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Assessment of the Accuracy of Certain Reduced Order Models used in the Prediction of Occupant Injury during Under-Body Blast Events

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14. ABSTRACT

It is of considerable interest to developers of military vehicles, in early phases of the concept design process as well as in Analysis of Alternatives (AoA) phase, to quickly predict occupant injury risk due to under-body blast loading. The most common occupant injuries in these extremely short duration events arise out of the very high vertical acceleration of vehicle due to its close proximity to hot high pressure gases from the blast. In a prior study [16], an extensive parametric study was conducted in a systematic manner so as to create look-up tables or automated software tools that decision-makers can use to quickly estimate the different injury responses for both stroking and non-stroking seat systems in terms of a suitable blast load parameter. The primary objective of this paper is to quantitatively evaluate the accuracy of using such a tool in lieu of building a detailed model for simulation and occupant injury assessment.

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DeltaV, accuracy, Effective G, Blast, ROM, reduced order models, MADYMO, occupant, injury, pulse, loading, descriptor, calculator

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ABSTRACT

It is of considerable interest to developers of military vehicles, in early phases of the concept design process as well as in Analysis of Alternatives (AoA) phase, to quickly predict occupant injury risk due to under-body blast loading. The most common occupant injuries in these extremely short duration events arise out of the very high vertical acceleration of vehicle due to its close proximity to hot high pressure gases from the blast. In a prior study [16], an extensive parametric study was conducted in a systematic manner so as to create look-up tables or automated software tools that decision-makers can use to quickly estimate the different injury responses for both stroking and non-stroking seat systems in terms of a suitable blast load parameter. The primary objective of this paper is to quantitatively evaluate the accuracy of using such a tool in lieu of building a detailed model for simulation and occupant injury assessment.

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INTRODUCTION

Underbody blasts have become one of the most widespread reasons for warfighter casualties in recent wars [1, 2, 3]. Spinal injuries to occupants have particularly increased in theater from these roadside blast incidents, followed by tibia and lower leg injuries. To support the design and development of military ground vehicles, mine blast underbody hull kits and mine blast seats, a suite of underbody modeling methods were quickly developed [4, 5, 6, 7, 8, 9, 10, 11]. These modeling and simulation (M&S) methodologies are being continuously enhanced with ever-increasing capabilities to predict vehicle structural and occupant injury responses, including fast running models for the same [16, 17].

It is of considerable interest to developers of military vehicles, in early phases of the concept design process, to quickly predict occupant injury risk due to under-body blast loading [12, 16]. The most common occupant injuries in these extremely short duration events arise out of the very high vertical acceleration of vehicle due to its close proximity to hot high pressure gases from the blast. A typical blast vertical acceleration history which is predominantly triangular shaped in nature is shown in Fig 1, and is often measured at a rigid location on the military vehicle to serve as a representative measure or a "signature" of the blast severity, and is often referred to as the "blast pulse". This can be thought of as being analogous to the crash pulse used in the automotive industry to serve as the loading that is experienced in the fore-aft direction by the structural components as well as the occupants in the vehicle interior. For example, the seats experience the pulse as a load at their structural attachment points. Other than the obvious difference in sign and direction (frontal crash results in vehicle deceleration in the fore-aft direction, while underbody blasts result in vehicle acceleration in the vertical direction), there are two other major differences between the blast pulse, and its frontal safety crash counterpart, namely:

- the peak acceleration - the blast pulse tends to be 5-10 times *larger* in magnitude, and
- the duration of the pulse - the blast pulse tends to be 3-5 times *shorter* than its crash counterpart.

As a common feature, both pulses serve as design criteria for development of seats, restraints and other safety features, and are even measured in a similar manner. The frontal crash pulse is usually the average fore-aft deceleration measured at one or more accelerometers at the stiff B-pillar/Rocker joint areas, while the blast pulse is usually the average vertical acceleration measured at one or more accelerometers at the stiff pillar/roof joint areas.

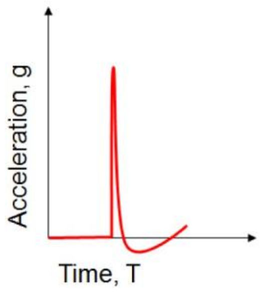

Fig 1. A typical blast pulse

There has been a continual quest in the blast community of practice to define one or more loading parameters from the "blast pulse" that would by themselves, or in combination thereof, serve as indicators of blast severity and therefore occupant injuries, similar to the crash pulse scenario. For example, in automotive frontal crashes, the peak value in the crash deceleration pulse is one quantity that directly correlates to occupant injuries, everything else being constant. In the past, several similar loading parameters have been proposed for blast pulses. Of these, the design community has mostly used change in velocity Δ*v*, or to a much lesser extent, peak acceleration G_{peak}, to determine the severity of, and classify any given blast pulse.

BLAST LOAD INDICATORS/DESCRIPTORS

In the past, several blast loading parameters have been proposed, alone or in combination, to serve as indicators or predictors of occupant injuries[16]. Some examples of these are:

1. magnitude of the peak acceleration, G_{peak} in g's,
2. time duration of pulse, *T* in milliseconds (ms),
3. rate of onset of acceleration, $\dot{G}$ in g/ms,
4. change in velocity, Δ*v* in m/s
5. effective-g[13] (slope of the velocity profile) G_{eff} in g's,
6. specific power[14], *SP* (G_{peak}*Δ*v*) in g-m/s, or m^2/s^3.

Of these, the Survivability design community has largely used change in velocity Δ*v*, or to a lesser extent, peak acceleration G_{peak}, to describe the severity of, and classify any given blast pulse.

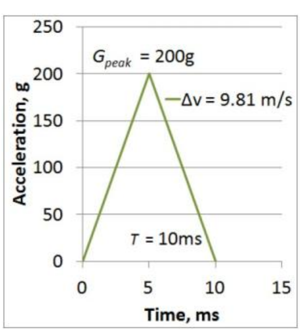

Fig 2. An example triangular blast pulse (Acceleration profile)

Fig. 2 shows an example of a blast pulse with a peak acceleration of 200g and 10ms duration. The corresponding velocity profile is shown Fig. 3, along with illustrations of how Δ*v*, G_{avg} and G_{eff} are calculated for the same.

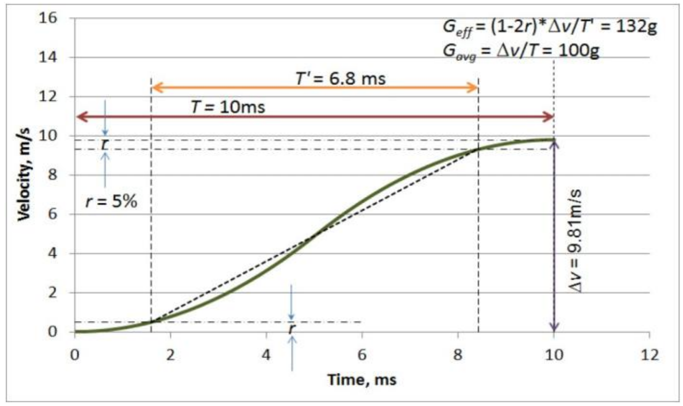

Fig 3. An example triangular blast pulse (velocity profile) Definition of effective-g [15] is also shown

BACKGROUND OF PRIOR STUDY [16]

Blast Simulation Model

Two setups of the dynamic simulation model of a vertical blast loading simulator are shown in Fig. 4 and 5. In the first setup, a 50th percentile Humanetics' Finite Element (FE)-based Hybrid-III Anthropomorphic Test Dummy (ATD) model (v7.16) in LSDYNA® format is seated on a rigid finite element seat with a five point seat belt as shown in Fig. 4. This seat is rigidly placed on a vertically sliding platform (not shown in figure) where the blast pulse was input as base excitation. A typical run time for a FE-based simulation lasting 100ms duration is about two hours using 16 processors on an Intel x86-64 based Linux server.

