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Occupant-Centric Platform (OCP) Technology-Enabled Capabilities Demonstration (TECD)

Comparing the Use of Dynamic Response Index (DRI) and Lumbar Load as Relevant Spinal Injury Metrics

Presented at the ARL Workshop on Numerical Analysis of Human and Surrogate Response to Accelerative Loading, Jan 09 2014

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14. ABSTRACT
The two most commonly used injury criteria for Spinal injuries today are Dynamic Response Index (DRI) related to structural accelerations, usually of the seat pan, or even more directly, lumbar force measurements taken within the Hybrid-III ATD as the evaluation criterion. With respect to continued use of these two criteria for spinal injuries, this report examines the following aspects in detail:
1) Any existing correlation between Peak Lumbar loads and DRI for un-encumbered occupants, in the whole blast loading regime or at least within different loading regimes
2) Re-evaluate (1) for encumbered occupants, that is, with heavier upper torsos
3) Potential changes to DRI calculations and Injury Assessment Reference Value (IARV) thresholds for encumbered occupants
4) General discussion on continued use of DRI as a design criterion for spinal injuries given the availability of the more direct Lumbar load from fully encumbered ATDs in underbody blast testing.

15. SUBJECT TERMS
DRI, Lumbar Load, Blast, LSDYNA, MADYMO, occupant, injury, pelvic, IARV, Encumbered, Mertz, Geertz, Brinkley, lumbar, spine, Anton

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Comparing the Use of Dynamic Response Index (DRI) and Lumbar Load as Relevant Spinal Injury Metrics

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Comparing the Use of Dynamic Response Index (DRI) and Lumbar Load as Relevant Spinal Injury Metrics

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Workshop on Numerical Analysis of Human and Surrogate Response to Accelerative Loading
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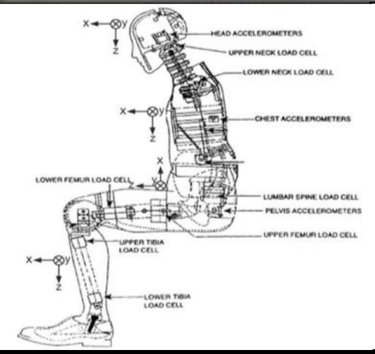
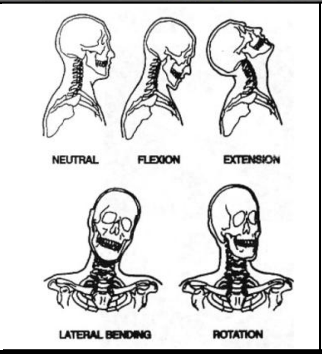
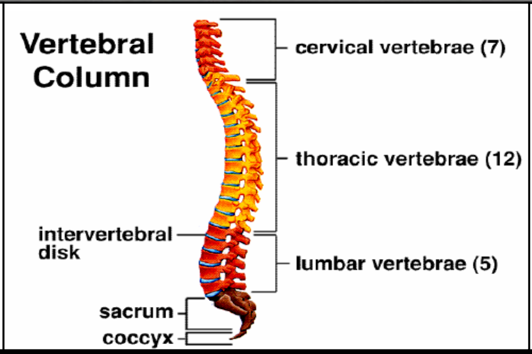
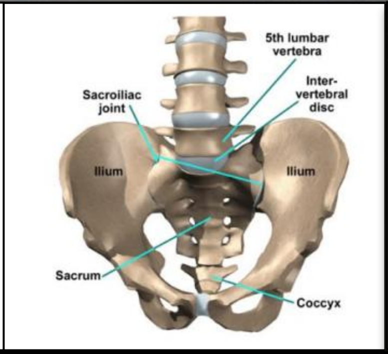
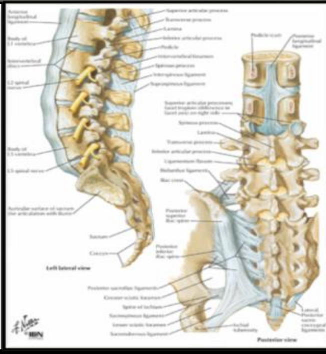










OUTLINE

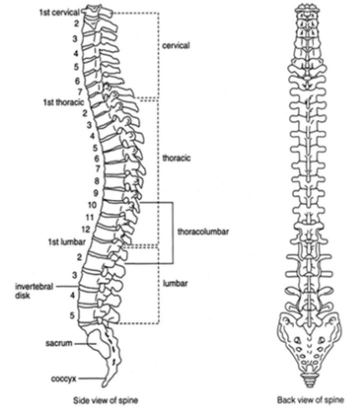
- Mechanical and Injury Models for DRI
- Mechanical and Injury Models for Lumbar Load
- DRI and LL: Temporal Behavior
- M&S Model Descriptions
- Behavior of Peak Compressive LL vs. DRI Cross-plots
- Proposal for Mechanical Model for Encumbered DRI
- Known Issues with DRI
- Summary / Conclusions

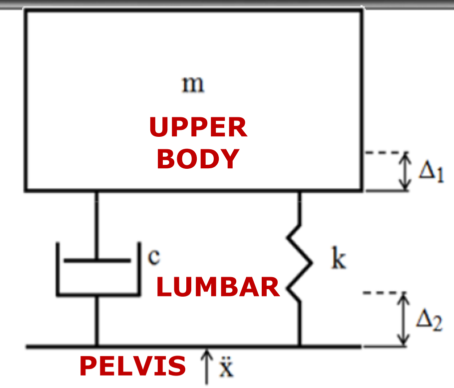
Dynamic Response Index (DRI) – Mechanical Model

