D. Review of Biomechanical Impact Response and Injury in the Automotive Environment

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Mr. Melvin submitted as a summary of the material presented in his talk the executive summary of Task B of Phase 1 of a DOT contract titled Advanced Anthropometric Test Device Development Program. His summary is titled, "Review of Biomechanical Impact Response and Injury in the Automotive Environment" and was co-authored by Kathleen Weber.

"This review includes literature through 1984 and is divided into chapters covering the following body regions: head, spine, thorax, abdomen, pelvis, and lower extremities. Each chapter includes information on anatomy; clinical injury experience; biomechanical response to impact; and injury mechanisms, tolerance, and criteria from laboratory studies. Each chapter also contains its own reference list and thus can stand alone as a review of the literature on that region of the body. Summaries of each chapter follow.

Head

The head is considered the most critical part of the body to protect from injury because of the irreversible nature of injury to the brain. In the Injury Priority Analysis, head injury constitutes nearly 45% of the total Injury Priority Rating (IPR, this percentage indicates the relative contribution of injuries in each body region to the total societal cost of all automotive injuries). Facial injury, however, accounts for an additional 10.5%. The costly facial injuries are primarily lacerations to younger occupants. While these injuries are not likely to be life threatening, the impairment to the individual from facial nerve damage and/or facial disfigurement as well as the need for reconstructive surgery make such injuries relatively costly to society.

A variety of mechanisms have been postulated for mechanical damage to the brain from impacts to the head. They include: (1) direct brain contusion from skull deformation at the point of contact; (2) indirect brain contusion produced by negative pressure on the side opposite the impact; (3) brain contusion from movements of the brain against rough and irregular interior skull surfaces; (4) brain and spinal cord deformation in response to pressure gradients and motions relative to the skull, resulting in stress in the tissues; and (5) subdural hematoma from movement of the brain relative to its dural envelope, resulting in tears of connecting blood vessels. The latter three mechanisms have also been postulated for mechanical damage resulting from head motions due to indirect impact.

The data presently available for defining the response of the head to impact are limited to rigid impacts and are predominantly based on embalmed cadaver tests. The data are adequate to define general response specifications for rigid impacts to the front and side of the head, in terms of peak contact force over a range of impact velocities from 1 to 8 m/s. The corresponding acceleration response data are limited to an impact velocity range of 1 to 5 m/s.

There is a need for additional studies to define the impact response of the human head using unembalmed cadavers with rigid impact surfaces and current acceleration measurement techniques. A repeatable and reproducible method for producing padded impacts also needs to be developed to allow cadaver studies to be conducted for padded head impact response definition.

The parameters of head motion that have been associated with the production of brain injury are translational acceleration, rotational acceleration, and rotational velocity. Of these, most attention has been given to translational acceleration in terms of developing head injury criteria. For direct impacts to the head, the Wayne State Tolerance Curve and the Japan Head Tolerance Curve, both based on head translational acceleration, are in close agreement. Injury criteria that have evolved from the tolerance curve approach would be expected to provide accurate assessment of injury potential during direct head impacts.

The Head Injury Criterion (HIC), based on the resultant translational acceleration of the center of gravity of the head, is the most commonly used method of evaluating head impact data. Statistical analysis of direct head impact cadaver test data has been used to define the relationship between HIC values and the probability of sustaining a particular level of injury, thus providing a continuous ability to interpret HIC values. A HIC level of 1000 was found to produce an expected 16 percent incidence of life-threatening brain injury to the adult population.

The validity of the HIC for long duration and non-contact head accelerations remains in question. Injury criteria based on head angular acceleration and angular velocity have been proposed for such situations, but they lack the extensive evaluation and review that has been given the HIC for short duration (less than 15 ms) head impacts. Mathematical models of the head hold promise for evolving into injury predictive models given proper development and evaluation. Simple models, such as the mean Strain Criterion (MSC), which are based on translational acceleration, have the potential for describing the dependence of the injury response on impact waveform and direction of impact. The application of the MSC to dummy head accelerations, however, remains to be developed. More sophisticated finite element models of the brain and skull have been developed, but their complexities and lack of validation have hampered their development into injury predictive models.

The response of facial structures to impact loads has been studied to a limited extent. The fracture and collapse of the facial bones during distributed loading significantly reduces the peak forces and resulting head accelerations in comparison to those produced by similar impact tests to the skull.