Fig 4. LSDYNA dynamic simulation model including Humanetics v7.1.6 Hybrid III 50th percentile ATD model

In the second setup, a vertical blast loading sled is constructed [13] using finite elements of assigned rigid material. A MADYMO® rigid multibody (RMB)-based model of the 50th percentile Hybrid-III ATD (Q version) seated on a RMB seat, with a three point seat belt, is attached to a vertically sliding platform where the blast pulse was input as base excitation. A typical run time for a RMB-based simulation lasting 100ms duration is about 20 minutes using a single processor on an Intel x86-64 based Linux server.

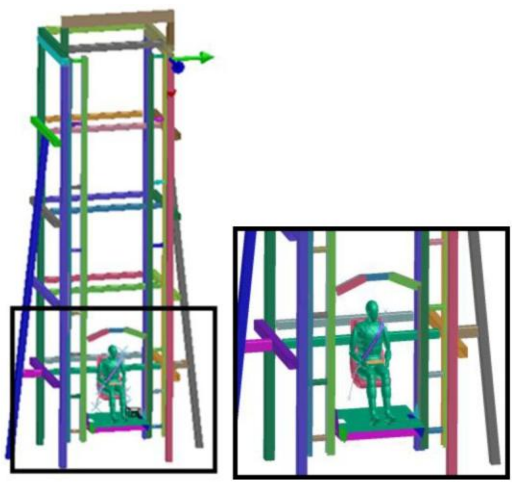

Fig 5. MADYMO dynamic simulation model including Q-version of Hybrid III 50th percentile ATD

Fig 6. Photograph of a typical drop tower test fixture

A vertically sliding platform also known as a drop tower test fixture is shown in Fig 6. ATDs can be seated and the platform, including the seat and ATD, dropped from a suitable height to achieve any desired Δ*v* over the duration *T*. The target pulse can be achieved by controlling the energy absorption characteristics of the floor on which the platform is dropped upon. Alternatively, a base excitation can be provided to the sliding platform in the upward vertical direction to achieve any given pulse if the fixture is so equipped (Also known as vertical sled). These two scenarios are completely equivalent in the occupant response behavior for the same given pulse.

Occupant injuries recorded from both these approaches were compared against those measured during physical tests. The resulting comparison showed no significant differences between the two approaches. Therefore the ensuing parametric modeling and simulation (M&S) study, involving a large number of simulations, was conducted on the latter multibody-based MADYMO® model since the run times for the latter are significantly lower than that of the FE-based model.

Parametric Study

A parametric study was conducted by varying the peak acceleration from 10g-1200g, and duration of the pulse from 2.5ms to 60ms (a total of thirteen distinct duration levels) such that Δ*v* is varied to a maximum value of about 15 m/s. The blast pulses with the minimum and maximum amplitudes from these thirteen duration levels are shown in Fig 7. In addition to a rigid seat without an energy-absorbing (EA) mine blast feature, two other generic EA blast mine seats of different ratings (EA1, EA2) were also used in the study (Fig 8). As may be observed from the figure, both EA seats have the same amount of stroke, but EA2 is softer in that it strokes at a lower limiting force level than EA1.

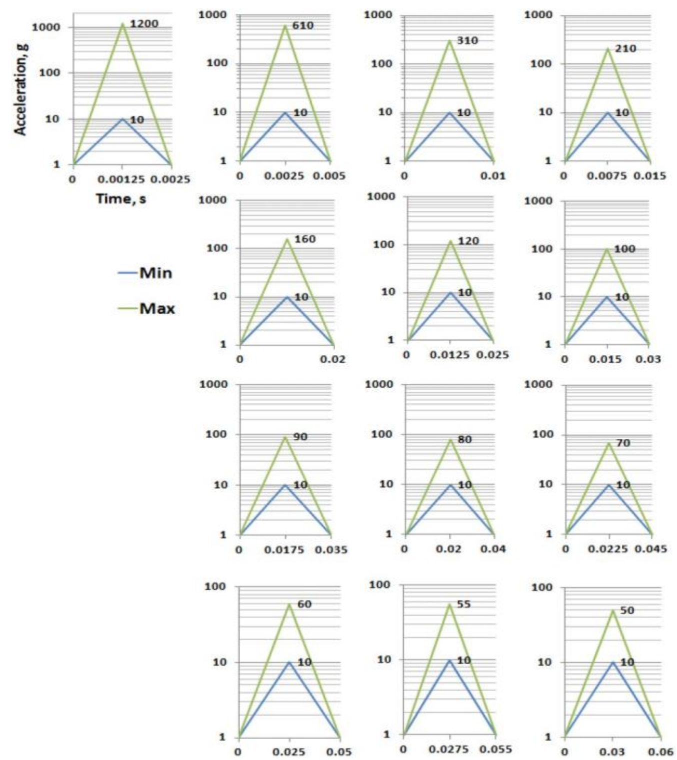

Fig 7. Blast pulses with minimum and maximum amplitude for the thirteen duration levels considered in this study.

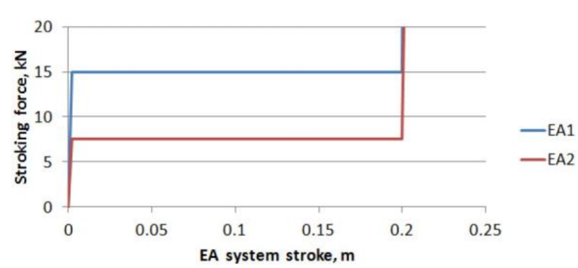

Fig 8. Two generic seat EA systems characteristics considered. Baseline EA (EA1) and Softer EA (EA2)

A total of ten different upper body injuries were recorded and monitored, namely:

(1,2,3) Peak, 2ms clip and HIC_{15} criterion of the head resultant acceleration,

(4) N_{ij} criterion for the neck injury,

(5,6) 3ms and 7ms clips of chest resultant acceleration,

(7) 7ms clip of pelvic vertical acceleration,

(8,9) Peak and 30ms clips of lumbar spine compression, and

(10) Pelvic vertical Dynamic Response Index (DRI)[12].

The sample size for each of the three seating variants consisted of 230 MADYMO® simulations, for a total of 690 data points. For each simulation, the ten occupant injuries are plotted and assessed using the post processing tool Hypergraph® as shown in Fig 9. In the next section, each of these ten injury responses are plotted against three different blast loading parameters, grouping them by the pulse duration *T*, to look for trends (Fig. A2).