- Simple lumped mass parameter model (single spring-mass-damper) to simulate the biomechanical response of the human upper body/vertebral column/pelvis [2,3]
- Values of **m, k, c** (and thus ω_n, ζ) were derived by compressive strengths of individual vertebrae [1], and load-deflection curves [4]
- Values established for a representative population of Air Force pilots with a mean age of 27.9 years [3]

m = 34.51 kg
k = 9.66E04 N/m
c = 818.1 Nsec/m

- ω_n = 52.9 rad/s, and $\zeta = 0.224$
- Lumbar Force = **k*δ**
- Maximum Lumbar Force = **$k*\delta_{max}$**

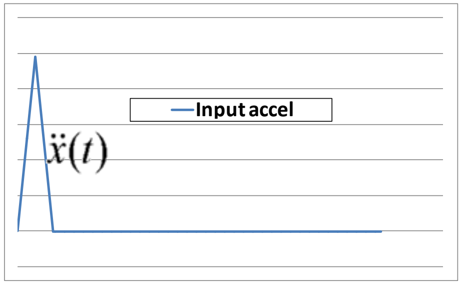


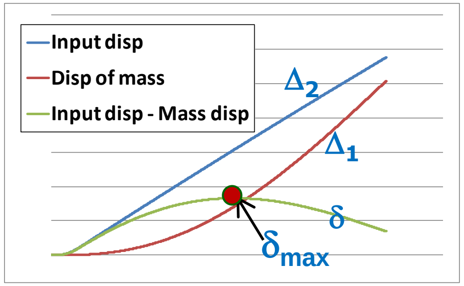


$$\ddot{x}(t) = \ddot{\delta} + 2\xi\omega_n\dot{\delta} + \omega_n^2\delta$$

- δ is the relative displacement between the upper body and pelvis ($\delta = \Delta_1 - \Delta_2$)
- ζ is the damping coefficient[33] (0.224) $\zeta = c/(2*sqrt(m*k))$
- ω_n is the natural frequency[33] (52.9 rad/s) $\omega_n = sqrt(k/m)$
- **Normalized Lumbar Force**
  = $(k*\delta_{max}) / (mg) = \omega^2_n * \delta_{max}/g$

$$DRI = \frac{\omega_n^2 \delta_{max}}{g}$$

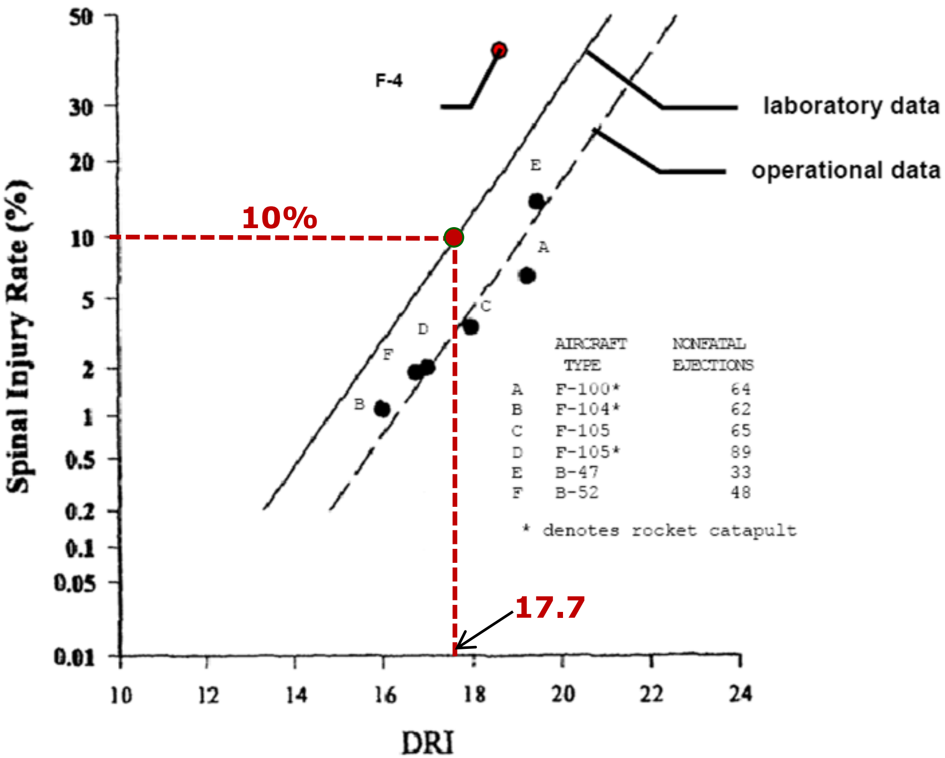




Maximum Lumbar Force, when normalized by the weight m*g, is called DRI

Dynamic Response Index (DRI) – Injury Risk Model

- During World War II, Geertz generated data on compressive vertebral strengths either with individual vertebrae or vertebral complexes of PMHS between 19 and 46 years old [1]
- Stech and Payne [3] used the above to relate the DRIz to an injury risk of 50% vs age, and for an average age of 27.9 of Air Force pilots, estimated a DRIz of 21.3 (7220N/1622 lbf). Brinkley used a normal distribution around this to set up the **laboratory data curve** [5]
- DRI value of 17.7 leads to a 10% risk of spinal injury (corresponds to 5992 N/1346 lbf)
- Injury model based on the laboratory data curve has the pelvis as point of initiation, so as far as possible, the pelvic acceleration rather than the seat acceleration should be used to calculate the DRI [13]

