MODELLING OF DIRECT HEAD IMPACT INJURY MECHANISMS APPLIED TO TRANSPORT AIRCRAFT: ARE LONG PITCH SEATS SAFE?

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ABSTRACT

This paper describes recent AmSafe Aviation research conducted in cooperation with Cranfield Impact Centre. The research addresses questions raised during consideration of safety regulations FAR/JAR 25.785 and 25.562 for the design of modern seat and restraint systems. Finite Element (FE) techniques with human models were used in conjunction with traditional dynamic testing to assess the efficacy of existing test methods and tools.

The subject of the research is long-pitch passenger seats which use a “no strike” compliance approach for the Head Injury Criteria (HIC). The research objective was to first determine if the potential for head to leg impact during a survivable crash event could result in severe injury or death within the context and limits of FAR/JAR 25.785 and 25.562. Because human models are not routinely available for design evaluations, the second objective was to assess the capability of the current regulatory test devices to provide meaningful injury response data that can be used to consider the head to body impact injury mechanism.

The explicit FE computer code LSDYNA was used to create a complete virtual seat, restraint, and occupant system with a human head model and human leg material properties. The human head model is validated against existing biomedical data. The system kinematics represent a typical economy class seat with a 50% male occupant, based on dynamic tests. A parametric study was used to evaluate the factors of impact severity, seat back recline angle, and restraint slack. The response from a baseline simulation plus eight additional simulations support the conclusions.

The potential for injury from head to leg impact with long pitch seats was found to be severe or lethal using the 16 G longitudinal regulatory impact pulse. The HIC response was found to correlate well with human injury response measures, and was also found useful in evaluating seat design and impact severity factors. It is recommended that HIC be recorded and reported for all dynamic compliance tests. This will provide regulatory authorities with the data needed to assess the safety of long pitch seats, and provide the means to evaluate the relative safety of design trends.

INTRODUCTION

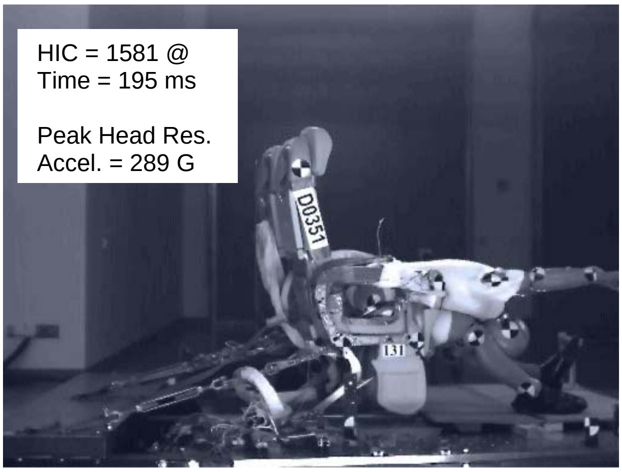
FAR/JAR 25.562 added specific dynamic performance requirements to the existing 25.785 safety regulation. [1] Regulation 25.785 is more general and establishes the intent of the safety regulations, while 25.562 provides specific limits. It quantifies both the limit of protection afforded passengers to remain conscious and ultimately evacuate the aircraft and the limit of the survivable impact. The three measured occupant injury responses (head, spine, femur) have become critical seat and interior design parameters, in effect, representing a “safe” environment.

The limit criteria for spine and femur have been straightforward to satisfy, but the HIC limit value of 1000 units has been a challenge. [2] HIC is a computed value based on the head resultant acceleration of the Anthropomorphic Test Dummy (ATD), and is thus affected by the occupant’s movement and interaction with the surrounding environment. Two compliance options for HIC have evolved for the selection of aircraft interior layouts.

The first option is to stop upper torso articulation early and produce minimal head velocity. This can be done with close seat pitch combined with seat-backs designed to absorb energy, upper torso restraint, or aft facing seats. The second option is to move vertical strike hazards (interior furniture) out of the head path envelope. FAA Advisory Circular 25.562-1A does not require the applicant to measure HIC for these cases. [3] The inference is that this is safe and results in a meaningful compliance to 25.562. The difficult question with no-strike compliance is that the occupant's head can strike the occupant's legs with sufficient velocity to be injurious or even lethal. This has become the preferred means of compliance for special economy (front row, exit rows), business class, and premium class seats. Current estimates run between 20% and 30% of total transport category seat placements.