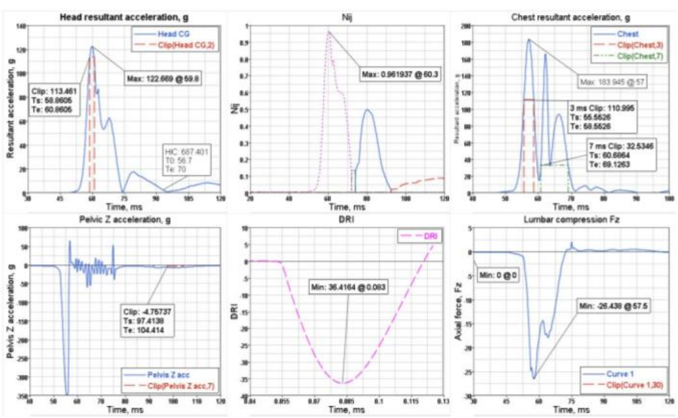

Fig 9. Recording occupant injury metrics

Occupant Injury Responses

Figs A1 shows ten occupant injuries, plotted against the change in velocity, Δ*v*, for the three seat types. The first column of plots shows the ten occupant injuries for the rigid seat, plotted against Δ*v*. The second and third columns of plots are the same injuries for the seats EA1 and EA2, respectively.

In each of the 30 plots in Fig A1-a & b, several curves can be seen. Each of these curves corresponds to a constant value of *T*, the duration of the blast pulse. In essence, they are the same as iso-*T* curves.

Findings

A detailed regression study was conducted to see if there was a single blast loading parameter which could be used to adequately characterize occupant injuries. Three blast loading descriptors proposed in the literature [12, 13, 14], viz., Δ*v*, G_{eff}, and Specific Power (*SP*) were evaluated for this purpose. Figure A2 shows occupant injuries HIC_{15ms} (Fig. A2-a) and DRI (Fig. A2-b) plotted against these three blast loading parameters for the three different seat designs considered in this study. It may be observed from Fig A2 that the injury responses are much more bunched together when plotted against Δ*v*, as opposed to the other two blast loading parameters, thereby indicating a higher potential for Δ*v* to be the single indicator for occupant injury characterization. This trend to bunch together is even more pronounced in smaller intervals of *T*, that is, in 10 ms groupings such as 0-10, 10-20, 20-30, etc. It must also be mentioned here that the iso-*T* curves, for all three load descriptors, tend to bunch together much better for larger values of pulse duration *T* (>30 ms), but these durations are not typical of underbody blasts but more representative of standard automotive crash events. These observations were also validated by the rigorous regression study, the key finding being that Δ*v* is the best single indicator for estimating injury criteria, for typical blast pulse duration ranges (0-20 ms), independent of seat type.

REDUCED ORDER MODELS (ROM) / INJURY LOOK-UP TABLE

Three different Reduced Order Modeling approaches to construct injury look-up tables with increasing levels of fidelity and thereby accuracy are described below.

Approach #1 (Single Parameter)

Using the linear/quadratic regression equations resulting from this parametric study, occupant injuries for any triangular-shaped pulse can be easily computed. As an example, Figure 10 shows one such occupant injury, viz., Head acceleration 2ms clip vs. Δ*v*.

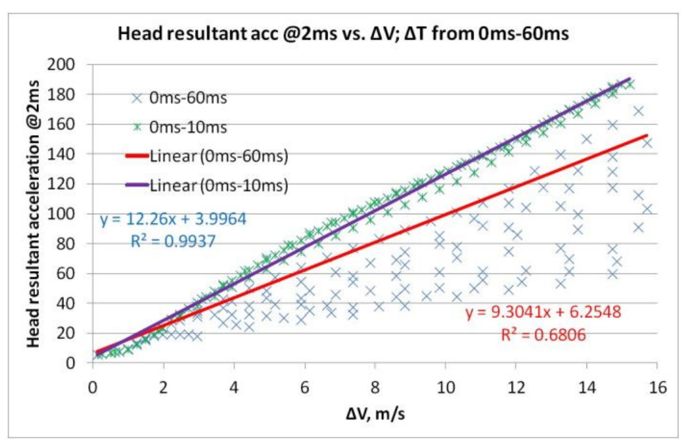

Fig 10. Head resultant acceleration vs. Δv for two duration ranges

Head acceleration (2ms clip) now can simply be calculated by using the linear regression equations from the curve above. For the entire range of *T*, i.e., 0 ≤ *T* ≤ 60ms,

$$Head\ Acc\ (2ms\ Clip) = 9.3041 * \Delta v + 6.2548 \tag{1}$$

Similar regression equations were constructed for other injuries and seat designs. One limitation of this approach is that if a single simple relationship for the entire range of pulses and durations of interest considered in this study were to be constructed, it would result in significant error. To minimize this error, for those pulses in the *typical blast loading range* i.e., $0 \leq T \leq 10$ms, another set of regression equations can be derived from the trend line above as,

$$Head\ Acc\ (2ms\ clip) = 12.26 * \ \Delta v + 3.9964 \tag{2}$$

The comparison is demonstrated in the Results section.

Approach #2: (Iso-T Regression Lines)

This is an extension to the previous approach where regression analyses for the entire range of pulse durations, by suitably grouping them in to a finer (5ms in this study) intervals, is performed. This approach results in a set of regression lines for each group of pulse durations and therefore results in improved accuracy. For example, Head acceleration (2ms clip) regression equations are:

$$Head\ Acc.(2ms\ clip) = 12.368 * \ \Delta v + 4.7557\ (0 \leq T \ \leq 5ms) \tag{3}$$

$$Head\ Acc.(2ms\ clip) = 12.03 * \ \Delta v + 1.3809\ (5 \ \leq T \ \leq 10ms) \tag{4}$$

▪

▪

$$Head\ Acc\ (2ms\ clip) = 3.469 * \ \Delta v + 8.2534\ (55 \ \leq T \ \leq 60ms) \tag{5}$$

Similarly linear regression equations can be derived for other occupant injuries. As with approach #1, the regression analyses were performed using Microsoft Excel®. Based on the linear/quadratic regression equations resulting from this approach, a macro-enabled tool has been built using Microsoft Excel® to predict occupant injuries for any triangular-shaped pulse (within a certain range of $0 \leq \Delta v \leq 15$ m/s). This look-up table is shown in Table A1 in Appendix. Users can simply select and input two key blast descriptors such as peak acceleration and time duration (shown as yellow cells in Table A1-2) and the injury table gets automatically updated (Table A1-3). User inputs can be provided by directly entering in the yellow cells, or by dragging the sliders shown in Table A1-2 to desired values. The injury values, as determined by the various appropriate best-fit equations depending on *T*, are computed and compared against the reference values and color-coded as low, moderate and high risk. The users have the option of setting the reference values per their specific program needs/targets. Results are shown for all three seat types as well. In addition, users can also choose the factor *r* (default value of 0.05) to determine effective-g, G_{eff}, for the chosen pulse which also gets updated and displayed graphically as shown in Table A1-1.

