the peak load. No secondary impact to the specimen occurred after pendulum rebound. Maximum forces ranged from 1.86 to 2.74 kN with a mean of 2.29 kN and a standard deviation (SD) of 0.38. Peak thoracic deflections ranged from 76 to 125 mm (mean: 101 mm $\pm$ 17). Approximately 1500 frames of the video were digitized to determine the deflection-time history. It should be noted these thoracic deflections represents the chest deflection including the rib cage and the overlying flesh/skin; it was measured as the change in the distance from the surface of the pendulum impactor to the photo target on the dorsal aspect of the specimen in a direction longitudinal to the line of travel of the impactor mass. As expected, force-deflection responses were nonlinear. Rib fractures (non displaced) and liver trauma were identified during detailed autopsy; liver lacerations occurred in two out of the five specimens..

DISCUSSION

Many impact biomechanical studies have been conducted in the past to determine the force-deflection response of the human chest from crashworthiness point of view. These include human cadaver experiments at injurious and non injurious levels, and human volunteer tests at the subinjury threshold. However, to the best of our knowledge, force-deflection data are not available to describe the biodynamics of the human lower thoracic region. This data is important as recent studies are aimed at redesigning the dummy to predict real world trauma in crash studies [10, 11]. This was the impetus for our research.

As stated in the Introduction, the hybrid III thorax design is totally based on blunt impact to the sternum of human cadavers and it is particularly applicable to an unrestrained occupant in a frontal impact. However, the dummy is being consistently used in vehicular product analysis and design with two-point belts, three-point belts, air bag-knee bolster restraints, or a combination of these systems. Researchers have recently recognized the need to redesign this manikin primarily due to the changing pattern of real world injuries secondary to the legislation as well as the variations in vehicle restraint system. Different types of restraint systems and occupant locations (driver, right front seat passenger) induce different modes of loading on the human thorax in a frontal impact. For example, the airbag can be a source for thoracic injuries for an out-of-position front seat occupant. Similarly, the diagonal harness in the absence of the lap belt has the potential to induce lower thoracic injuries including fatalities in certain instances. The force-deflection corridors developed from the symmetrical loading at the mid sternal region cannot be used as a replica for the lower thoracic region of the human due to the significant anatomical and structural differences. Consequently, fundamental data is needed to describe the biomechanical response of this region.

In this study, the biomechanical response was determined using dynamic loading directly on to the lower thoracic region of human cadavers. Although static/quasistatic loading mode could have been considered, pendulum type dynamic studies were conducted in this research because real-world simulations are dynamic in nature and static results cannot be directly extrapolated to impact situations. This methodology also allows for a direct comparison of the previous mid-sternal loading data which were conducted using similar techniques. The mid-sternal corridors reported earlier [7] generally lie above the force-deflection obtained in the present study suggesting a softening behavior of the lower thoracic cage compared to the sternum.

The rationale for selecting the 4.3 m/s velocity in our impact tests is as follows. The force-deflection corridors developed by Kroell et al used velocities of 4.3 and 6.7 m/s based on the analysis that an unrestrained front seat occupant interacts with the interior components of the vehicle at this speed during a 11 - 15 m/s (delta-V) frontal crash [7]. However, the restraints used in current vehicles considerably slows the occupant's forward velocity in a frontal impact. This is because, active restraints are generally in contact with the human body which will alter the occupant kinematics in terms of the time and severity of interior vehicle component interaction [12]. Occupant kinematic models (e.g., MADYMO, CVS) and analysis of high-speed human cadaver tests with belt restraints have indicated that the chest loading rates range from one to five m/s depending on the degree of slack present in the restraint prior to impact [10, 12]. Consequently, we chose the 4.3 m/s velocity to be representative of the occupant movement prior to loading. Furthermore, this velocity also facilitates a comparison with the earlier mid-sternum impact experiments.

All deflections were obtained by transferring every image of the high-speed video taken at 2,250 f/s to a computer using an image grabbing software. Because of the high-speed nature of the optical procedure, it was possible to obtain a detailed deflection-time history for each specimen. Since, the load cell data was gathered at a sampling rate of 12,500 Hz, in order to obtain a one-to-one correspondence between the deflection-time and force-time traces, the deflection-time signals were processed with a curve fitting technique. This procedure provided ordinates at identical abscissa in both traces. A force of 10 N was chosen to represent the loading of the lower thoracic region. This nominal low force level provided a consistent datum for deriving the force-deflection characteristics.

