Injury Biomechanics and Crash Dummy

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Abstract: This paper is to discuss with a biofidelity problem between crash dummy and injury biomechanics.

The crash dummy is a human surrogate for the impact simulation. There are over forty different dummies for crash tests under divergent regulations and standards in the world. However, their biofidelity is still a problem for most of those dummi-es due to several reasons. To solve this problem, the new dummies have to consider with the injury biomechanics and criteri-a harmonization.

According to the injury biomechanics, it is impossible to make crash dummy as a human body. As a tool of measurement for crash testing, the crash dummy is suggested to develop by the concept of one world with one dummy such as THOR dummy for frontal impact and WSID dummy for side im Systems, Inc., Detroit, jzhu@ftss.com）pact.

Looking in future, the computer model will be a new direction to solve the biofidelity problem between the measurement of crash dummy and injury biomechanics. A computer simulation can replace human body and crash dummy for the cra-sh testing if it is developed under the injury criterion. Onetime dummy will be another direction to solve this problem for crash testing measurement in future. The paper will describe this concept and dummy development in detail.

Keywords: Biomechanics, Biofidelity, Dummy, Injury

1 Introduction

There are over forty different dummies (Figure 1) for the crash tests under a series of regulations and standards in the world. However, their biofidelity is still a problem for most of those dummies due to several reasons. The major difficulty is how to understand the injury biomechanics and how to design the crash dummy under the injury criteria with a series of probable materials. Another difficulty is the crash dummy has been requested under repeat uses. It is impossible to run tests of crash dummy as a human body if is asked for repeat uses.

Figure 1 A Family of the Crash Dummy

- Crash dummy or ATD (Anthropometric Test Device) has a biofidelity problem although it is tried to develop on injury biomechanics with the measurement of impact injury for human body under a series of regulations and standards.
- Injury biomechanics is an area of research primarily associated with the protection of the human body or vehicular occupant in here.
- It is to understand how a body region is injured in an impact and seeks to minimize injury through environmental

modifications.

- Injury biomechanics can be broadly categorized into four areas; they are:

a) Mechanisms of injury
b) Human response to impact
c) Human tolerance to impact
d) Development of human surrogates for impact simulation

- Each area plays a role in the design of automotive restraint systems and interior structures and in understanding how injury is caused so that effective countermeasures can be taken to minimize injury.
- According to a study, crash injury has become the leading cause of death for road traffic accidents in the world.
- The mortality cost of injury was over $47 billion per year in USA and over $2.4 billion per year in China (Ref. 1). Roughly intentional injuries comprised of 1/3 of all injuries and the automotive related injuries constitute one-half of all unintentional injuries.
- Evolution of injury biomechanics was started as a collaborative effort of H.R.Lissner and E.S.Gurdjian at Wayne State University, USA in 1929 (Ref. 11).

The crash injury is a foundation for the crash dummy development. This paper will discuss with it and use it for the development of next generational crash dummies as well as dummy models.

2 Crash injuries

All crash injuries of the human body can be divided into the injury of Head, Face, Neck, Thorax, Spine and Extremity.

2.1 Injuries to the head (Including in brain)

The brain can be injured by the processes fracture or impingement mechanism. Excessive acceleration by itself can cause brain injury through a variety of effects.

Relative motion between the brain and the skull can induce a wide range of debilitating effects. The periphery of the brain can be contused. The blood vessels leading from the brain and the skull can be ruptured by relative impact pressure. Internal brain matter can be sheared by relative motion between its parts. And the brain stem can be distorted by extrusion through opening at the base of the skull. Finally, excessive tensile stresses can occur on independent of any large brain displacement.

This usually takes place opposite the impact site and can disrupt a variety of brain functions

2.1.1 Types of head injury lesions

A. Scalp
   a) Bruise (leakage of blood from a vessel into adjacent tissue).
   b) Abrasion (traumatic removal of some outer layers of scalp).
   c) Laceration (cutting injury, tearing of scalp).
   d) Avulsion (extreme laceration causing peeling of whole scalp).

B. Skull
   a) Suture separation (diastalsis, more often in younger skulls).
   b) Indentation (e.g. ping-pong fracture, usually in younger skulls).
   c) Linear fracture (may occur at points remote from the impact location due to tensile stresses generated by sudden return of skull to its original shape after deformation).
   d) Depressed fracture (may be accompanied by perforation, fragmentation or communion of the skull).
   e) Crushed skull (massive communion, usually due to extreme static loading).

C. Extra-cerebral bleeding (local or diffuse)
   a) Sub-arachnoids hemorrhage.
   b) Epidural hematoma (with skull fracture in 90% of the cases).
   c) Sub-dural hematoma (with skull fracture in 50% of the acute cases – usually due to torn bridging veins).

D. Brain tissue damage (neural and/or vascular)
   a) Brain concussive injuries (including classic cerebral concussion and associated with increasing intensity and distribution of DAI, diffuse axonal injury).
   b) Brain contusions (bruises of brain tissues located at any site, e.g., cortical, intra-cerebral brain stern).
   c) Intra-cerebral hematoma (visible intra-cerebral blood clots).
   d) Cerebral laceration (visible tearing of the brain tissues).

The above listing of head injuries shows only their anatomic loci, while suggesting some of their physiopathologic effects, does not clearly indicate their effects on consciousness.

It is important to relate each injury type to its mechanism and to show how the functional effects follow. That is to connect the trauma input biomechanics to the physiopathologic response and its contribution to the outcome of the patient. The two tools used for grading head injury severity are:

- **Glasgow Coma Scale** is used by clinicians to record the intensity of coma and by repeated use can also show its duration.
- **Abbreviated Injury Scale** is a simple estimate of the threat to life caused by the injury at an early stage of the trauma and is used primarily by motor vehicle accident investigators.