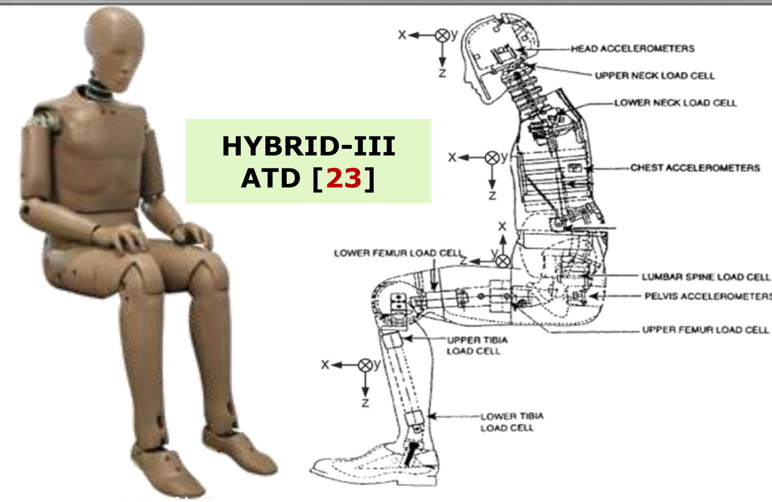


Spinal Injury Risk Calculated from Laboratory and Operational Data valid for AIS 2+ Injuries [3,5] .

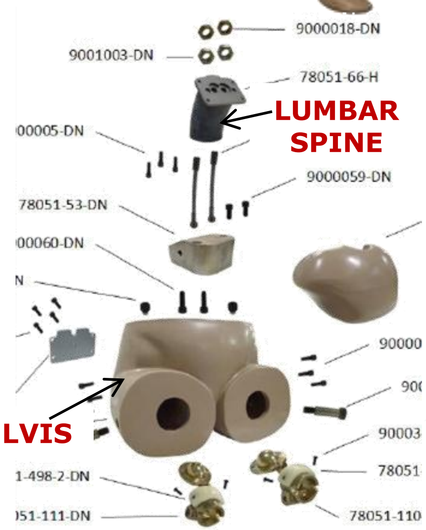
- **Quasi-static testing on PMHS specimens led to a 10% risk of spinal injury for DRI = 17.7**
- **Underlying principles for DRI model are based on lumbar load**

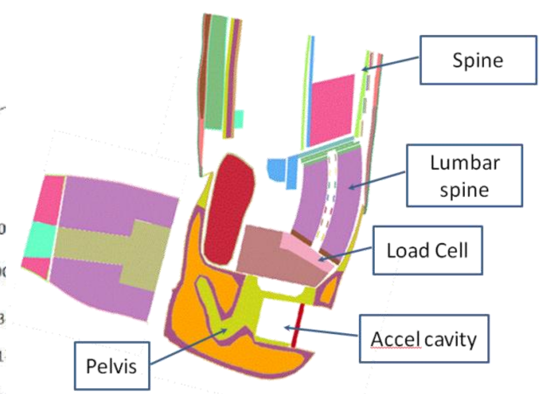
Compressive Lumbar Load (LL) – Mechanical Model

- Development of anthropomorphic test devices (ATD) and subsequent addition of load transducers in them represents a revolutionary increase in capability [20]
- Curved lumbar spine is incorporated to replicate typical seated automotive occupant positions, also used in military vehicle applications
- Three-Axis Lumbar Spine Load Cell measures time-dependent forces/moment at desired sampling rates during blast/crash
- Lumbar load cell did not adversely affect measured accelerations and forces, nor modify the spinal flexural characteristics [17]
- ATDs capable of producing reproducible results in greater detail under controlled testing conditions
- Biofidelic enhancements to the Hybrid III design were made which support its use in predicting human injury during high-speed dynamic events [20]



HYBRID-III ATD [23]

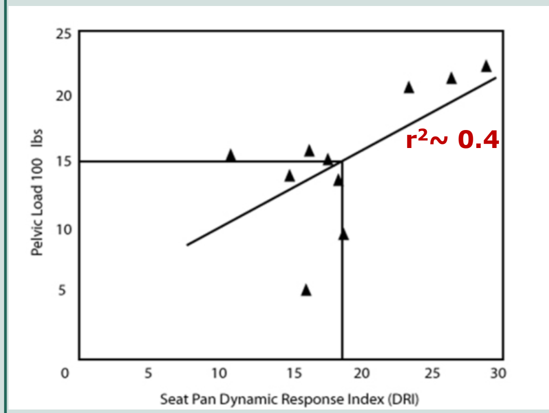
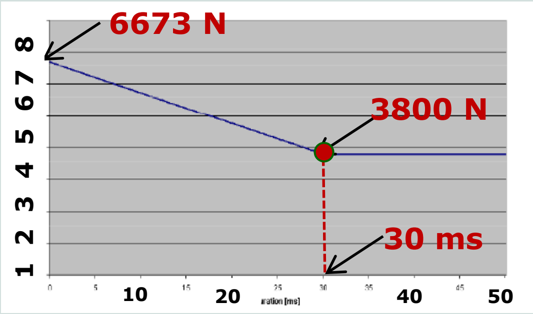


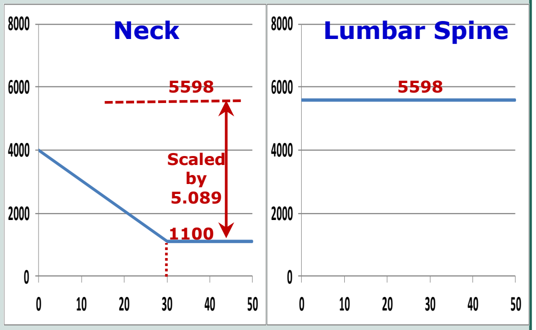


Part 572 Hybrid-III Lower Torso Assembly [23]

Lumbar Force measured here is a "direct" representation of lumbar response/injury