Dynamic tests such as the baseline test shown below suggest severe. This test yielded an ATD head resultant acceleration with a peak value of 289 G's, and HIC = 1581. [4] As this compliance option becomes common, it is the responsibility of industry and regulators to take a critical look at the no strike means of compliance and consider if this satisfies the intent of FAR/JAR 25.562 and ultimately 25.785. This research seeks to provide the information to address these questions.

Baseline Test for Study, D0351



OBJECTIVE

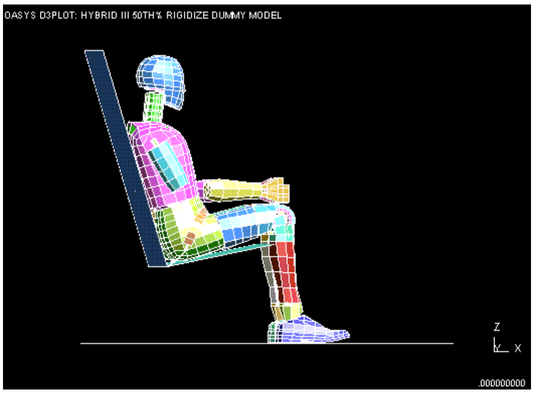
The research objective was first to determine if the potential for head to leg impact during a survivable crash event could result in severe injury or death within the context and limits of the safety regulations. Understanding the true nature of the head injury severity potential required consideration of the human tissues ability to absorb energy and evaluation of direct human response measures. Aircraft system survivability expertise and Finite Element (FE) modelling were beyond AmSafe's capabilities, and so proposals were solicited from several organizations in both the USA and Europe. Cranfield Impact Center (CIC) was selected, and the research was conducted as a team. The work was performed in five phases during the calendar year 2003 and first quarter of 2004. Follow up work, phase 6 in summer of 2004, refined the brain pressure injury response. CIC conducted all finite element modelling while AmSafe conducted the sled testing and parametric studies. Decisions on protocols and literature survey were made by mutual agreement.

Extensive FE analysis is not practical during the typical seat and interior design process. Thus the second objective was to assess the capability of the current regulatory test devices to provide meaningful injury response data for use in seat and restraint design relative to the head to body impact injury mechanism. This was accomplished by evaluating the efficacy of HIC as compared to the direct injury response measures. A parametric study with impact and seat design factors produced a wide range of results.

SEAT SYSTEM MODEL

A typical dynamic sled test was selected from 12 AmSafe tests evaluating a variety of factors (seat, restraint, ATD, yaw angle). [4] Designated by AMSAFE sequence number D0351, the test was at 10 deg Yaw and used the Hybrid III. The Hybrid III provides the most injury response data and also best matches available FE ATD computer models.

CIC FE D0351 ATD Model



CIC created a finite element model of the seat system. The setup parameters and impact were based on the seat drawings and data provided by AmSafe. Variation in contact surface friction resulted in small differences in the articulation timing between the sled test and simulation. This model is intended to generate typical kinematics response for evaluation of injury. Further refinement of frictional coefficients was not pursued in favor of developing representative human properties. All models are in the LSDYNA code with components as listed below.

FE Mesh ATD: Existing LSTC Hybrid III

Seat and Floor: 4-noded shell elements, mid-surfaces in center planes of real structures with seat base, seat back and floor connected through rigid links

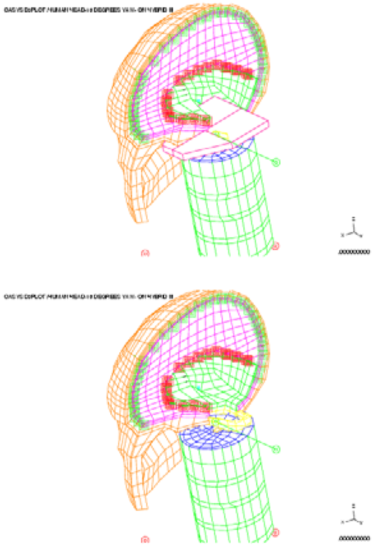
Lap Belt Restraint System: Pre-processor belt-generation facility, with interactions between ATD and Belt included