Approach #3: (Response Surface Metamodel)

Using the injury data obtained from the parametric M&S study, a surface-based metamodel was constructed using LSOPT®. Three-dimensional injury response surfaces were obtained for the ten injury parameters considered in this study from the LSOPT® simulations. Each of the injury surfaces was created as a function of the blast loading descriptor and the loading duration *T*. For example, Fig 11a shows contours of the peak lumbar compression response surface as a function of Δv and *T*, while Fig 11b shows contours of the same injury as a function of G_{eff} and *T*. Similar surfaces were also constructed for other injury parameters and they can be used as injury lookup tables with the blast loading descriptor and *T* as independent variables. The peak head acceleration injury response surface is shown in three dimensions plotted against Δv and *T* (Fig 12a) and against G_{eff} and *T* (Fig 12b). One important observation that may be made is that while both surfaces are mathematically equivalent, the uniformity of the surface against Δv makes it a more suitable candidate for reduced errors during the numerical interpolations required for injury predictions using the response surface.

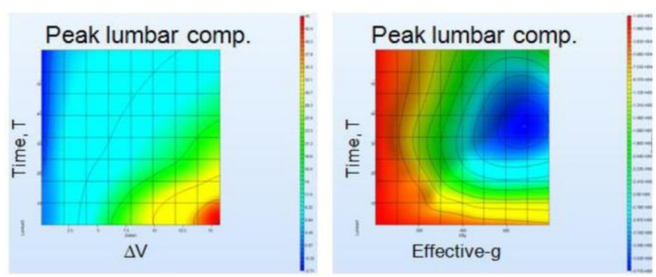

Fig 11 a & b. An example injury (Peak lumbar compression) look-up chart; ΔV and Effective-g as blast load descriptors along with duration of blast pulse

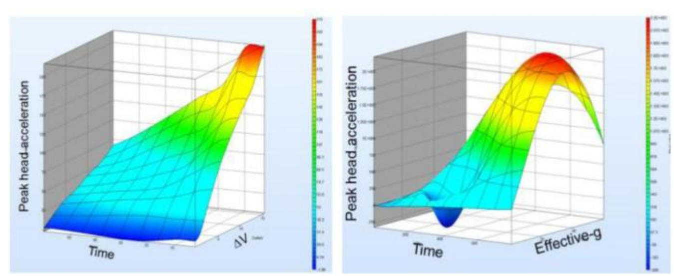

Fig 12 a & b. An example injury (Peak head acceleration) response surface in 3D; ΔV and Effective-g as blast load descriptors along with duration of blast pulse

RESULTS AND DISCUSSION

In this paper, as mentioned earlier, our objective is to demonstrate the accuracy and efficiency of using injury look-up tables in the occupant injury assessment without performing a direct blast simulation. Therefore, five arbitrary pulses as shown in Figure 13 and Table 1 were selected, *which were not in the original seed simulations*, of the parametric M&S study, to determine injuries as predicted by the three different approaches for comparison against corresponding results from the direct MADYMO® simulations.

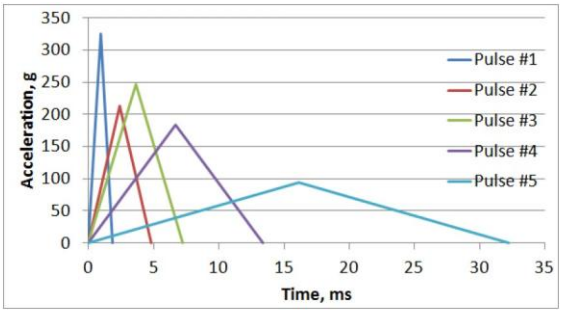

Fig 13. Various pulse shapes arbitrarily chosen to verify the validity of the predictor tool

Table 1. Arbitrarily chosen pulses to verify the validity of the three predictor approaches

Pulse #	Peak, Dec., g	Duration, ms	ΔV, m/s	Eff-g	Sp. Power
1	324.43	1.85	2.94	213.51	955.11
2	212.40	4.80	5.00	139.78	1062.15
3	246.90	7.23	8.76	162.48	2161.82
4	183.63	13.40	12.07	120.85	2216.31
5	94.32	32.23	14.91	62.07	1406.39

Approach 1

Table A2-a in Appendix shows occupant injuries predicted using the general regression equation (Eqn. 1) and actual MADYMO® simulations. It can be seen from the data that the difference is significant (Table 2a) as noted earlier.

Table 2a. Percentage error between predicted values from using Equation 1 and similar for other injuries (Approach #1) and actual simulation

Occupant Injury	Pulse #1	Pulse #2	Pulse #3	Pulse #4	Pulse #5
Head resultant acceleration 2ms-clip, g	-18%	-25%	-23%	-7%	27%
Peak Head resultant acceleration, g	-17%	-26%	-24%	-10%	26%
HIC 15	-74%	-14%	3%	15%	144%
Nij	-11%	-17%	-14%	-8%	19%
Chest resultant acceleration, 3ms clip, g	0%	-14%	-10%	10%	51%
DRI	-1%	-2%	-3%	-4%	2%
Peak Lumbar Compression, kN	-11%	-24%	-27%	-12%	50%

Table A2-b (Appendix) shows similar results from using Equation 2. For those pulses which are in the blast loading range (Pulse #1-3), the errors in predicted occupant injuries show significant reduction (Table 2b). This further validates our findings [16] that Δ*v* is the single best blast loading parameter which has a direct relationship with occupant injury. The maximum error was within a range of ±15% except for those cases when the injury magnitude is very low, e.g., HIC_{15ms} for Pulse #1. However, using this same equation to predict occupant injuries for those pulses which are outside the blast loading range (Pulse #4-5) the error is still significant (Shaded cells of Table 2b), as may be expected.

Table 2b. Percentage error between predicted values from using Equation 2 and similar for other injuries (Approach #1) and actual simulation

Occupant Injury	Pulse #1	Pulse #2	Pulse #3	Pulse #4	Pulse #5
Head resultant acceleration 2ms-clip, g	-3%	-7%	-2%	19%	63%
Peak Head resultant acceleration, g	-1%	-8%	-3%	16%	64%
HIC 15	-74%	-14%	3%	15%	144%
Nij	-5%	-5%	3%	11%	45%
Chest resultant acceleration, 3ms clip, g	1%	-4%	1%	18%	52%
DRI	0%	1%	1%	0%	7%
Peak Lumbar Compression, kN	3%	-5%	-3%	19%	105%

Approach 2

Table A2c in Appendix shows typical injuries predicted from the Excel® tool (Approach #2) as well as from the direct simulations. The maximum error was within a range of ±10% (Table 3) except for those cases when the injury magnitude is very low or very high, e.g., HIC_{15ms} for Pulse #1 and #5 respectively. It may be observed that the results from this approach are much more accurate than those shown in Table 2. This is to be expected since in Approach 1, one single equation was used over *all* the data points, whereas in Approach 2, a set of regression lines for each sub-group of pulse durations is utilized in the look-up tables.