Compressive Lumbar Load (LL) – Injury Risk Model

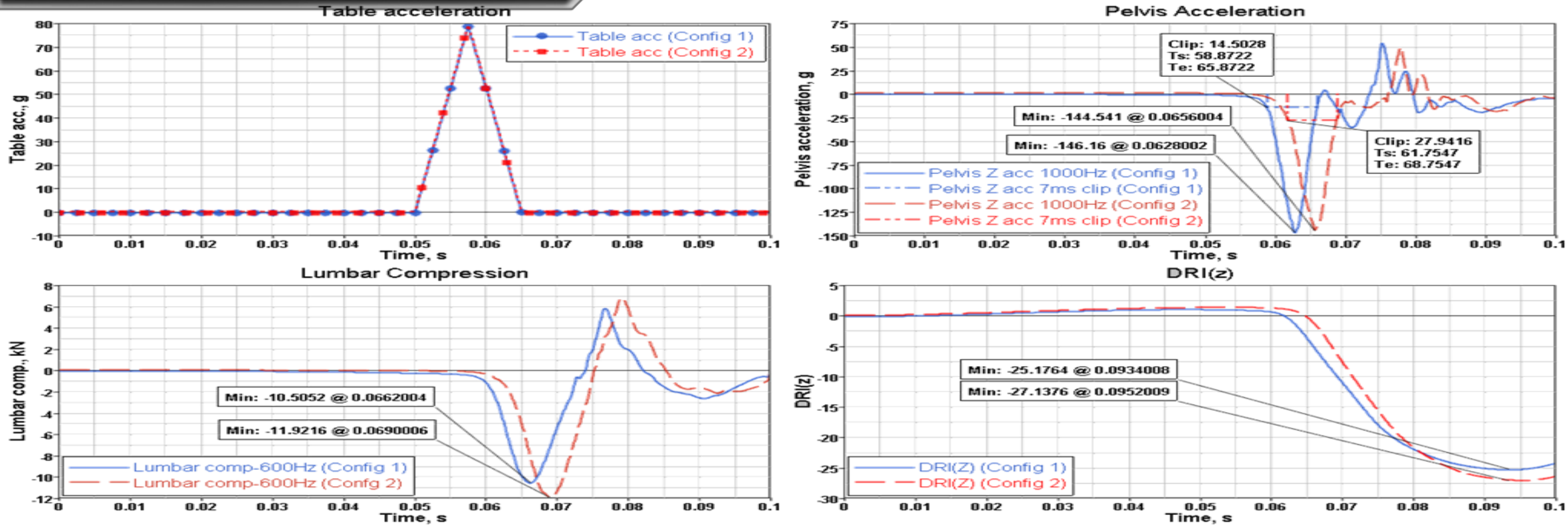
<table>
<tr><th>Source</th><th>Chandler [7,10]</th><th>Tremblay [24], Ripple [25]</th><th>Mertz [18,21,22,12]</th></tr>
<tr><td>Approach</td><td>Proposed for aircraft seats
Derived a compression force criterion by correlating the DRI and maximum compression force measured on a H-II lumbar load spine cell in 12 tests
1500 lb / 6675 peak value corresponds to a DRI of ~19
Adopted by FAA Regulations in Title 49/CFR 572</td><td>Proposed by Tremblay based on Ripple and Mundie's paper, but that paper doesn't specify any tolerance values, so not clear on origin
NATO RTO-TR-HFM-090 suggests that these criteria arise from Mertz criteria [19] for a scaling factor of 3.4-3.8, which also does not match the paper.
Perhaps Tremblay meant to refer to Alem [6], who also refers to a factor 3.4 that was used on Mertz's neck data in estimating 6675 N for peak lumbar load</td><td>Tolerance curves for compressive neck loading in high school football players and the adult populace using a H-III ATD outfitted with a football helmet impacted by a tackling block
Scaling factor from neck to lumbar based on waist and neck dimensions
Limiting force rationale (ratio applied to large-duration value); more conservative in mitigating lumbar spine injuries</td></tr>
<tr><td>ATD</td><td>H-II Straight Lumbar Spine</td><td>Unknown</td><td>H-III Curved Lumbar Spine</td></tr>
<tr><td></td><td>
</td><td>IARV for Compression Force


→ Duration of loading over given force level (ms)</td><td>IARV for Compression Force


→ Duration of loading over given force level (ms)</td></tr>
<tr><td>Peak LL Criterion (C)</td><td>1500 lb / 6675 N</td><td>1500 lb / 6673 N</td><td>1258 lb / 5598 N</td></tr>
</table>

Different approaches, but they lead to similar injury criteria for Compressive Lumbar Load