The Loading and Boundary Conditions are representative of the full-scale laboratory sled test. The actual sled acceleration pulse was imposed as the forcing function. The Model used 9 different material types, contains 3,205 Nodes, 1,434 Shell Elements, and 2,456 Solid Elements. The model also contains 22 Contact Interfaces and 114 different materials. [5]

HUMAN HEAD ATTACHED TO ATD

The human head model was used to replace the FE head of the ATD. Originally it was not known if a reasonable kinematics response would be achieved. An alternate approach using the ATD head to establish the impact criteria and then apply it to the human model was considered.

Head Attached to ATD



However, similarities between the human and ATD head and convenient interface points provided a means of direct attachment. The CG is positioned correctly with excellent kinematics response. The human head FE model was attached to the dummy neck as shown, by connecting the foramen magnum rigidly to the C1 plate.

HUMAN LEG PROPERTIES

The impacts were found to occur between the head and the lower leg. Appropriate consideration of the human impact characteristics requires a human like leg as well as head. It was beyond the scope of the research to use a full human leg model, thus the material properties of the ATD leg were changed to represent human tissue.

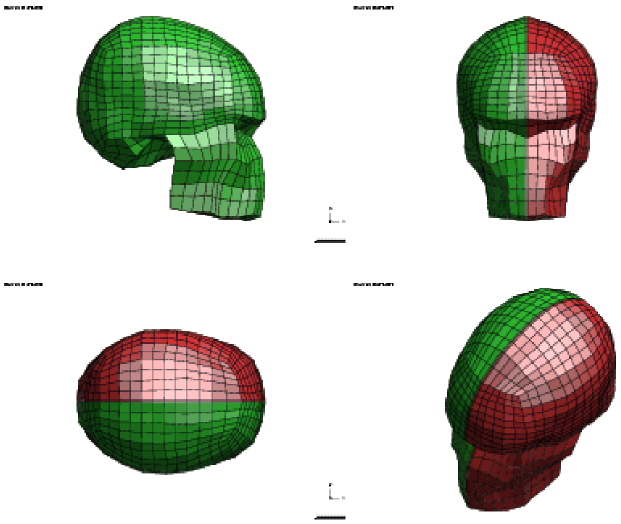
The contact force generated as a result of the head impacting the lower legs is a function of the Young's modulus of each of the two parts. The leg soft tissue is represented by the rubber/foam skin of the ATD and the tibia bone is represented by the round bar tibia of the ATD. The properties of soft tissue were found to be in the range already used for the rubber/foam, and were not changed. The Young's modulus of the tibia bone was defined as 2650 MPa, a common value cited for compact bone in literature. [6,7] This is a significant reduction from the original ATD value of 5000 MPa.

HUMAN HEAD MODEL

The head model used was developed by Taleb at Cranfield Impact Center [5]. Human material properties, geometries and boundary conditions have no universally accepted standards, and thus were chosen by the creator of the head model based on published post-mortem data. The model consists of

6,168 solid elements, 2,002 shell elements and 118 spring elements.

Taleb FE Head Model



The Taleb model was adjusted to achieve comparative results between the ATD and the human model. However the approach was to adopt minimal changes and all default parameters were kept unchanged. The scale of the model was slightly increased to match the Hybrid III. The mass (Skull density) correlated well, with the human head 4.33 kG vs Hybrid III 4.40 kG. The center of gravity of the human head is 15 mm in front and 10 mm above the Hybrid III cg position.

Validating the model necessitated the following calculation process changes: integration option modifications, element stabilization, more robust internal contacts, and re-meshing option changes. The elastic modulus of the skull was reduced to account for the scalp, which was not included in the original model, and found to add 6 mm cushioning material. The model components are summarized below.