2.1.2 Paradigm for head injury biomechanics

The following chart (Figure 2) shows the paradigm for head injury biomechanics.

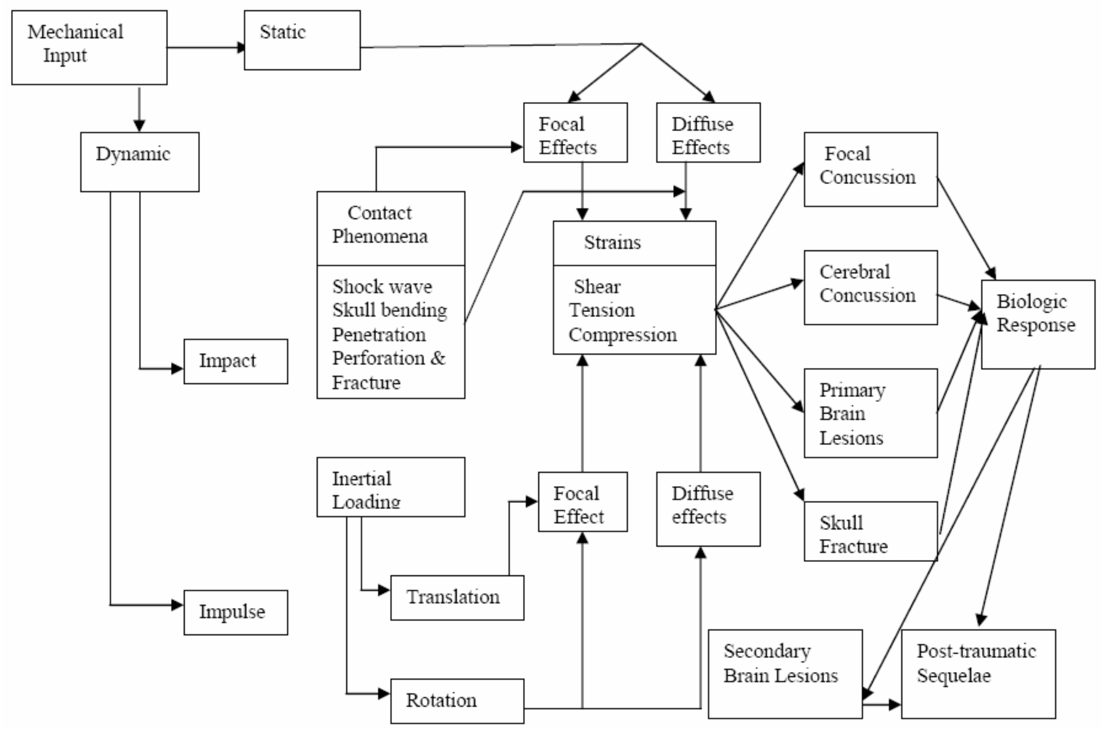

Figure 2 Head Injury Biomechanics

2.1.3 Concepts of head injury mechanisms

The above paradigm can be used to discuss the current concepts of how the various types of head injuries. Applied loads to the head may be either static (occurring at time duration's exceeding 200 ms) or dynamic (time duration's less than 200 ms and usually in the range of 5 to 50 ms). Static loads causing head injury occur infrequently. E.g., when the jack of an automobile fails and allows the car to crush the head of the victim trapped underneath.

Contact phenomena can be produced only by direct impact and inertial loading can be produced also by indirect impacts or impulsive loading of the head, for example, during violent flexion-extension movements of the head after a rear end car collision. Although shock waves are one of the contact phenomena, thus raising the possibility of diffuse brain injuries caused by this mechanism. But, it has been argued that the extremely short transit times for such waves (microseconds) are exceeded by an order of magnitude by the impact duration's and, therefore be ignored. Inertial loading results in two types of head motion: translation and rotation. The translation would be harmless and only the rotation could initiate the diffuse tensile and shear strains required to produce the diffuse effects required to generate the concussive brain injuries.

2.2 Injuries to the face

The principle facial bones are the mandible (lower jaw), maxilla (upper jaw), and the two zygoma (cheek bone). All are prominent and can be struck in a variety of locations from a variety of directions. In addition, these bones can be loaded individually or collectively depending on the size, shape and conformability of the impacted surface. The impacts were delivered to the most prominent feature of the bone and essentially normal to it.

2.2.1 Zygoma

An impactor (diameter 2.9 cm) was use to strike the zygoma of a cadaver. The studies employed blows to the frontal portion of the zygoma (near its junction with the maxilla). The results were similar and their findings can be summarized by the results reported by Nahum et al (Ref. 3).

a) The minimal tolerance load was 200 lb (0.89 kN), their recommended value for a clinically significant fracture was 225 lbs (1.00 kN).
b) Embalming the cadaver did not appear to impact the test results for the areas studied.
c) Thickness of the overlying soft tissue played and important role.
d) In another study, Hodgson (Ref. 4) explored the effect of increasing the area of the impactor. He conducted paired tests on five cadavers. The zygoma on one side of the face was struck with a 1 1/8 inch (2.9 cm) diameter impactor while the opposite zygoma was struck with a 29/16 inch (6.5 cm) diameter impactor. The average fracture loads were 283 lbs (1.26 kN) and573 lbs (2.55 kN) respectively.

2.2.2 Maxilla

The maxilla is the weakest of the facial bones when the impact is directed to the thin bone covering the maxillary sinus. Schneider et al (Ref. 5) reported that every one of the fractures in their maxilla study was "depressed and comminuted" due to the breakage of this bone shell.