The deflections reported in the present study includes the skeletal and the soft tissues (e.g., overlying skin). The existing hybrid III manikin measures the "central" chest deflection through a potentiometer attached internally within its thorax. Consequently, the dummy deflections reflect only the skeletal part without contributions from the overlying skin and other soft tissues. The deflections presented in this study must be scaled down if one were to use the above methodology for determining the lower thoracic dummy deflections. Although Lobdell et al suggested a common decrement of 12.7 mm for the mid-sternum impact at a velocity 6.7 m/s, similar information, to the best of our knowledge, is not available for lower thoracic impact at 4.3 m/s velocity [9]. Therefore, no such recommendations are made to scale the present values. It may be appropriate to use dimensional analysis combined with mathematical models such as the finite element method to achieve this objective.

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DISCUSSION

PAPER: **Dynamic Characteristics of Lower Human Thorax to Pendulum Impact**

PRESENTER: N. Yoganandan, Medical College of Wisconsin

QUESTION: Guy Nusholtz, Chrysler Corporation

It appears that the difference between your force and your deflection-based force deflection is an experimental process. It is not necessarily something that is fundamental but merely the way the experiments are run. In one case, you've got a direct loading (a flat loading) and in the other case, you've got an off axis loading. Part of that could be because we use a one dimensional response to characterize really a three-dimensional phenomena. Could you comment on that?

ANSWER: The force deflection corridors traditionally specified for almost all the components of a dummy are based on a one dimensional response. So, to be in line with what has been done before, and to continue on to suggest argument of designs to provide fundamental data, we are stuck with the one dimensional analysis. That is why we had a load cell which was in one dimension so the analysis could be done only in one dimension. Did I answer your question?

Q: You sort of answered the question but I think the problem is, it's due to the fact that we are using the one dimension to characterize the dummy and so the distinctions that we come up with between force and deflections are artificial due to the method that we are using to characterize how the dummy is responding or how a person is responding. You've drawn a distinction but the distinction may be the way that we are handling that particular piece of information.

A: You are correct. The distinction also comes from the fact that you have little control over the anatomy of Joe vs. John and that is one of the reasons you are seeing this kind of a difference in the shift of the force deflection curve when you use two different criteria.

Q: John Cavanaugh, Wayne State University

Have you made any comparisons between the deflection obtained from film analysis to that of the chest band?

A: We looked at the peak deflections obtained from the chest band and the peak deflections obtained from the high speed video and we did not find a significant change in the deflections. In other words, the deflections were well within the acceptable range for biomechanical testing of cadavers.

Q: Richard Morgan, NHTSA

As chair, let me ask Yoga a question. Yoga, these impacts were into the abdomen such that you were going into ribs mostly, I think. If that's true, could you tell how you fixed the chest band over the anatomy.

A: The chest band was at the R8 level so whatever was the anatomy of that particular specimen, we would wrap it around the lower thoracic cage at the R8 level in the anterior region.

Q: How did you stick or fix the chest band to the outer skin? How did you have it keep it there?

A: The specimens were closed, so to speak, in a cotton clothing. Then we used double sided tape to fix the chest band to the clothing of the human subject, of the human cadaver, and then instrument it.

Q: Let's say that some of the experiments that Dr. Cavanaugh ran before were (what I'd call) straight into the abdomen and didn't go through a rib cage. Do you have any ideas how you would fix the chest band for that type of experiment, one where you are going into just soft tissue? What would you think?

A: I'd start off with what I'm doing now and see where I go and probably I would change it if needed.

Q: Could you foresee any difficulties?

A: Not standing here.

A: John Cavanaugh, Wayne State University

Methods to fix a chest band are probably work in progress. We've had some success with having sutures around the band at periodic intervals and also the method of using a Velcro. I think, Rick Morgan, that helps to keep it taut.

Q: N. Yoganandan, Medical College of Wisconsin

And you don't clothe the human cadaver. I mean you don't put any clothing on it? Just to the skin only?

A: John Cavanaugh, Wayne State University

Well, the suture goes through the clothing into the skin and through the subcutaneous tissue and back around again.