Skull: single material by several layers of solid ‘brick’ elements

Dura: single layer of thin shell elements 1 mm coincident with the inner Skull

Subarachnoid Fluid: surrounds Brain, solid elements using fluid material properties

Brain: solid elements with visco-elastic material properties

Falx: separates the right and left sides of the main Cerebral Hemisphere

Membrane: layer of shell elements of elastic material properties

Foramen Magnum: opening in base of Skull , solid elements

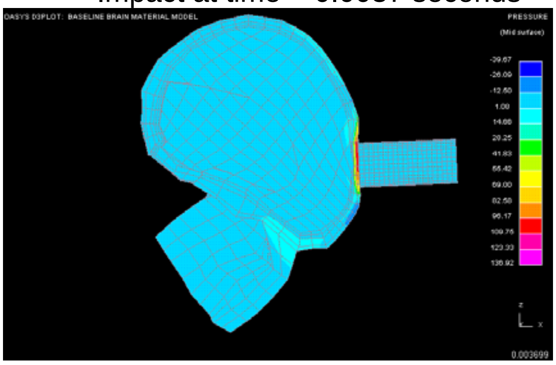
Connecting Veins: Skull, Brain along centre line, spring elements

VALIDATION

The head model is validated against published cadaver tests which have served to validate other FE Head models [8]. All FE head validations use the head only, treating body effects as negligible. The short impact duration (6 ms) makes this a reasonable assumption. The validation case is a steel cylinder projected at forehead of an instrumented cadaver. This duration is also similar to the impact spike experienced during the head to leg contact. The cylinder mass is 5.23 kG with a speed of 6.33 m/s

Head Model Pressure Response

Impact at time = 0.0037 seconds



The human head model was validated to the injury responses of acceleration measured at the head Center of Gravity, contact force measured at the skull bone impact region, and coup / contra coup pressure measured at brain elements in the region of impact (coup), and at the region directly opposite (contra-coup). Good correlations were obtained with published cadaver data for acceleration and contact force. The peak values are shown in the table below.

Validation vs Cadaver Results

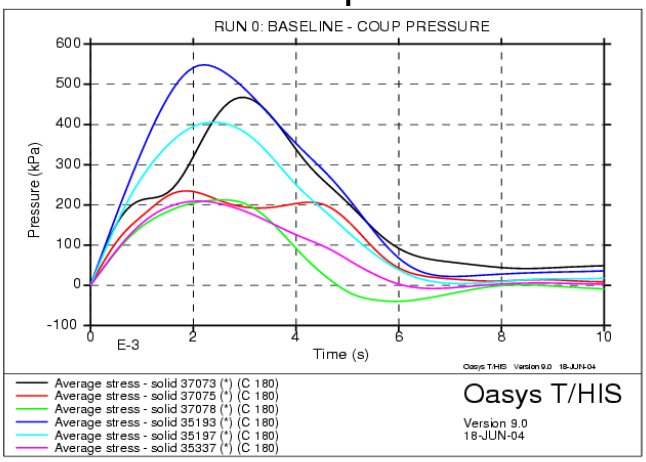
Peak Value	Cadaver	FE Model
Contact Force	8.1 kN	7.9 kN
Acceleration	210 G	200 G

COUP / CONTRA COUP RESPONSE

Good correlation with the brain pressure response of the cadaver test case was not achieved. A separate phase of the study was added to improve this response. Material properties, modelling parameters, and alternative test cases were investigated based on other published work. [9-14] Evaluation of six different options for brain tissue concluded that the original material/method was best, but improvements were made to the contact definitions. It was also found that because the pressure gradient changes rapidly, relative to the measurement location, reported values are subjective relative to the chosen element.

Pressure response is a valid measure of trauma severity as reported in several sources, but the measurement location and gradient must be taken into account. [8] A conservative approach was taken to assign the pressure values used for injury assessment. Rather than select the maximum pressure value, a group of six brain elements in the surrounding area were measured. The mid pressure value in this range was then selected as the value to compare with the threshold criteria. The validation case shown below has peaks that range from 200 to 550 kPa, thus the value of 375 kPa is used for assessment.