Thirteen impacts were conducted (producing eleven fractures) with a 2.9 cm diameter flat impactor. Their average fracture force was 257 lbs (1.15 kN) and their fracture range was 140-445 lbs (0.62-1.98 kN). A previous study of the maxilla by Nahum et al (Ref. 3) had reported a range of 175-210 lbs (0.78-0.93 kN) as a "clinical fracture tolerance".

2.2.3 Mandible

The size and shape of the mandible presents a wide range of impact possibilities. Schneider et al (Ref. 5). noted a indeterminacy in delivering impacts to the center of the mandible. If the blow was directed towards the cranium, and the teeth were in contact, high forces could be sustained before failure occurred at the mandibular body or its symphysis. In these series of tests by Schneider, fractures occurred at all the three locations; the fracture force levels for the nine specimens tests ranged from 425-925 lbs (1.89-4.11 kN) with an average value of 639 lbs (2.84 kN) for the failures obtained. In a previous series of tests by Nahum had found a “clinical fracture range” of 350-400 lbs (1.56 to 1.78 kN) for impacts to the symphysis of the mandible.

Lateral impacts to the body of the mandible have been undertaken both with a 1 1/8 inch (2.9 cm) diameter impactor and a 1 x 4 inch (2 ½ x 10 cm) rectangular impactor aligned along the body. The study by Nahum obtained “lower fracture values” of 290-325 lbs. (1.29 to 1.44 kN) while the latter study by Schneider produced a fracture range of 184-765 (0.82 to 3.41 kN) and an average fracture load of 431 lbs (1.92 kN).

2.3 Injuries to the neck

There are three ways by which a neck injury can occur to the neck:

1) Direct impact to the neck
2) Neck injury due to head inertia loading
3) Neck injury due to head loading

2.3.1 Direct impact to the neck

The anterior portion (front) of the neck contains two stiff tissues which are delicate and vital. These stiff tissues, the thyroid and cricoid’s cartilages are found out at the upper end of the airway passage in the neck; hence their collapse can obstruct airflow. The thyroid cartilage is shaped like a wishbone with a relatively blunt apex. The cricoid’s cartilage is immediately beneath the thyroid. It is ring shaped and completely encircles the trachea.

2.3.2 Neck injury due to head inertia loading

The neck injuries can occur as a result of its bending from head inertial loading. When the Torso is violently accelerated or decelerated, potentially injurious neck loads and deflections are generated by the injuries of the head.

Neck bending can occur in any direction. In medical terminology, backward bending of the neck is called extension; forward bending of the neck is termed as flexion, sideward bending of the neck is called as lateral flexion. The “no” gesture of the head is termed as rotation. Following are he neck injury mechanism.

2.3.2.1 Hyperextension injuries and associated mechanisms

The rear collision accounts for most of the diagnosed neck injuries that occur to vehicle occupants.

The neck lesions are generally classified as hyperextension trauma and include symptoms such as:

1. Localized neck pain
2. Pain radiating to the shoulders
3. Vague aches
4. Discomfort
5. Vertigo due to strained muscles
6. Damaged ligaments
7. Injured articular joints
8. Fractures of various parts of the cervical vertebrae

2.3.2.2 Hyperflexion injuries and associated mechanisms

Hyperflexion injuries to lap/shoulder belted occupants have not been reported with any degree of frequency in the field accident studies. In a study by Vazy and Holt (Ref. 6) on the fatalities to car occupants wearing shoulder/lap belts indicated that only two out of the 136 fatalities were due to neck injuries. And in these two cases the occupant compartments were severely compromised.

A series of one hundred frontal collision sled tests is run with using lap/shoulder belted cadavers as vehicle occupants. Sled impact speeds of 30, 40 and 50 km/h were used. The cadavers were unembalmed with an age distribution at time of death ranging from 12 to 83 years. Forty-six of the 100 cadavers had neck damage with most of this damage being concentrated at the level of the C-7 and T-1.

Shear forces are important in the flexion prior to the chin contacting the chest. For the vertebrae C-3 through C-7, there are bone-to-bone interlocking joint surfaces and ligaments to carry these shear forces as the neck is flexed. This is not a case for the upper neck joints (occipital condyles/C-1 and C-1/C-2). Here the ligaments must carry the shear loads. These upper joints therefore are likely to be injured by shear.

2.3.2.3 Lateral flexion injuries and associated mechanisms

Lateral flexion injuries occur less frequently that the other two types of neck injuries. Usually in a lateral (side impact) collision, severe lateral flexion of the neck does not occur. For a far side collision, the upper torso is accelerated but may be free to rotate towards the impacted side, minimizing the neck forces required to accelerate the head. For a near side collision, the torso is accelerated upright, but the head usually impacts the side door window or upper side structures minimizing the neck forces.

If severe lateral flexion should occur, ligament injury and/or fractures of the articular processes of the vertebrae may be found at C-5 and C-7 level.

2.3.3 Neck injury due to head loading

The neck can be injured by loading of the head. During head loading, some or all of head load is transmitted to the torso by the neck structure. The magnitude of the transmitted load is dependent on the location and direction of the head load, the inertia of the head, and the configuration of the cervical spine when the head load is applied. For a given applied head load, the anterior cervical body strains were the lowest when the vertebrae were aligned; that is, the neck is straight. This implies that the neck behaved as a column and that the neck compressive load should be a good indicator of the potential for neck injury.

When the neck is flexed, the cervical vertebrae are subjected to a combined axial compression and bending moment. For this condition, the axial compression load alone may not be a good indicator of the potential for neck injury.

2.4 Injuries to the thorax (Including ribs, soft tissues)

The human thorax (or chest) is a ribbed shell (rib cage) containing the following important organs: heart, lungs, trachea, esophagus, and major blood vessels. The size and shape of the thorax depends on the age and sex of the individual, but roughly it may be described as a truncated cone with its depth less than its breadth. The rib cage is a semi-rigid structure which provides protection to the internal organs and facilitates the mechanics of respiration.