Table 3. Percentage error between predicted values from the Excel tool (Approach #2) and actual simulation

Occupant Injury	Pulse #1	Pulse #2	Pulse #3	Pulse #4	Pulse #5
Head resultant acceleration 2ms-clip, g	1%	-6%	0%	-1%	0%
Peak Head resultant acceleration, g	4%	-7%	-2%	-5%	0%
HIC 15	-74%	-10%	5%	-6%	11%
Nij	0%	-3%	3%	-2%	-2%
Chest resultant acceleration, 3ms clip, g	7%	-2%	3%	-5%	3%
DRI	1%	0%	0%	-1%	2%
Peak Lumbar Compression, kN	7%	-3%	-1%	-7%	4%

Approach 3

Table A2d shows similar comparison data between injury values measured from the response surface (Approach #3) vs. those from actual simulation. Since the response surface is smooth and continuous throughout the region of our interest, the errors (Table 4) from the predicted values are further lowered from Approach 2. The Excel® tools used in approach #2 can be suitably modified to utilize this response surface instead of the discrete regression equations for increased accuracy in injury predictions.

Table 4. Percentage error between injury values from the response surface (Approach #3) and actual simulation

Occupant Injury	Pulse #1	Pulse #2	Pulse #3	Pulse #4	Pulse #5
Head resultant acceleration 2ms-clip, g	0%	0%	-3%	0%	0%
Peak Head resultant acceleration, g	1%	0%	-1%	0%	0%
HIC 15	3%	0%	-1%	1%	0%
Nij	0%	0%	0%	0%	0%
Chest resultant acceleration, 3ms clip, g	0%	0%	-1%	0%	0%
DRI	0%	0%	0%	0%	0%
Peak Lumbar Compression, kN	-1%	0%	-2%	-1%	1%

These results clearly show that the methodology has enormous advantages as a reduced order modeling tool, taking merely a few seconds to predict the injuries accurately as opposed to over 20 minutes of computation time, followed by hours of post-processing, plotting, tabulation, interpretation, etc. by an expert user/analyst. Also, casual users of the predictor tool avoid the costs of owning, maintaining and learning expensive M&S software tools.

CONCLUSIONS

The following broad conclusions may be made from the analysis, results and discussions of the preceding sections:

1. Three different reduced order modeling approaches of increasing fidelity and accuracy were constructed and evaluated for their ability to predict occupant injury behavior.
2. An easy-to-use, rapid injury estimator tool was constructed in Microsoft Excel® as a function of input load descriptors, using the occupant injury regression trends obtained from a detailed parametric study.
3. This tool takes mere seconds to arrive at accurate injury predictions when compared to the direct method which takes a minimum 20 minutes with additional time required for post-processing, plotting, and tabulation, etc. by an expert user. Also this tool does not require the expensive software, training and hardware associated with the direct method.
4. This tool will enable decision makers to quickly arrive at informed decisions during early concept design stages, Analysis of Alternatives (AoA) studies, etc.
5. It is noteworthy that these results are only representative of the underlying power of the technology. By extending this methodology to one or more seats with the EA as one of the design variables, family of better validated ATDs of different sizes, new and improved injury criteria from the bio-medical research the tool can be made extremely useful in ground vehicle acquisition.
6. The methodology used in this project is being planned for extended use elsewhere in the Army for data from physical drop tower/vertical sled tests, as well as from Live-Fire blast tests to develop similar empirically-based tools for use by designers, program managers, evaluators, etc.
7. This methodology can also be used elsewhere in the automotive industry to develop reduced order models using occupant injury tables to assist conceptual studies during early phase of product development.

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DISCLAIMER

DEFINITIONS/ABBREVIATIONS

AoA - Analysis of Alternatives

ATD - Anthropomorphic Test Device

ATEC - Army Test and Evaluation Command.

ARL - Army Research Laboratory

COTS - Commercially-Off-The-Shelf

DoA - Department of the Army

DoD - Department of Defense

DRI - Dynamic Response Index

DTIC - Defense Technical Information Center, https://www.dtic.mil/

Δv - Delta-V

EA - Energy Absorbing

FE/FEA - Finite Element/Finite Element Analysis

G_{Avg} - Average acceleration of pulse

G_{Eff} - Effective-g

G_{peak} - Peak Acceleration value of pulse

HIC - Head Injury Criterion

IED - Improvised Explosive Device

LSDYNA® - structural dynamics software from Lawrence Livermore Software Corporation, CA

LSOPT® - COTS optimization software from Lawrence Livermore Software Corporation, CA

MADYMO® - MAthematical DYnamic Models, COTS multibody dynamics software from TASS, Netherlands

ms/msec - Milliseconds

M&S - Modeling and Simulation **MSU** - Mississippi State University **MB/RMB** - Multi-body/ Rigid Multi-body **N_{ij}** - Neck injury criterion