Coup Pressure Measured at 6 Elements in Impact Zone

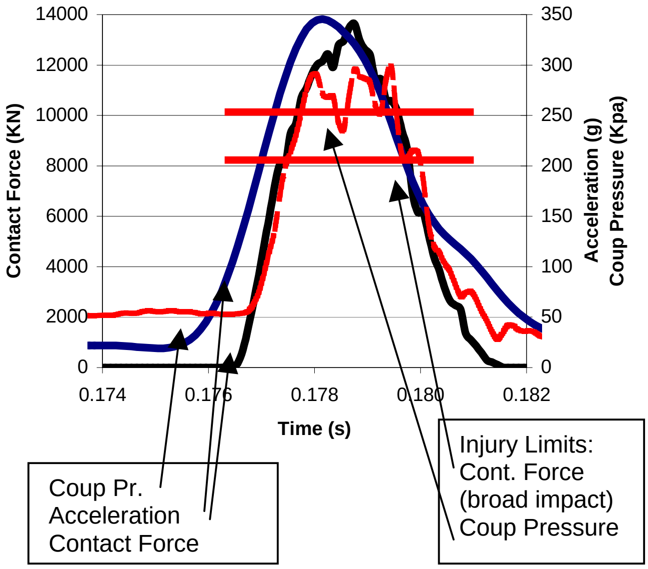


The contra coup response measures the pressure reaction at the back of the skull. This occurs after the brain rebounds from the front. The complex motions of this behaviour did not provide reliable correlation to the validation case. This response is not included in the injury assessment.

INJURY THRESHOLDS

The two most common measures for head impact injury are the contact force of the skull impacting the strike hazard and the pressure/stress response of the brain impacting the skull. The contact force threshold for skull fracture depends on the bone strength, which varies according to various factors such as age and formation. The adopted threshold value of 7650 kN is a middle range fracture level for broad surface impact. A three-inch radiused impactor lowers this number to 4581 kN. [15] Threshold criteria for serious brain injury has been cited in several sources, and is defined as the onset of brain contusions exhibited in post impact cadaver autopsies. Serious injury is defined to occur when the pressure exceeds 234 kPa. The figure below illustrates the baseline response versus the adopted threshold criteria.

FED0351 Injury Response



PARAMETRIC STUDY

A parametric study was conducted to evaluate injury potential as a function three factors. A simple, linear orthogonal, eight run fractional factorial design was chosen. The factors and levels are:

Factor A: IMPACT PULSE
High (+): 16G Triangle, 180 ms duration
Low (-): 12G Triangle, 120 ms duration

Factor B: BELT SLACK
High (+): 10% Slack
Low (-): 20% Slack

Factor C: SEAT RECLINE ANGLE
High (+): Upright seat back (0 deg.)
Low (-): Reclined seat back (10 deg.)

The fractional factorial experiment is structured to derive equations for the response as a function of the factors and their interactions. Factor levels can then be selected and the performance predicted. The test matrix defining the eight model simulations is given below, followed by the response equations for HIC, contact force, and coup pressure responses.

Fractional Factorial Matrix

	A: Impact Pulse	B: Belt Slack	C: Backrest Angle
1	12G 120 ms	20%	10 deg recl.
2			upright
3		10%	10 deg recl.
4			upright
5	16G 160 ms	20%	10 deg recl.
6			upright
7		10%	10 deg recl.
8			upright

Response Equations

$$Y = y_{avg} + (\Delta^{A}/_{2}*A) + (\Delta^{B}/_{2}*B) + (\Delta^{AB}/_{2}*AB) + (\Delta^{C}/_{2}*C) + (\Delta^{AC}/_{2}*AC) + (\Delta^{BC}/_{2}*BC) + (\Delta^{ABC}/_{2}*ABC)$$