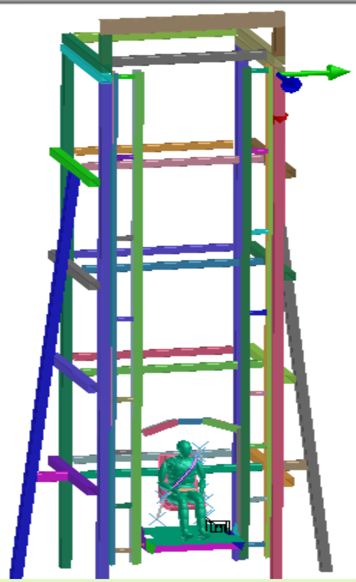
DRI and LL: Temporal Behavior



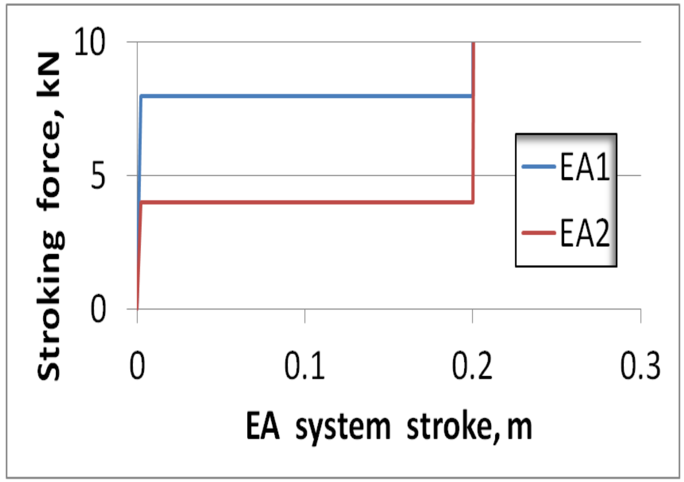
- Same pulse applied to two slightly different (both rigid) seat configurations in M&S
- Direct value of LL for C1 is 10.5 KN, indirect value (from DRI) is 8.5 KN; In addition, time of occurrence of the peak is quite different (8.7 ms after table peak for LL vs. 35.9 ms for DRI)
- The larger shift means that later changes in Pelvic accel will still affect DRI, but not peak LL
- Peak LL for C2 shifted by 2.8 ms vs. C1, and value higher by 13%
- Pelvic acceleration peak for C2 shifted by 2.8 ms, only 1% less, but wider (based on 7 ms clip)
- DRI for C2, calculated from pelvis acceleration, is shifted by 1.8 ms, value higher by only 8%

DRI does not track direct LL well in the time domain and can be affected by minor late data

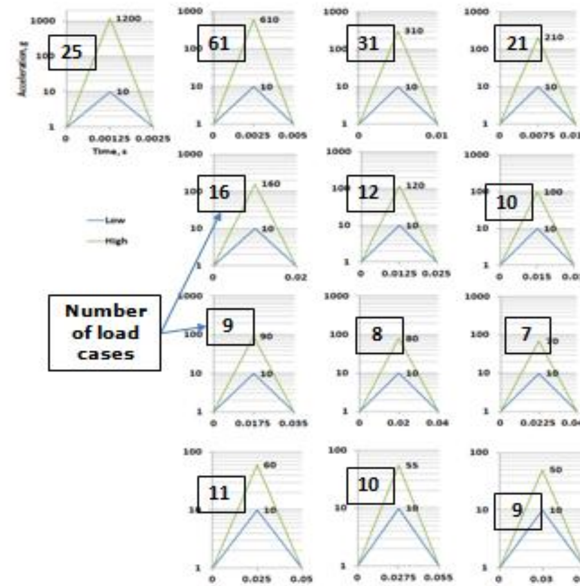
M&S Model Description - MADYMO



MADYMO Dynamic simulation model including Q-version of AM50 H-III ATD



In addition to Rigid Seat, two other EA seats (4 and 8 KN limiting force) were also included

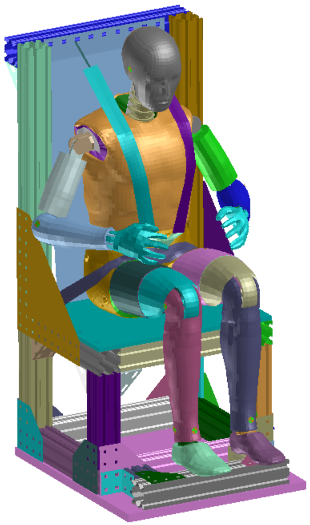


230 Triangular blast inputs were used for each of the 3 seats

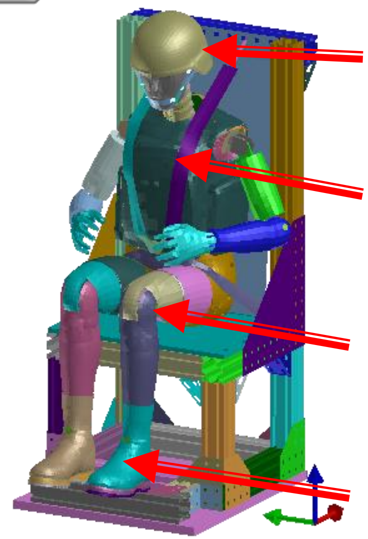
- A triangular blast wave pulse was applied to the vertical drop tower/sled.
- For unencumbered occupant: A total of eleven duration levels are studied; from 2 ms to 60 ms. At each ΔT, peak deceleration was varied from 10g to 1200g in 10g increments up to the point when Δv reached ~15m/s, where $\Delta v = 0.5 \times$ Peak acceleration$\times \Delta T$ (230 runs for each seat type)
- ~30 kg of encumbered PPE mass on the typical AM50 soldier was lumped on the upper torso, and 31 simulations were run covering Peak acceleration between 10-360g, **ΔT** between 5-15 ms (corresponding to a Δv between 1.5-12 m/s) for encumbered occupants

Systematic study was performed on MADYMO ATD setup for a large sample of input pulses