r_c - Correlation coefficient

R&D - Research & Development

RO/ROM - Reduced Order / Reduced Order Model

SimBRS - Simulation Based Reliability and Safety

SLAD - Survivability and Lethality Analysis Directorate

SP - Specific Power

TACOM - Tank Command

TARDEC - Tank Automotive Research, Development and Engineering Center

TASS - TNOAutomotive Safety Solutions division

UBM - Underbody Blast Modeling/Methodology

APPENDIX

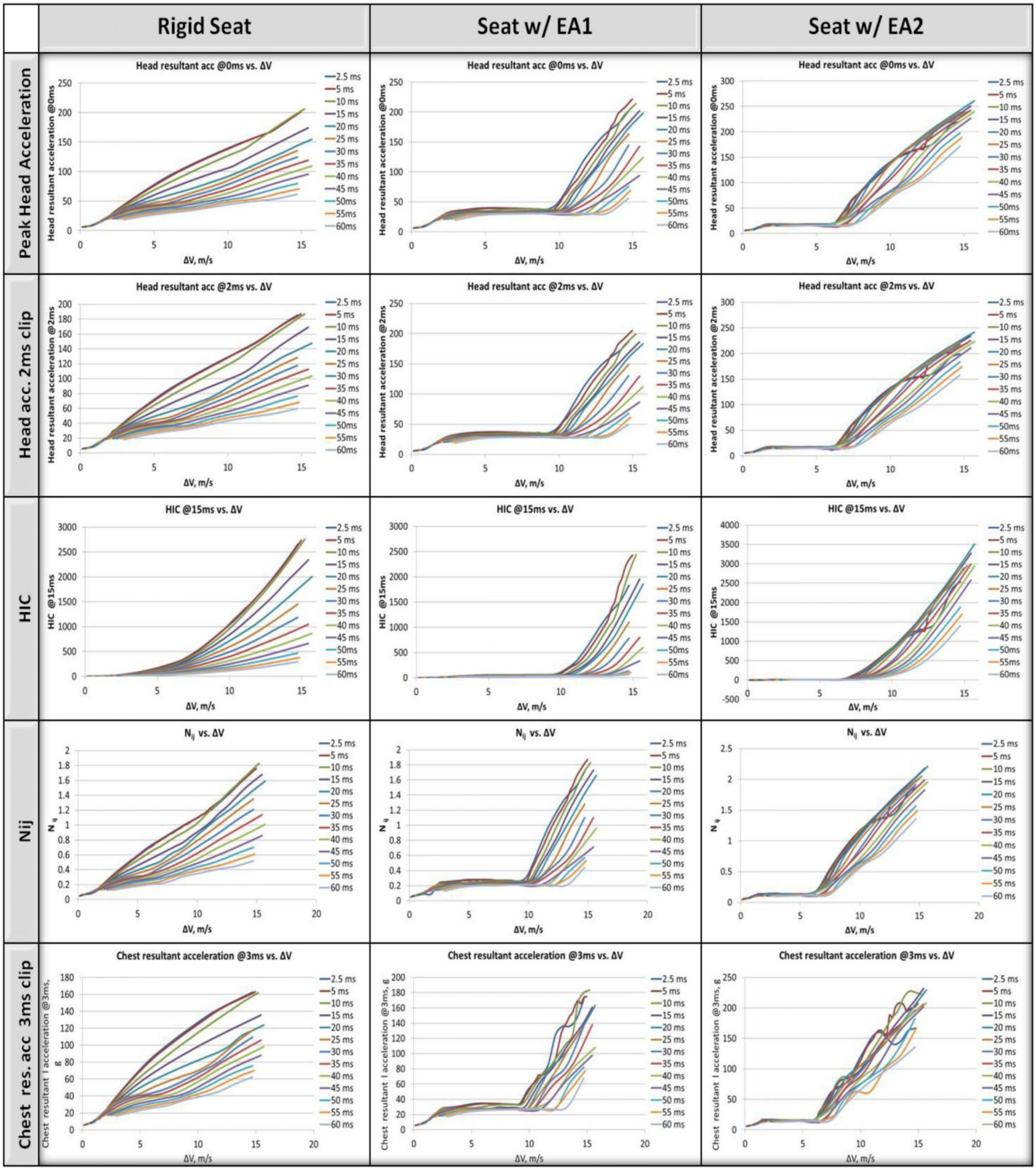


Fig A1-a. Occupant injuries vs. blast loading parameter for three different seat types

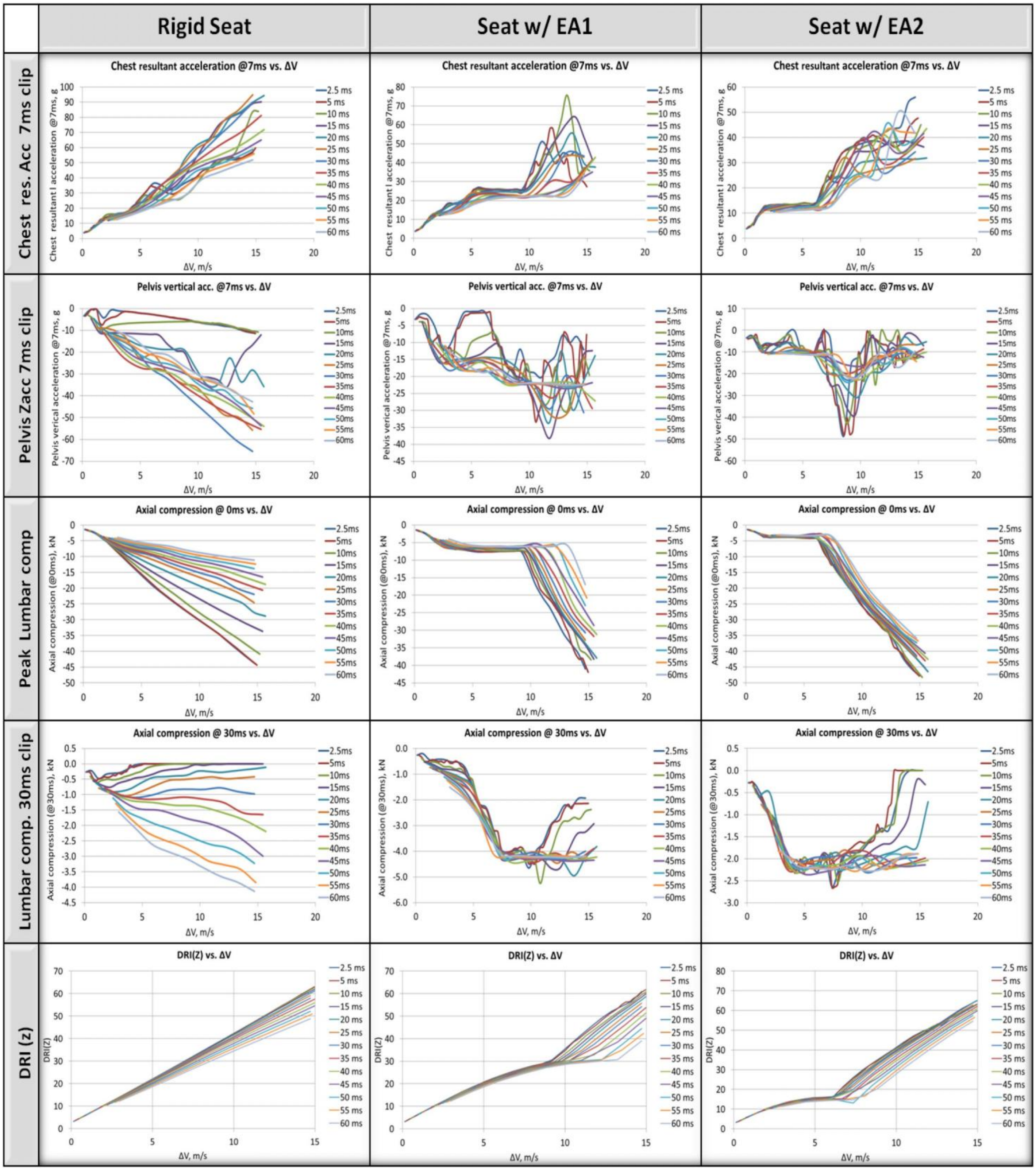


Fig A1-b. Occupant injuries vs. blast loading parameter for three different seat types