The tolerance of the facial bones to direct impact loading has been studied by a number of researchers, and fracture loads for individual bones and the whole face have been determined. The failure characteristics of facial soft tissues due to laceration from sharp edges have been studied, and rating systems for the assessment of the severity of the lacerations have been developed. There is a need, however, to study the mechanisms of lacerations to facial tissue due to blunt impact.

Spine

The vertebral column is the principal load-bearing structure of the head and torso and provides a flexible protective pathway for the spinal cord. Injuries that affect the function of the spinal cord can result in death, quadriplegia, or paraplegia. Despite these potentially serious consequences, the actual incidence of such injuries is relatively low, and thus they contribute probably less than 6% to the total IPR. (This figure is uncertain because NASS does not code the spine directly but rather incorporates it into the neck and back regions.)

The static and dynamic response of the head/neck system to indirect inertial loading at low crash severities has been studied extensively in volunteers and, to a lesser extent, in cadavers. These studies have included frontal, lateral, and oblique impacts. Specifications for suitable neck linkage systems, ranges of motion, and joint resistance characteristics are available from the published literature. Direct crown loading experiments have also produced data on the superior-inferior compliance of the cervical spine in cadavers.

The static midsagittal bending response of the thoraco-lumbar spine has been studied in volunteers for flexion and extension. Specifications in terms of overall rotation ranges and bending resistance characteristics of the rotation of the thorax relative to the pelvis have been produced. The equivalent dynamic data are quite limited but do indicate the presence of upper thoracic spine mobility with values similar to those for lower spine mobility.

The status of knowledge on the tolerance of the neck to loading is limited. Of necessity, all volunteer data are below the injury threshold. Additionally, injury mechanisms can be quite different than those mechanisms controlling response. Most injury threshold data are either based on cadaver tests or on reconstructions of accidents with instrumented dummies. As such, the threshold values are subject to the limitations associated with the surrogate used to obtain the data. These data sources have been used to develop limiting tolerance values for neck bending moments in midsagittal flexion and extension, axial compressive and tensile neck forces, and neck shear forces. No efforts have been made at this time to develop limit values associated with combinations of the various forces and moments. Corresponding studies of the tolerance of the thoracolumbar spine are not available. The only tolerance studies done on the thoracolumbar spine are those related to vertical accelerations.

Thorax

The thorax houses most of the body's vital organs and is thus the next most critical region, after the head, to protect from injury. Injuries to the chest constitute nearly 19% of the cost to society of injuries sustained by automobile occupants, as calculated using the Injury Priority Analysis. The nature of thoracic injury, however, is such that there are few long-term disabilities. In general, the victim either dies soon after impact or recovers completely.

The most critical injuries are those to the internal organs. In most experimental studies using cadavers, however, injury rating has been based on skeletal damage. As thoracic skeletal deflection increases under dynamic loading, the force resisting the motion remains somewhat constant. Further deflection begins to produce rib fractures, which can be followed by the sudden appearance of internal soft tissue injuries as the skeletal structure collapses. It is necessary, therefore, to be conservative in defining thoracic injury criteria in terms of deflection levels related only to rib fracture because of the instability of the thoracic structure under such conditions. Applied load by itself is also inadequate as an injury criterion, because of its insensitivity to increasing deflection in the force-plateau region characteristic of dynamic thoracic response.

Another factor that must be considered in defining thoracic injury criteria is the fact that thoracic response to impact loading is highly rate-sensitive. Viscous and inertial forces dominate the initial response, and elastic forces become significant only as large deflections of the system occur. Some forms of pulmonary and cardiac injuries have been found to occur only in conditions of high impact velocities with very little chest deflection. The rate of thoracic deflection as well as the degree of deflection can both be important parameters in describing the injurious effects of an impact to the chest, and they should both be considered in the development of general thoracic injury criteria.

In terms of response, the sensitivity of the thoracic structure to the rate of loading makes it difficult to interpret the findings from different types of experiments without accounting for this variable. For instance, the strip loading produced by the shoulder belt may produce an apparent stiffness that is lower than that produced by a flat circular impactor, due to differences in shape and area of loading. The rate of loading in shoulder belt tests, however, is usually much lower than that of the typical impactor test, thereby confounding the interpretation of shoulder belt interactions with the thorax. Impactor mass is a variable that can also strongly affect the apparent response of the thorax and must be accounted for when comparing experimental results.