M&S Model Description - LSDYNA



Un-Encumbered AM50 H-III Occupant



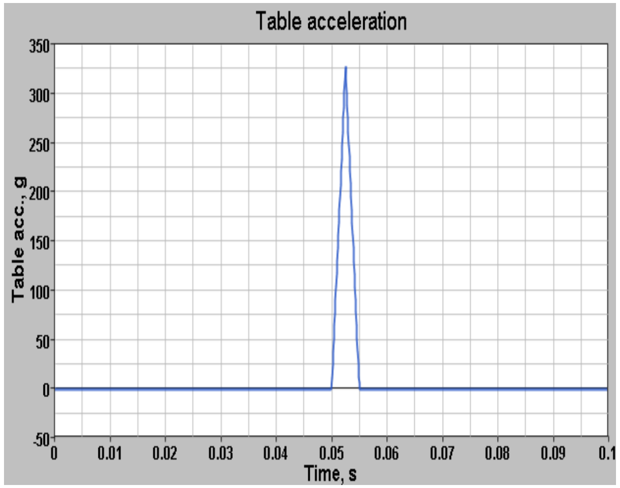
Helmet 1.4 kg

Upr Body PPE 29.4 kg

AM50 ATD 78.1 kg

Boots 2.3 kg

Encumbered AM50 H-III Occupant

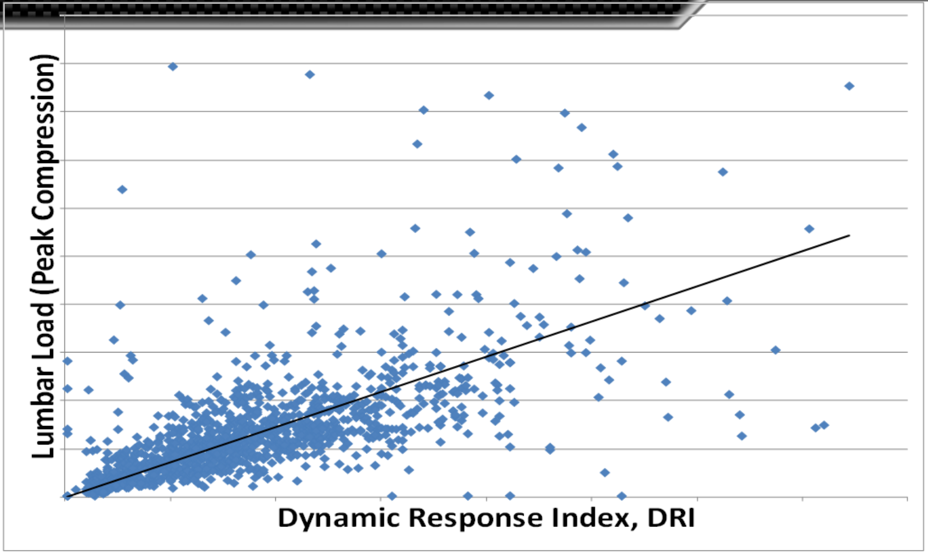


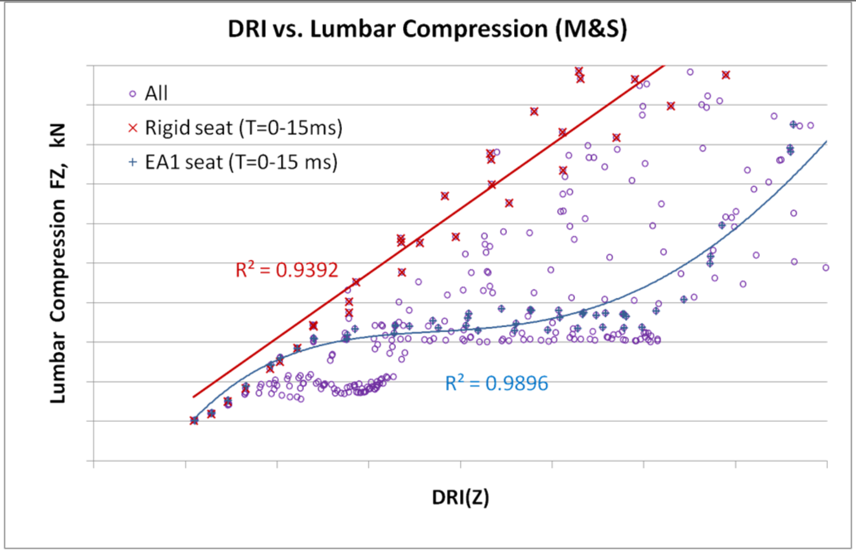
Triangular pulse applied to table of magnitude Ap and time duration ΔT

- A triangular blast wave pulse was applied to the vertical drop tower/sled.
- Only rigid seats were used in the LS-DYNA simulations, and the Humanetics version [23] of the LS-DYNA ATD (military version) were used
- For encumbered occupant studies, the vest and helmet were modeled in FEA using finite elements. The remaining PPE mass on the upper body of a typical AM50 encumbered soldier (~30kg - mass of vest) was lumped on the vest
- 31 simulations were run covering Peak acceleration between 10-360g, **ΔT** between 5-15 ms (corresponding to a Δv between 1.5-12 m/s) for both unencumbered and encumbered occupants

A reduced set of simulations were performed on LS-DYNA ATD setup

Behavior of Peak Compressive LL vs. DRI





Data from underbody mine tests (~1200 samples)

Data from MADYMO M&S (~700 samples)

- DRI and LL test data have been obtained for restrained occupants (usually encumbered) from a multitude of vehicles of different sizes and weights subjected to underbody mines of different sizes, positioned in different seats in different vehicle positions and configurations.
- While amount of scatter is reduced for M&S data, it is clear that there is a lack of a general overall governing relationship between DRI and Peak LL.
- When some other factors are also included, for example, only data for a seat type and a reduced range of DT, some patterns can be discerned in the M&S data.
- One interesting observation is that based on previously described IARVs, for 94% of the samples in test and 89% in M&S, DRI and LL both predict the same outcome (incapacitation, or no incapacitation)

Mechanical Model for DRI of Encumbered Occupant (DRI')

M corresponds to the weight to be added to m in DRI calculator to account for encumbrance on upper body of occupant. It is usually only a fraction of the actual physical weight of the encumbrance.