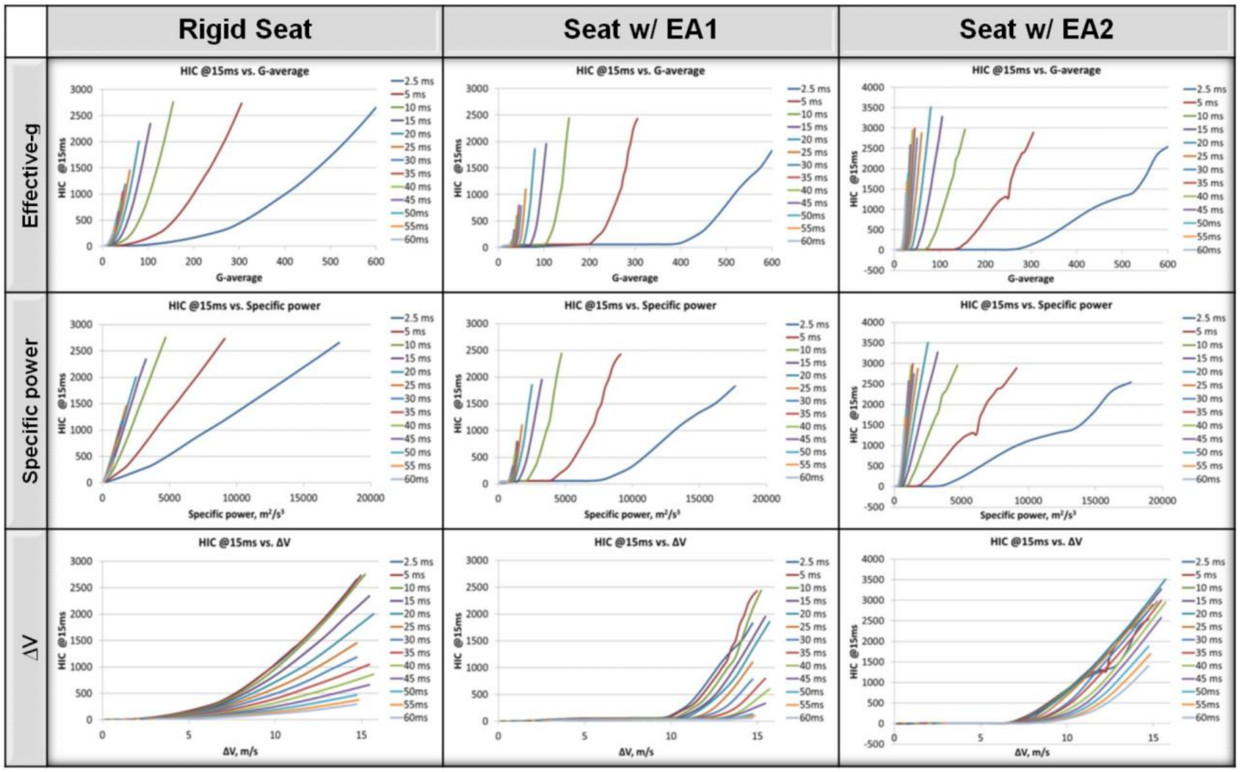


Figure A2-a. HIC15ms vs. three different blast loading parameters for three different seat types

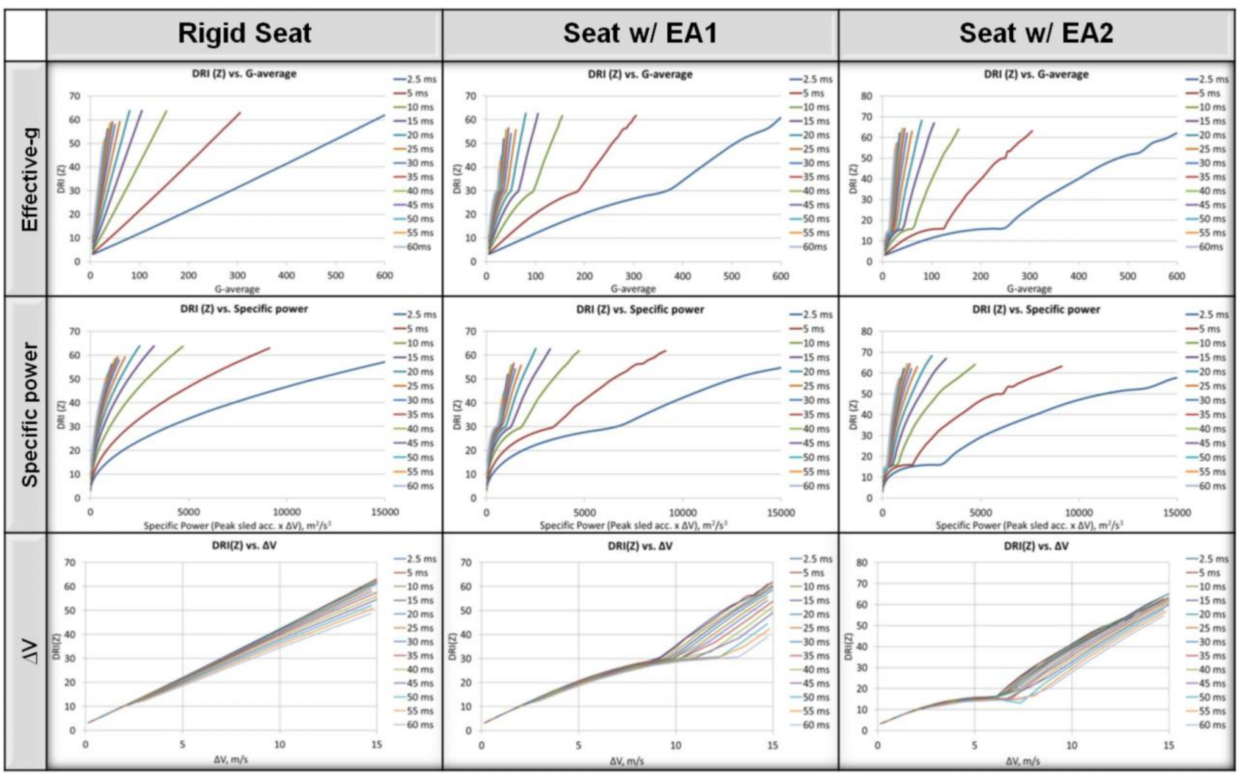
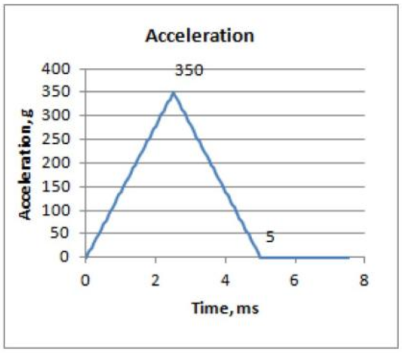


Figure A2-b. DRI vs. three different blast loading parameters for three different seat types

Table A1-1, A1-2, A1-3. Blast injury predictor tool

RDECOM | Occupant Injury estimator for vertical blast sled | TARDEC



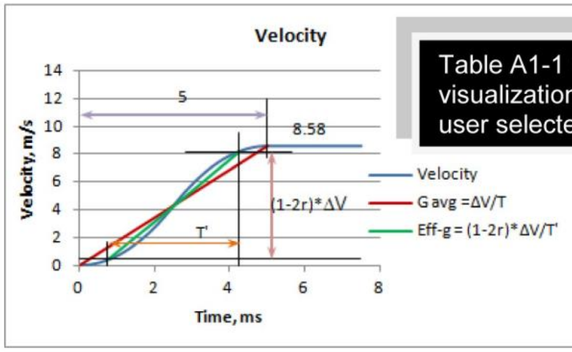


Table A1-1 Graphical visualization of the user selected pulse

Enter the value of time duration of pulse in milliseconds, and peak acceleration in g's in the yellow shaded cells. Conversely you can use the arrow buttons to desired values such that $1.5 \le \Delta V \le 15$ m/s. Please note that the solution range is only valid for the following ranges;