Thoracic injuries may be divided into injuries - the internal thoracic organs, the rib cage and the soft tissue.

2.4.1 Injuries to the internal organs

1. Pneumothorax.
2. Hemothorax.
3. Ruptures of the heart.
4. Ruptures of the arteries connected to the heart.
5. Injury to the cardiac muscle.
6. Lung contusion.
7. Bruising.
8. Rupture.

Of these the most frequent and the serious is the rupture of the thoracic aorta which is the major artery attached to the heart. Cardiac injuries are thought to be caused by compression of the heart between the spinal column and the sternum (breastbone). There is a increased possibility of cardiac rupture if the heart in that portion of its pumping cycle where it is full of blood. Tears of the aorta usually occur just beyond the aortic arch at its junction with the subclavian artery. The tears are usually traversed to the vessel axis.

2.4.2 Injuries to the rib case

1. Fractures of the ribs and sternum.
2. Dislocations and fractures of the thoracic vertebrae.

Rib fractures can become dangerous if the broken rib ends are displaced to the point where they can puncture internal organs or are numerous enough to inhibit adequate inspiration.

2.4.3 Injuries to the soft tissues

The development of injury criteria for soft tissue trauma is an extraordinarily complex subject which is only in its early developmental stage. Progress in this field is likely to be slow for the following reasons:

a) A wide variety of possible injury mechanisms exist.
b) Small difference in location or level or injury can have vastly different consequences to the injured person.
c) The capability to analyze and model the organs is very limited.

2.4.3.1 Skin: The procedures for assessing skin injuries are summarized in the SAE information report J202 (deals with Chamois, PPG formulation and Inland Skin and Flesh). One test procedure is to expose a synthetic skin material to a standardized impact test and then to evaluate the injury level either by subjective observation or measurement of the resultant damage to the synthetic material.

This is one of the practical ways to evaluate skin injury levels when the multiplicity of skin injury mechanisms is considered. Skin trauma includes:

a) Avulsion (tearing away)
b) Contusion (bruising from direct impact)
c) Laceration (cutting)
d) Puncturing
e) Splitting
f) Abrasion

2.4.3.2 Internal soft tissues: Internal soft tissues are vulnerable to all of the above types of trauma except for abrasion. In addition, they can be injured by excessive displacement which may detach an organ from its vascular or ligament connections. In the brain, rapid displacement may result in injury due to cavitations.

2.5 Injuries to the spine

Spinal injuries can be categorized as a sprain, disc disruption or vertebral fracture. A sprain refers to the stretching of soft tissue beyond its elastic limit (that is, it will not return to its original shape and size, even though the stretching is discontinued) so that it begins to tear.

Disc disruption most frequently refers to a tearing or compression of the inter-vertebral disc, possibly with the extrusion of some of the contents. The fracture of a vertebra refers to actual cracking or breaking of bones. These injuries are not due to some unique movement but rather due to extremes of the various loadings and movements which are part of normal functioning of the spine. These include tension, compression, flexion, extension, lateral flexion (lateral bending), rotation and their combinations. It is the spinal cord, contained within the framework of the bony spine that is of ultimate concern with regard to injury. It is when the spinal cord is damaged that paralysis and loss of sensation may occur. Thus, the vertebral fracture is not sufficient to cause neurological deficit.

The spinal injury and their consequences can be broken down into three groups:

1. The transection of the spinal cord in the lumbar region would generally result in at least a partial loss of lower limb

movement (paresis) and possibly a total loss of lower limb movement (paralysis or plegia). Typically, if the cord is transected above the L-2 level, ambulation, even with a cane or a brace cannot be achieved. Other possible consequences of cord injury include neurogenic bladder dysfunction (i.e., urinary bladder dysfunction due to injury to the nervous system rather that due to the direct injury to the bladder) and impotence.

2. The transection of the spinal cord in the thoracic region would mean paralysis of the entire region below the waist. This may also include the region between the waist and the mid- chest (this loss would still be considered paraplegia, since the paralysis affects the lower limbs but not the upper limbs).
3. The transection of the spinal cord in the cervical region would result in quadriplegia (sometimes referred to as tetraplegia). The term “quadraplegia” is used at times to indicate total paralysis of all four limbs, while the term “quadraparesis” is used to indicate partial loss of motion in all the four limbs. Cervical cord injuries, if in the upper cervical region, can have additional deleterious effect. The phrenic nerve, which controls the respiratory diaphragm (and hence, the ability to breathe), is usually a combination of nerve branches from the third, fourth and fifth cervical levels. If the cord is severed above this level, the patient would be unable to breathe and hence death would result.

Other injuries to the spine include in burst fracture, the wedge fracture, the fracture dislocation and subluxation. A burst fracture generally results from a compressive force on a “straight” vertebral column. Typically, a part of a disc is forced through the end plate (uppermost and lowermost boundaries of vertebral body, in the horizontal plane) of a disc and results in a comminuted (bone broken into at least three pieces) fracture of the vertebral body. The wedge fracture is also typically due to compression but, in that instance, it is compression acting on a flexed vertebral column and usually resulting in a fracture to the anterior superior region of the vertebra. A dislocation refers to a complete disruption of the joint, that is, the articular surfaces of adjacent vertebra are sufficiently displaced so that there is not contact between the two.

Since vertebras are bones, the articulation of adjacent vertebrae is a joint. If a joint is dislocated one or both the vertebrae forming the joint are fractured, the injury is sometimes referred to as a “fracture dislocation”. If the articular surfaces maintain partial contact, the injury is then referred to as “subluxation” or partial dislocation. Complete dislocation is called as “luxation”.