Example: M=30.4 kg

	Unencum	Encum #1	Encum #2
m	m	m+M	m+M
k	k	k	k
c	c	c	c
ω^2_n	k/m	k/m	k/(m+M)
DRI	$\frac{\omega_n^2\delta_{max}}{g} = \frac{(k*\delta_{max})}{mg}$	$\frac{\omega_n^2\delta'_{max}}{g} = \frac{(k*\delta'_{max})}{mg}$	$\frac{\omega_n^{2\prime}\delta'_{max}}{g} = \frac{(k*\delta'_{max})}{(m+M)g}$
IARV	17.7	17.7	$\frac{17.7}{(1 + M/m)}$
RI = DRI / IARV	$\frac{k*\delta_{max}}{17.7*mg}$	$\frac{k*\delta'_{max}}{17.7*mg}$	$\frac{k*\delta'_{max}}{17.7*mg}$

	Unencum	Encum #1	Encum #2
m, kg	34.51	64.91	64.91
k, N/m	9.66E4	9.66E4	9.66E4
c, Ns/m	818.1	818.1	818.1
ω^2_n	2799.2	2799.2	1488.2
DRI	285.34* δ_{max}	285.34* δ'_{max}	151.86* δ'_{max}
IARV	17.7	17.7	9.4
RI = DRI / IARV	16.1* δ_{max}	16.1* δ'_{max}	16.1* δ'_{max}

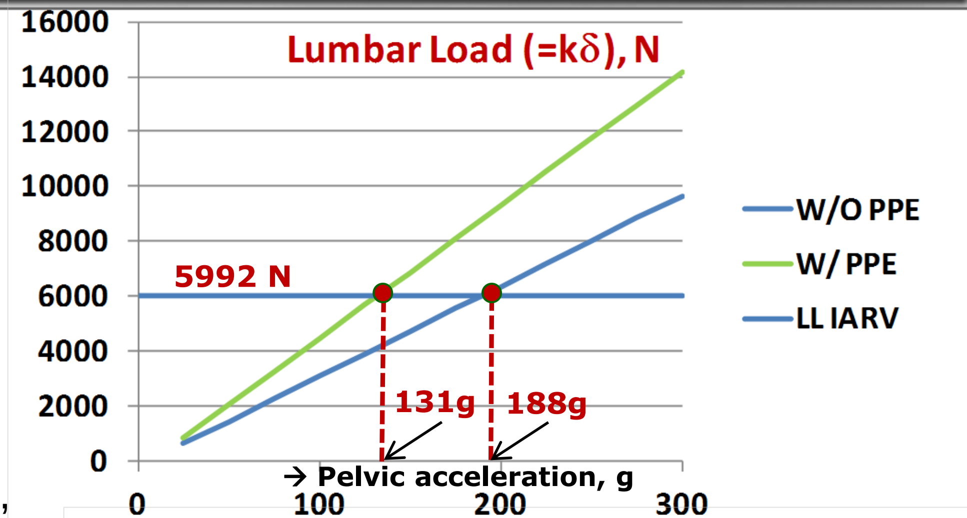
- The Mass quantity in the DRI SDOF calculator MUST be increased to compensate for the encumbrance (actual factor to be used (<1) on added mass is under review)
- Approach #1 is strongly preferred since the familiar IARV values (17.7) are still the same

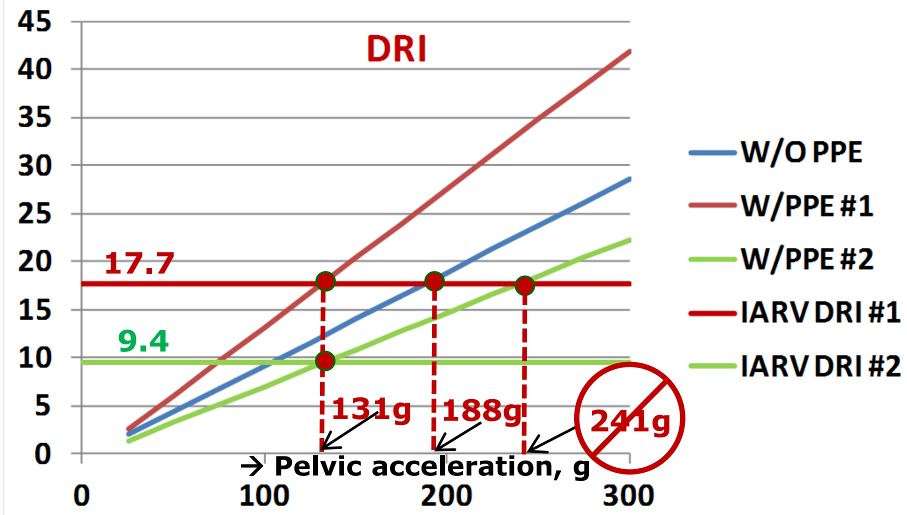
DRI for Encumbered Occupants (DRI') - Example

Let's calculate DRI for unencumbered (m=34.51 kg) and encumbered (m=64.91 kg) occupants based on the SDOF system, i.e, assuming that an added mass of <u>*30.4 kg*</u> to the upper body *fully affects lumbar load and DRI*.

The analysis is being done for triangular pelvic accelerations all of fixed duration 5 ms, but the amplitude is allowed to vary from 25-300g

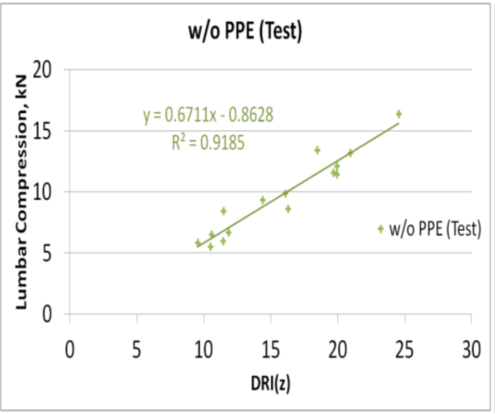
- From the max allowable lumbar load of 5992 N, it can be seen that a max pelvic acceleration of 188g and 131g can be withstood, for the unencumbered and encumbered occupants.
- Depending on the normalization constant used in determination of the encumbered DRI, the correct corresponding IARV must be used to get consistent and accurate results.
- Because the lumbar load (k*δ) is uniquely defined, it is a good idea to verify that DRI results are consistent with lumbar load.