HIC

$$Y = 1116 + (1650A) - (38B) - (86AB) - (209C) + (188AC) + (488BC) + (478ABC)$$

Contact Force

$$Y = 6869 + (4124A) - (986B) - (910B) - (2242C) + (504AC) + (1884BC) + (1808ABC)$$

Coup Pressure

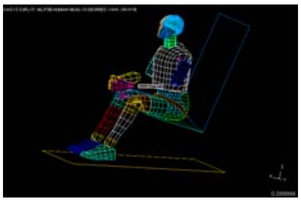
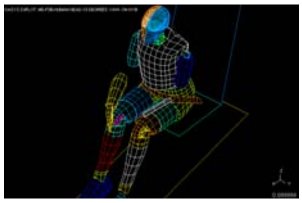
$$Y = 206 + (133A) - (29B) - (23AB) - (49C) + (24AC) + (41BC) + (35ABC)$$

The first value in each equation (y_{avg}) is the average of the eight runs for that response. In other words, y_{avg} is a prediction of the performance for a test conducted at the mid point of each factor range (14G, 15% belt slack, and 5 degree recline). HIC was on average just above the injury limit threshold with a value of 1116, while contact force and coup pressure were, on average, just below the threshold with values of 6869 kN and 206 kPa respectively.

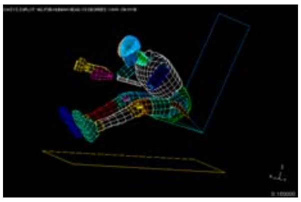
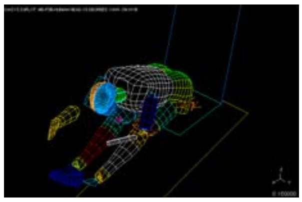
The coefficient in front of each variable represents half of the total effect or influence on the response. The sign indicates the direction relative to the defined low and high levels. Factor A's influence from low (12G) to high (16G) increased (+) all injury measures. Factors B and C low to high, (tightening the belt and putting the seat back upright), reduced (-) injury. The influence of B and C were much lower than impact severity, but the interactions were very significant. A slack belt and reclined seat together, especially at higher impact levels, causes large increases of injury potential.

Sample Impact Sequences

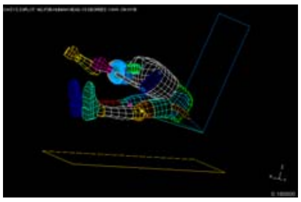
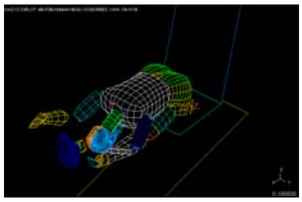
12 G, Loose, Upright **16 G, Tight, Upright**

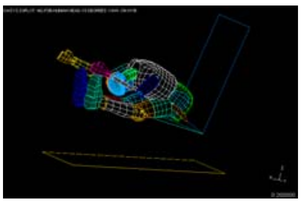
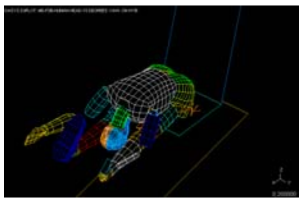
12 G 100 ms
16 G 100 ms

12 G 150 ms
16 G 150 ms

12 G 180 ms
16 G 180 ms

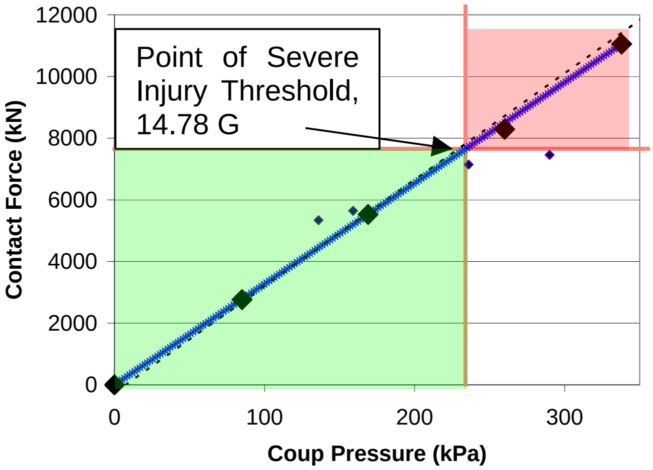
12 G 200 ms
16 G 200 ms

ONSET OF SEVERE INJURY

A primary objective of the research is to understand if and when aircraft impacts produce sufficient occupant articulation to cause injurious head impact to the lower legs. The following two graphs address this objective. Contact force vs Coup pressure are shown relative to the impact severity. The first graph has B and C fixed at Tight and Upright. The second graph adds the same response for B and C fixed at Slack and Reclined.