2.6 Injuries to the lower extremities (Including in pelvis, femur and tibia)

The structural elements of the lower extremities consist of the pelvis, femur, tibia, smaller fibulas and the ankle and the foot bones. In addition, there is the patella (bony knee cap), which cover the knee joints in front and serve as a terminus for ligaments and tendons. It can be divided into four cases for loading of the lower extremities.

2.6.1 Loading through the knee joint

It was conducted the lower limb studies to determine the strength of the patella/femur/pelvis complex in impacts simulating knees striking instrument panels by Patrick, Kroell and Mertz (Ref.7). The seated cadavers translated forward during sled deceleration to impact against four padded load cells. The head, chest and each knee struck a separate load cell. These load cells were geometrically arranged to simulate the forward surfaces of an automobile passenger compartment. These researchers concluded that the femur was slightly more vulnerable to fracture than the patella or the pelvis, but that distinction was too small to allow confident prediction as to which bone structure would fail first. Loads of around 1470 lbs to 1970 lbs (6.54 kN to 8.76 kN) were observed without fracture. Viano and Culver (Ref. 8) conducted sled tests with cadavers restrained by a shoulder-belt-plus-knee-bolster configuration (no lap belt). For six of these subjects the bolster was positioned so that the knees impacted it squarely. No injuries were produced for bolster loads which ranged from 1190-1800 lbs. (5.3 – 8.0 kN) per leg with an average of 1420 lbs (6.3 kN).

2.6.2 Loading below and across the knee joint

In the preceding studies the loading of the femur was primarily through the patella and/or femoral condoyle and resulted in patella, femur and/or pelvis fractures. If the loading is applied below or across the knee joint, damage to the knee ligaments and/or fractures of the tibia and fibula may result. Viano et al impacted seated cadavers on the anterior portion of the tibia just below the knee joint. They found that impactor forces ranging from 740 – 1550 lbs (3.28 to 6.89 kN) with an average of 1140 lbs (5.09 kN) produced knee ligament tearing and/or tibia-fibula fractures.

For the eight impacts which spanned the knee joint (involving both the patella and tibia), knee joint damage was produced for impactor forces ranging from 1330 to 1880 lbs (5.91 to 8.36 kN) with an average of 1580 lbs (7.02 kN). The predominant injury mode was avulsion of the posterior cruciate ligament from the tibia plateau. Sled tests in which the shoulder belted cadavers (without lap belts) impacted knee bolsters. The two below-the-knee leg impacts produced significant ligament tears at peak bolster loads of 790 lbs (3.5 kN) and 940 lbs (4.2 kN) per leg.

2.6.3 Static tests of the knee joint

Viano et al also performed low-speed ligament tolerance tests on five isolated knee joints mounted in a universal testing machine. In these tests, the knee joint angle was maintained at 90 deg while the tibia was displaced rearward relative to the femur until complete joint failure occurred. Loads corresponding to the initiation of joint failure ranged from 320 lbs. (1.43 kN) – 575 lbs (2.56 kN) with an average of 455 lbs (2.02 kN).

The corresponding displacement of the tibia relative to the femur at the initiation of joint failure ranged from 0.37 inches (9.5 mm) to 1.18 inch (30 mm) with an average of 0.57 inch (14.4 mm). The load corresponding to complete joint failure ranged from 375 lbs (1.67 kN) to 675 lbs (3.0 k) with an average of 560 lbs (2.48 kN).

2.6.4 Concentrated loading of the patella

The unique construction of the patella makes it vulnerable to concentrated loading. This phenomenon was studied by Melvin et al. (Ref. 9), employing three different impactor sizes, all unpadded. Two of the impactor were flat surfaced circular areas with diameters of 0.61 inch (15.5 mm) and 0.43 inch (10.9 mm), while the third impactor was ring shaped with an outer diameter of 0.50 inches (12.7 mm) and an inner diameter of 0.25 inch (6.4 mm).

3 Injury measurement in the crash dummy

According to the injury biomechanics, it is impossible to make crash dummy as a human body due to its repeat uses. As a tool of measurement for crash testing, the crash dummy is suggested to develop by one world with one dummy such as THOR dummy for frontal impact and WSID dummy for side impact.

There are examples for the development of new crash dummies. The THOR dummy will be a new generational dummy for the frontal impact and the WSID dummy will be a new generational dummy for the side impact. They are both developed by global designers under international harmonization.

3.1. Head tolerance leverls and injury criteria

3.1.1 Wayne state tolerance curve

The Wayne State University tolerance curve was first proposed by Lissner et al in 1960 (Ref 10). The abscissa is the duration of the effective part of the pulse spanning the principle impact. The ordinate is effective acceleration which is the average a-p (Anterior-posterior or front to back) acceleration of the skull measured at the occipital bone for the principle part of the impact of the forehead against plane, unyielding surfaces. This curve was derived from the following observations.

(a) It was observed clinically that the linear skull fracture is usually associated with unconsciousness or a mild concussion as reported by Gurdjian et al (Ref 11).

(b) The acceleration levels and pulse durations necessary to cause skull fractures in cadaver heads were measured in free fall impacts against a rigid surface. These results were considered to approximate the human tolerance level for concussion from the correlation. The fracture data provided points for the curve in the range of 0.001-0.006 s.

(c) Animals were concussed by air pressure pulses of varying magnitudes and durations applied directly to the membranous covering of the brain, Gurdjian et al (Ref 12).

(d) The pressure pulses measured in the parietal and temporal regions of the cadaver heads in the drop tests, Lissner et al (Ref 10) and Gurdjian et al (Ref 13) were compared with the animal data in the item (c), and the corresponding cadaver acceleration measurements were used to provide data points for the concussion curve between 0.006 and 0.010 s.