Flat circular impactor tests tend to produce a characteristic thoracic force-deflection response that consists of an initial linear region, followed by a plateau region of almost constant force, and finally, if the impact has sufficient severity, a third region of increasing stiffness. This general form of response has been shown to be true for both frontal and side impact and with volunteers as well as cadavers. Thoracic structural rate sensitivity appears to be responsible for much of the initial stiffness and for the subsequent plateau in force as the rate of loading decreases during the impact. However, the distribution of load by the flat impactor surface must play some role in determining the response, since shoulder belt loading does not appear to produce the plateau region, even when loading rates are taken into account. Such local loading effects are not, however, well documented.

Because of the complexities of thoracic response, simple elastic structural representations are inadequate to guide the designer of mechanical analogues of the thorax. Instead, representation by means of spring-mass-damper models and/or transfer function approaches are necessary to provide the designer with the proper insight into the relative contributions of elastic, viscous, and inertial forces to the overall system response.

The three-dimensional structure of the thoracic skeleton and its contents requires deformation descriptors that are global in nature to provide an omnidirectional description of response. In the cadaver, this has been accomplished to some degree by the use of arrays of accelerometers on the periphery of the thorax. Similar or alternative methods of global response measurement will be necessary in the AATD to ensure adequate capability to assess injury potential in different directions and under different types of loads and loading rates.

Abdomen

The abdomen includes the organs and viscera below the diaphragm and above the pelvic girdle. Although there is little bony structure to protect these organs from blunt impact, injuries to this region contribute only 7.5% to the total IPR. Like the thorax, the abdomen can be the site of injuries induced by restraint systems themselves, including belts and steering systems. As far as the crucial organs are concerned, the liver, spleen, and kidneys are most frequently injured, and these injuries tend to be the most serious and life- threatening.

Injury mechanisms in the abdomen are thought to be primarily the result of deformation or penetration of the abdominal contents along with significant force or pressure generation in the deformed organs. In addition, solid organs, such as the liver, may undergo severe damage due to pressure generation alone at high impact velocities. There is evidence to show that these organs are viscoelastic, that the rate of loading is a crucial factor in injury causation, and that a compressive stress of 300 kPa (43 psi) will cause a superficial liver injury. Regarding dynamic response of the abdomen, the problem is complicated by the fact that there is a variety of surface geometries and component materials that can impact the upper abdominal area in a vehicle crash environment. In side impacts, however, the surfaces such as doors and armrests are somewhat well-defined, and dynamic load-deflection response curves do exist to a limited extent for lateral impact. Much more research data are needed, however, before abdominal response to impact can be fully quantified.

Pelvis and Lower Extremities

The pelvis is a bony structure that transmits the weight of the torso to the lower extremities during normal locomotion and supports the torso in the seated position. In an automotive impact environment, it can sustain injury from both frontal and side impact, and, during aircraft ejection or vertical falls, it is called upon to take the entire inertial load from seat-to-head acceleration. Injuries to the pelvis, however, contribute only about 1% to the total IPR. This structure is important, therefore, primarily for its response during load transmission.

The lower extremities constitute approximately one-third of the body weight, and, during normal locomotion, are required to withstand large dynamic loads. Injuries to the lower extremities of automobile occupants are rarely fatal but require significantly longer periods of hospitalization and lost working days than injuries to other body regions at the same AIS level. Even so, injuries to this region constitute only a little more than 5% of the total IPR.

The frontal impact response of the knee/femur/pelvis complex during seated knee impacts has been studied extensively. This research includes information on the acceleration-time histories, impedance, and effective mass. Other studies have defined the geometry of engagement of the knee into crushable padding. Load-deflection data are also available for subluxation of the tibia with respect to the knee joint. Lateral response of the pelvis has been studied for both impactor and flat-wall impacts and has been described in terms of force-time histories and pelvic acceleration-time histories.

Injury tolerance data for the knee/femur/pelvis complex consists primarily of axial loads in the femur.

Lateral loading tolerances for the pelvis are available in terms of forces and peak accelerations. For the femur, tolerance to lateral impact can be defined in terms of maximum bending moment as can the loading tolerance of the tibia in the transverse direction. There is also information on the strengths of the knee-joint ligamentous structures."