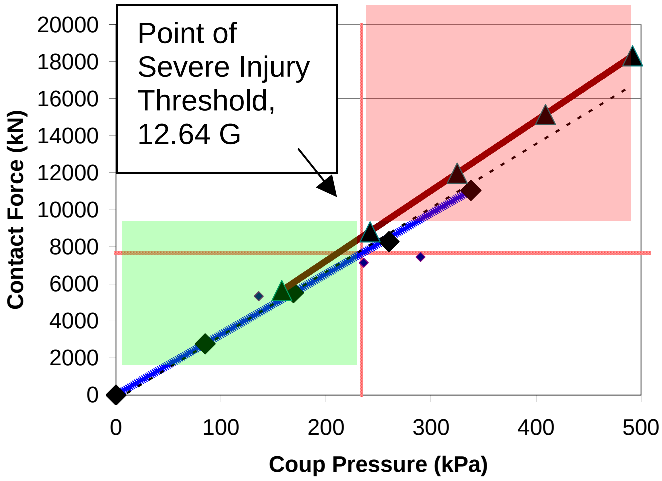
The severe injury thresholds were crossed between 12.6 to 14.8 G, depending on the setting of the other factors. It is also clearly shown that an occupant position more susceptible to "whip" forward generates higher head velocities and is much more susceptible to injurious head strike.

Injury Across Impact Range: Tight, Upright



The large black diamonds show the experiment points for the factors B and C set to 10% Slack and Upright, while Factor A varies from 12 to 16 G's. The blue line is the predicted response between the test points. The small diamonds are the test points at other factor levels, and dashed line is the trend line for the correlation.

Injury Across Impact Range, with both Tight/Upright and Loose/Reclined



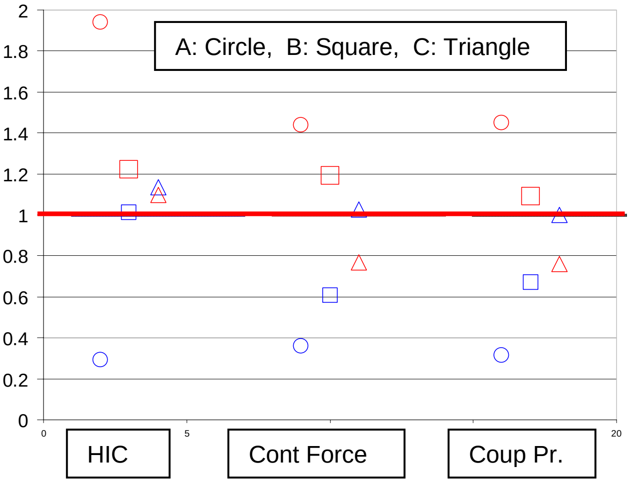
EFFICACY OF HIC

The other primary objective is to understand how well current regulatory test methods and tools (ATD and HIC) can provide an estimate of this injury potential. The responses were normalized at a common value of 1.0 in order to directly compare each estimate of injury severity relative to one another.

To achieve the nominal values, the average low and high response for each factor, A, B, C, were divided by the injury threshold. For example, HIC for factor A ranged from a low of 291 to a high of 1941. Normalizing these around the threshold of 1000 yields a low $(^{291}/_{1000})$=0.29 and high $(^{1941}/_{1000})$ = 1.94. HIC at the low level is only 30% of the threshold value, while at the high level is nearly 200% of the threshold.

The range for each factor indicates the magnitude of the influence. Factor A had a large affect on HIC (about 30% to 200%). Factors B and C indicated the injury trend, but had a relatively small affect on HIC for these factor settings. The contact force and coup pressure responses provided a more accurate measurable affect for factors B and C. This example illustrates the exponential nature of HIC. The response is very sensitive to changes in the acceleration versus time impact spike.