(e) The long-duration end of the curve, with the asymptotic value of 42 G, was obtained from whole body volunteer data reported by Stapp (Ref 16 & 17). Patrick et al. considered this value to be too low, since other volunteers had survived frontal crash simulations exceeding 45 G. They recommended that the value of the asymptote be raised to 80 G for padded impacts that avoid concentrated loads (Ref. 16).

3.1.2 Severity-Index (SI):

The Wayne State Tolerance Curve is difficult to apply to complex acceleration-time pulses because of uncertainties in determining the effective acceleration and time. To overcome this problem Gadd (Ref. 17) devised a weighted impulse criterion for establishing a Severity Index (SI):

$$SI = \int_0^T a^n dt$$

a = acceleration in G's
n = weighting factor, 2.5 for head impacts
T = pulse duration
t = time in seconds

The weighting factor of 2.5 is primarily based on the slope of the straight-line approximation of the Wayne State Tolerance Curve plotted on a log-log paper between 2.5 and 50 ms.

The influence of this impact criterion is shown by a 50% reduction in fatality incidents (normalized) when comparing the post 1971 seasons to the preceding equivalent period.

3.1.3 Head injury criterion (HIC)

The new head injury criterion (HIC) was defined as;

$$HIC = [(t_2 - t_1)\{(1/(t_2 - t_1)\int a(t)dt\}^{2.5}]_{\max}$$

Where $t_2 - t_1$ are the initial and final times (expressed in seconds) of the interval during which HIC attains a maximum value and a(t) is the resultant acceleration (expressed in G) measured at the head CG. The HIC replaced the SI in later versions of FMVSS 208 with a HIC value of 1000 being specified as the concussion tolerance level.

3.1.4 Lateral tolerance of the brain

Some lateral studies were carried out to determine the tolerance of the brain to lateral impacts. Some lateral studies employing cadavers and primates have been reported by Stalnaker et al. (Ref.18). They concluded that the threshold of irreversible closed skull brain injury to humans occurred when the translational head acceleration reached a peak of 76 G with the pulse duration of 20 ms. Melvin et al. (Ref. 19) investigated lateral impacts to unembalmed cadavers against rigid deformable structures. They found, for head impacts against rigid walls. The brain damage of AIS4 or greater began to occur at head impact speeds of 20 mph (33 km/hr).

Figure 3 is a head and neck of THOR dummy which is designed under biomechanical corridors. Its face mask can be taken off after testing and its injury level can be measured by a series of accelerometer, tilter and load cell.

Figure 3 Crash Dummy, Head and Neck (THOR)

3.2 Neck injury assessment values:

- NECK BENDING
  - Neck Flexion Moment Injury Assessment Value (Ref. 20 & 21) MF = 190 Nm
  - Neck Extension Moment Injury Assessment Value (Ref 20, 21 & 22) ME = 57 Nm
- NECK TENSILE, COMPRESSIVE AND SHEAR FORCES

  Injury Assessment defined by time dependent curves

Axial Tensile Neck loading	-	Ref. 23.
Axial Compressive Neck Loading	-	Ref. 24.
Fore/Aft Shear Force	-	Ref. 23.

3.3. Chest injury assessment values:

- RESULTANT CHEST SPINE ACCELERATION
  - Injury Assessment Value 60 G, 3 ms

If the resultant of chest spine acceleration is less that the 60 G, 3ms criterion, then significant thoracic organ injury due to gross chest acceleration is unlikely.

- FORE/AFT CHEST COMPRESSION
  - Shoulder Harness Assessment Value 50 mm

If the gross chest compression is less than 50 mm, then significant liver and/or spleen can be injured if the deflection of the lower portion of the lateral aspect of the rib cage produced by the shoulder belt's interaction is too great. Gross chest deflection provides an indirect measure of the magnitude of this interaction.

- Distributed, Frontal Chest Loading Injury Assessment Value 75 mm

If the gross chest compression is less than 75 mm, then significant thoracic organ injury due to distributed loads applied over large frontal chest areas is unlikely. Life- threatening injuries are expected if 75 mm is exceeded. This Injury Assessment Value is an Injury Threshold Level.

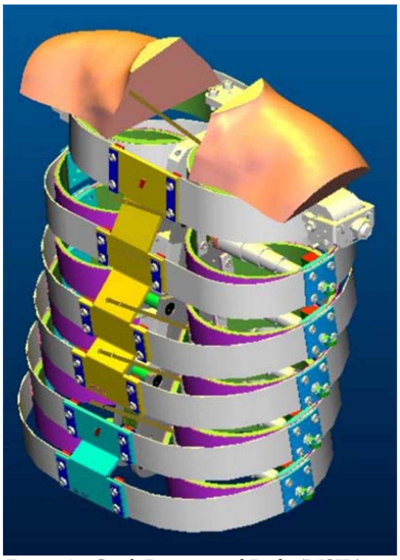

Figure 4 Crash Dummy and Body (WSID)

The thorax at Figure 4 is a design of WSID dummy which is considered with the human injury corridors. Its rib can be kept the shape with plastic deformation after testing and its injury criterion can be measured by a series of accelerometer, potentiometer and load cell.