$1.0 \le T \le 60$

$10 \le G_{peak} \le 1200$

Table A1-2 User input

#	Blast load parameters	Value
1	Peak accleration in g's, G_{peak}	**350**
2	Time duration of pulse, T, ms	**5**
3	"r" factor, % (default 5%)	5
4	ΔV, m/s	**8.58**
5	Adjusted ΔV for effective-g, m/s	7.73
6	Adjusted time for effective-g, ms	3.419
7	G-Average $= (G_{Peak}/2)$	175
8	Effective-g $= G_{Average} \times (1+\sqrt{2r})$	230

Computed blast load parameters

Table A1-3 Predicted Occupant Injury values

#	Occupant Injury Criteria		Seat EA system		
	Criteria	Reference values	None	EA1 (15kN)	EA2 (7.5 kN)
1	HIC @15ms	350, 700, 1050 (Low, moderate, high risk)	705	29	384
2	Head resultant acceleration (2ms clip)	150g	110.8	33.9	89.8
3	Head resultant acclelaration (Peak)	180g	117.9	36.2	95.1
4	Neck injury criteria, Nij	<1	0.97	0.25	0.77
5	Chest resultant acceleration (3ms clip)	60g	111.2	34.1	76.3
6	Lumbar spine compression (Peak)	6672 N	25,773	8,103	17,433
7	DRI_z	15, 18, 23 (low, moderate, high risk)	37	30	32

Table A2-a. Comparison of injury values obtained from a single parameter method (Approach #1) using regression equation (Eqn. 1) to those obtained from actual M&S

	Pulse #1			Pulse #2			Pulse #3			Pulse #4			Pulse #5		
Occupant Injury	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff
Head resultant acceleration 2ms-clip, g	41.1	33.6	-18%	70.6	52.8	-25%	113.3	87.7	-23%	127.7	118.6	-7%	114.6	145.0	27%
Peak Head resultant acceleration, g	43.4	35.9	-17%	76.1	56.1	-26%	122.3	92.9	-24%	139.8	125.4	-10%	121.1	153.2	26%
HIC 15	49.7	13.2	-74%	163.7	140.9	-14%	710.5	732.0	3%	1420.6	1637.7	15%	1105.1	2701.2	144%
Nij	0.33	0.3	-11%	0.6	0.5	-17%	1.0	0.8	-14%	1.2	1.1	-8%	1.2	1.4	19%
Chest resultant acceleration, 3ms clip, g	39.3	39.3	0%	69.9	60.2	-14%	109.8	98.4	-10%	120.2	132.1	10%	106.9	161.0	51%
DRI	13.8	13.7	-1%	22.0	21.6	-2%	37.2	36.0	-3%	50.7	48.6	-4%	58.1	59.5	2%
Peak Lumbar Compression, kN	-8.6	-7.6	-11%	-15.6	-11.8	-24%	-26.5	-19.3	-27%	-29.6	-26.0	-12%	-21.2	-31.7	50%

Table A2-b. Comparison of injury values obtained from a single parameter method (Approach #1) using regression equation (Eqn. 2) to those obtained from actual M&S

	Pulse #1			Pulse #2			Pulse #3			Pulse #4			Pulse #5		
Occupant Injury	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff
Head resultant acceleration 2ms-clip, g	41.1	40.1	-3%	70.6	65.3	-7%	113.3	111.3	-2%	127.7	152.0	19%	114.6	186.8	63%
Peak Head resultant acceleration, g	43.4	42.9	-1%	76.1	69.8	-8%	122.3	118.8	-3%	139.8	162.1	16%	121.1	199.2	64%
HIC 15	49.7	12.9	-74%	163.7	140.0	-14%	710.5	729.2	3%	1420.6	1632.4	15%	1105.1	2693.2	144%
Nij	0.33	0.3	-5%	0.6	0.6	-5%	1.0	1.0	3%	1.2	1.4	11%	1.2	1.7	45%
Chest resultant acceleration, 3ms clip, g	39.3	39.5	1%	69.9	67.3	-4%	109.8	110.7	1%	120.2	141.4	18%	106.9	162.0	52%
DRI	13.8	13.9	0%	22.0	22.2	1%	37.2	37.4	1%	50.7	50.8	0%	58.1	62.3	7%
Peak Lumbar Compression, kN	-8.6	-8.9	3%	-15.6	-14.8	-5%	-26.5	-25.7	-3%	-29.6	-35.3	19%	-21.2	-43.5	105%

Table A2-c. Comparison of injury values obtained from the predictor tool (Approach #2) to those obtained from actual M&S

	Pulse #1			Pulse #2			Pulse #3			Pulse #4			Pulse #5		
Occupant Injury	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff
Head resultant acceleration 2ms-clip, g	41.1	41.6	1%	70.6	66.4	-6%	113.3	112.9	0%	127.7	126.4	-1%	114.6	115.0	0%
Peak Head resultant acceleration, g	43.4	45.0	4%	76.1	71.0	-7%	122.3	120.1	-2%	139.8	133.4	-5%	121.1	120.8	0%
HIC 15	49.7	13.0	-74%	163.7	147.0	-10%	710.5	743.0	5%	1420.6	1336.0	-6%	1105.1	1230.0	11%
Nij	0.3	0.3	0%	0.6	0.6	-3%	1.0	1.0	3%	1.2	1.2	-2%	1.2	1.2	-2%
Chest resultant acceleration, 3ms clip, g	39.3	42.0	7%	69.9	68.8	-2%	109.8	113.0	3%	120.2	114.2	-5%	106.9	109.8	3%
DRI	13.8	14.0	1%	22.0	22.0	0%	37.2	37.0	0%	50.7	50.0	-1%	58.1	59.0	2%
Peak Lumbar Compression, kN	-8.6	-9.2	7%	-15.6	-15.1	-3%	-26.5	-26.3	-1%	-29.6	-27.5	-7%	-21.2	-21.9	4%

Table A2-d. Comparison of injury values obtained from the response surface metamodel (Approach #3) to those obtained from actual M&S

	Pulse #1			Pulse #2			Pulse #3			Pulse #4			Pulse #5		
Occupant Injury	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff	M&S	Predicted	% diff
Head resultant acceleration 2ms-clip, g	41.1	41.3	0%	70.6	70.5	0%	113.3	110.0	-3%	127.7	128.3	0%	114.6	114.3	0%
Peak Head resultant acceleration, g	43.4	43.6	1%	76.1	76.1	0%	122.3	121.0	-1%	139.8	139.7	0%	121.1	121.1	0%
HIC 15	49.7	51.0	3%	163.7	163.7	0%	710.5	705.9	-1%	1420.6	1440.0	1%	1105.1	1105.0	0%
Nij	0.3	0.3	0%	0.6	0.6	0%	1.0	1.0	0%	1.2	1.2	0%	1.2	1.2	0%
Chest resultant acceleration, 3ms clip, g	39.3	39.4	0%	69.9	69.9	0%	109.8	109.0	-1%	120.2	120.4	0%	106.9	107.1	0%
DRI	13.8	13.8	0%	22.0	22.0	0%	37.2	37.1	0%	50.7	50.7	0%	58.1	58.1	0%
Peak Lumbar Compression, kN	8.6	8.5	-1%	15.6	15.6	0%	26.5	26.1	-2%	29.6	29.4	-1%	21.2	21.5	1%