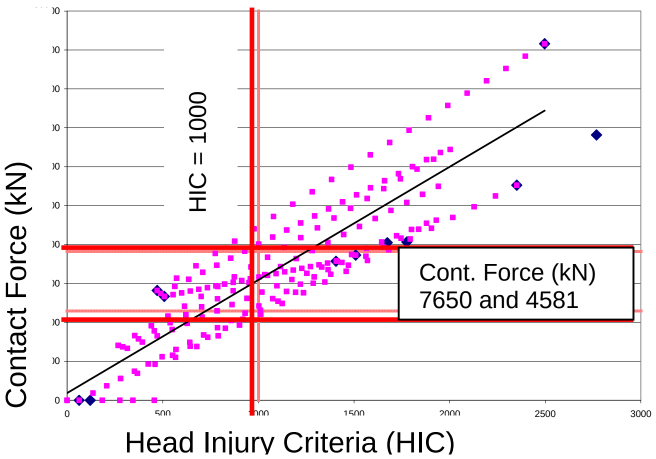
Normalized Factor Effects



HIC as a predictor of injury correlates well with that of contact force or coup pressure, although at smaller impact pulses the results are more variable. Of the eight parametric runs, all three responses yielded the same pass/fail result. A correlation was made between HIC and contact force, and large population of predicted results were generated. The pass / fail ratio was then compared to give a better estimate than only eight runs.

The simulation range and intervals were:
A Impact: 12 to 16G by increments of 0.1G
B Belt Slack: 20% to 10% by increments of 5%
C Backrest Angle: 10 to 0 deg by incr. of 5 deg

Contact Force vs HIC, Predicted Population



The results indicated that HIC reaches the threshold value of 1000 units at about 13G, where contact force, using a broad impactor crossed the threshold of 7650 kN at about 15 G. The correlation of pass/fail ratio for contact force versus HIC is 88% (12% failed HIC that would have passed contact force). Note that if the skull fracture threshold for a 3 inch radius impactor (4600 kN) is used, the correlation is 96% (13.6 G). The leg is curved suggesting that the radius value may be more appropriate, but the research did not look in detail at shape related factors.

RESULTS

Dynamic testing and Finite Element tools were successfully applied to evaluate the injury potential for long pitch seat configurations. The evaluation method addressed the limits of ATD biofidelity, providing insight to both the direct human response and the efficacy of the ATD response.

The baseline test and analysis clearly indicates flailing head to leg contact causing severe head impact injury. The parametric study indicates the importance of seat and restraint design factors on the injury risk. Characterization with respect to impact severity indicates that the severe injury threshold is reached for impacts significantly below the 16G pulse.

CONCLUSIONS

FE methods were applied to use simulated human response to measure the potential for head to leg impact injury during a survivable impact. The results indicate severe injury potential within the scope of FAR/JAR 25.562 and 25.785. Further, the parametric study indicates that seat and restraint design factors have a strong influence on the injury potential. Removal of the strike hazard alone does not eliminate potential for head impact injury.

The second objective was to assess the capability of the current regulatory test devices. HIC was found to be a meaningful measure of head injury potential for the head to lower leg impacts. The parametric study indicated reasonable correlation between HIC and the simulated direct injury responses. The affect of individual seat factors were also reflected in the HIC response. The head acceleration response and HIC are useful in optimizing seat designs for the injury mechanism studied. The occurrence rate of injurious impacts for long pitch seats is unknown, and needs to be characterized relative to the seat and restraint design factors.

RECOMMENDATIONS

It is recommended that HIC be recorded and reported for all dynamic compliance tests. This will provide regulatory authorities with the data needed to assess long pitch seats in general, and a means to evaluate relative safety of design trends.

Industry should consider HIC data generated from "no strike" seat configurations during the design phase to avoid this injury mechanism. Regulators should review data for these seat configurations to consider if the "no strike" means of compliance is consistent with the intent of safety regulations 25.785 and 25.562.

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