3.4 Lower extremities measurements injury assessment values

- Injury Assessment Value for the knee loading (Refer Figure 5 below)

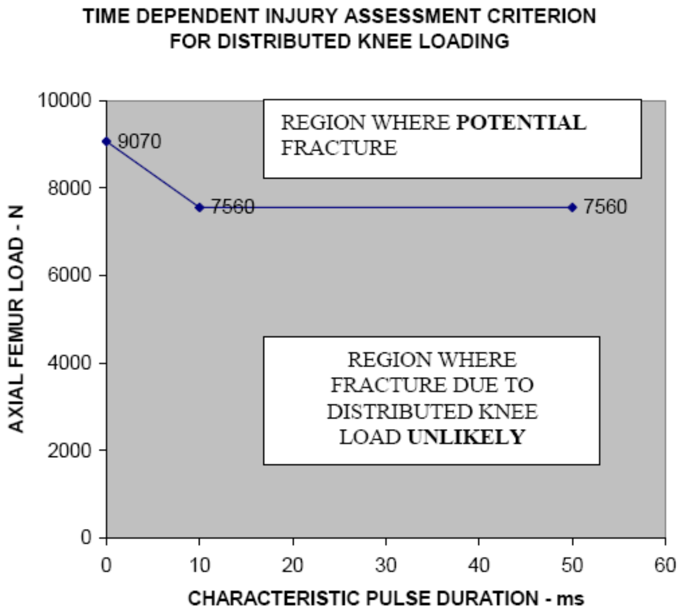

Figure 5 Assessment Value for the Knee Injury

If axial compression femur load is below the time dependent curve, then fractures of the patella, femur or pelvis due to well distributed loads applied to the knee are unlikely.

- RELATIVE TRANSLATION OF FEMUR AND TIBIA AT KNEE JOINT
- Injury Assessment Value 15 mm

If relative translation is less than 15 mm, then rupture of the posterior cruciate ligament of the knee joint is unlikely.

- COMBINED BENDING AND AXIAL COMPRESSIVE LOADING OF LEG
- Injury Assessment Value:

$$\frac{M}{M_C} + \frac{P}{P_C} = 1$$

Where Mc = 225 Nm, Pc = 35.9 kN

If the combined loading expression is less than one, then the fractures of the tibia and fibula shafts are unlikely. M is the resultant bending moment and P is the corresponding axial compressive force.

- MEDIAL AND LATERAL TIBIAL PLATEAU COMPRESSIVE FORCES
- Injury Assessment Value 4000 N

If the medial or lateral knee clevis compressive load is less than 4000 N, then fracture of the medial or lateral aspect of the tibia plateau due to femur/tibia interaction forces is unlikely.

- MEDIAL AND LATERAL ANKLE COMPRESSIVE FORCES
- Injury Assessment Value 4000 N

If the medial or lateral ankle clevis compressive load is less than 4000 N, then fracture of the medial or lateral aspect of the ankle is unlikely.

- KNEE LACERATION
- Injury Assessment Value - Moderate to major cuts of the interlayer of the two layers of chamois covering the knee.

It is potential for severing tendons or ligaments surrounding, the knee joint or fracturing of the knee cap.

4 The new dummy under the injury biomechanics

Regarding the injury levels in real accidents, the new dummy can be developed under the injury biomechanics as close as possible if the material property can be closed to corridors (Ref.25).

According to the injury biomechanics, it is impossible to make the biofidelity of crash dummy as a human body. As a tool of measurement for crash testing, the crash dummy is suggested to be developed by the concept of one world with one dummy such as THOR dummy for frontal impact and WSID dummy for side impact.

4.1. The correlation between THOR dummy and biomechanics

The THOR dummy is a frontal crash dummy of the next generation (Figure 6). It has been made two versions so far. THOR dummies have been evaluated in different regions in the world over several years. Its correlation is the best than other frontal dummies for the biofidelity.

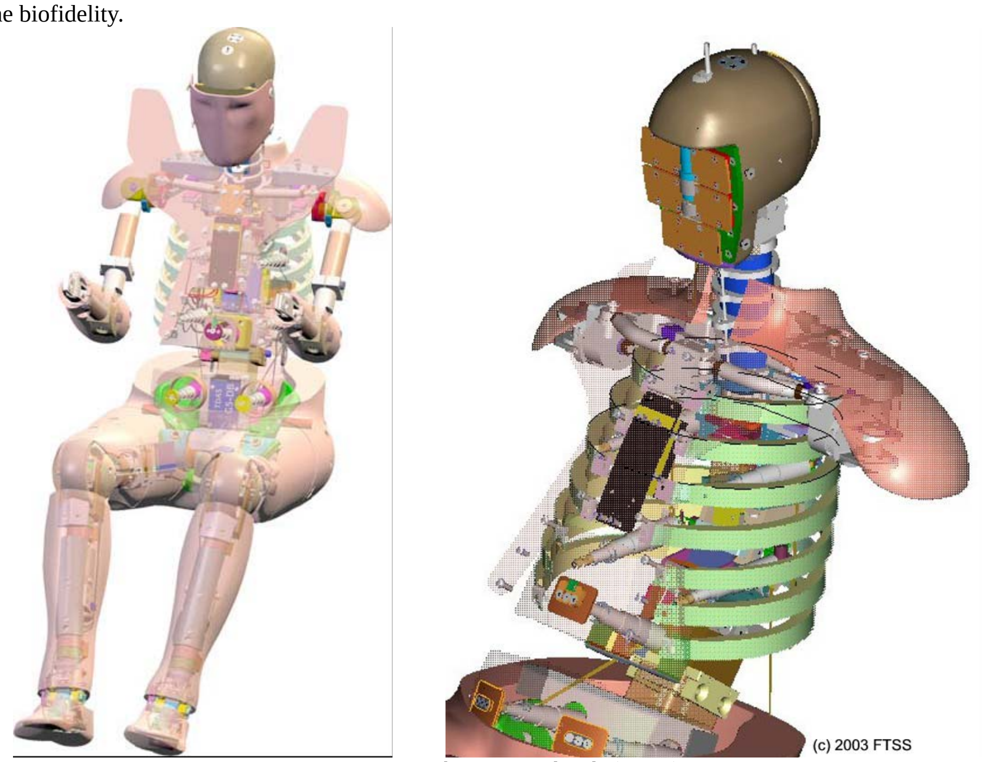

Figure 6 Crash Dummy and Body (THOR)

4.2. The correlation between WSID dummy and biomechanics

WSID – World Side Impact Dummy (Figure 7) is developed by world exports of crash dummy. It has included WSID-50 and WSID-5 dummies. Its biofidelity is high as 7.3 (Table 1). WSID dummy is the best dummy for side impact tests of the next generation.

Figure 7 Crash Dummy and Body (WSID)

Table 1 Dummy biofidelity Comparison (WSID)

Body Region	SID	ES-2	Biosid	WorldSID prototype	WorldSID prototype Rev. 1
Head	0	5	6.7	5	10
Neck	2.5	4.4	6.7	3.6	5.2
Shoulder	0	5.3	7.3	5.8	6.7
Thorax	3.1	5.2	6.3	6.9	7.7
Abdomen	4.4	2.6	3.8	6.5	6.6
Pelvis	2.5	5.3	4	5.4	7.3
Overall	***2.3***	***4.6***	***5.7***	***6.2***	***7.3***

- ISO Biofidelity Rating Scheme. 1 to 10. Higher is better.

4.3. The correlation between computer model and biomechanics

The computer model (virtue dummy) has been developed and used for automotive crash. It is not liked as physical dummy for a real barrier or sled test. The virtue dummy can be used for virtue crash tests under computer simulation. This new methodology has been introduced into the

automobile crash tests rapidly. It is not only for research and academic fields, but also for all of the automotive manufacturers in worldwide. The virtue dummy is also divided into Crash Dummy Model and Human Body Model. Their parameters of computer model must follow their biomechanical corridor as Figure 8.

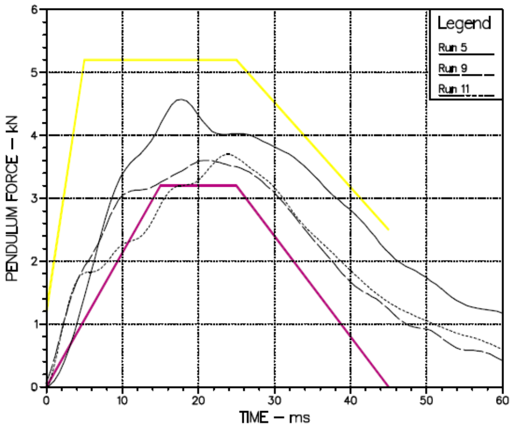

Figure 8 – Biomechanics Corridor for Human Body (Thorax)

The computer simulation has been used for crash testing and dummy design for a decade. It is more powerful for this area after the computer technology has been developed dramatically. The Finite Element Analysis (FEA) model and Rigid Body model have both been the trend in virtual crash dummy design. The predictive capabilities of both models allow engineers to fully understand a crash event in a virtual environment, thus limiting the number of physical tests that need to be executed. It has been supplied to the crash safety CAE community with innovative virtual dummies available in the different codes such as LS-DYNA, MADYMO, PAM-CRASH, and ABAQUS for the different applications (Figure 9).

Computer simulation of human body will be a direction to solve the injury biomechanics problem of crash dummy due to its accuracy (Ref.26) although it needs to have more biomechanical knowledge.

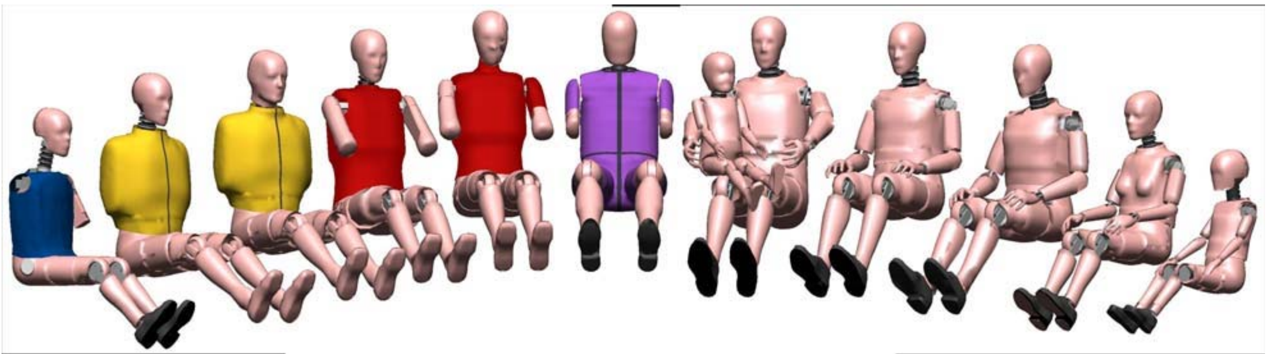

Figure 9 – A Series of Computer Models for the Crash Dummy and the Human Body

To solve the contrary between material property and injury biomechanics, onetime dummy (OD) will be also considered as another method to solve the biofidelity problem for crash testing measurement in near future because it needn't to concern with the repeat using conditions. Onetime Dummy can be used for the crash tests, earthquake rescue research, land mine explosive protection and other rescue trainings. The dummy body can be used with onetime and an instrumentation box can be used repeatedly.

5 Conclusions

The relationship is very important between the injury biomechanics and the crash dummy. After a series of discussion and analysis, several conclusions can be listed as following:

- Injury biomechanics is a key factor for the crash dummy research and development.
- It is impossible to design and manufacture a crash dummy as a human body.
- Computer simulation of human body will be a better direction to solve the injury biomechanics problem of crash dummy due to its accuracy of the injury biofidelity.
- Onetime dummy can become another method to solve the biofidelity problem for crash testing measurement in near future because it needn't to concern with the repeat using conditions.

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