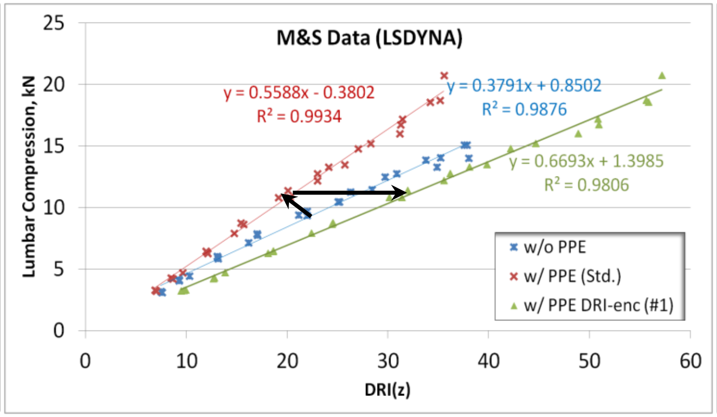


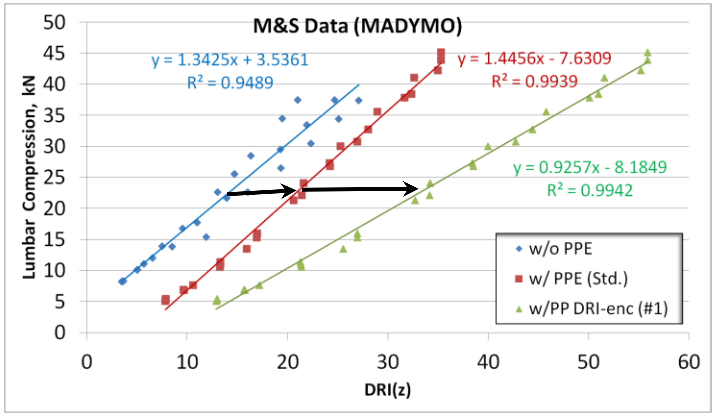


Depending on how the DRI calculator is coded and which normalization factor is used, the DRI value will be different and must be compared against the right IARV

Behavior of Peak Compressive LL vs. DRI







LL vs DRI behavior from drop tower tests (left), LSDYNA M&S (middle) and MADYMO (right) M&S

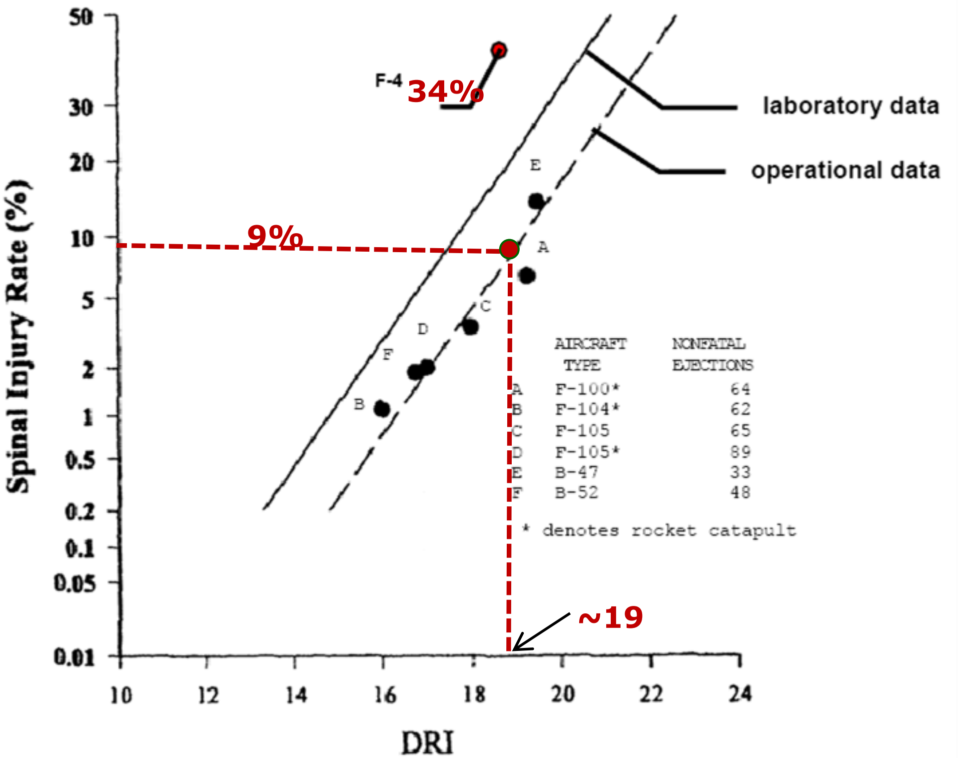
	Peak Acceleration,g	Duration, ms	ΔV, m/s
M&S	10 - 360	5 - 15	1.5 - 12
Test	3 - 285	5, 20	3 -7

- For a specific seat and limited ΔT range, DRI-LL relationship is linear in both tests and M&S
- For the LS-DYNA results, when the pelvic accels from increased upper body weights are run through the DRI calculator without changing the mass, the DRIs are seen to drop (red curve), which is not realistic, since the lumbar loads increased. Using DRI calculations as per #1, shifts to green curve, indicating higher DRIs as expected.
- For MADYMO results, because all the added weight affects pelvic accel, the DRI curve moves from blue to red, and increases even further as per #1. This indicates that a smaller factor (<1) needs to be applied to the physical added mass in order to accurately capture vest separation.

DRI (properly calculated) and LL both go up for encumbered occupants

DRI - Applicability to Ejection Seats

- Stech and Payne also presented the injury risk for operationally experienced non-fatal spinal injuries in ejection seat tests, shown as **operational data curve** [3, 5?]
- F-4 operational data [5] does not match the injury trend. For the F-4 DRI value of ~19, the operational data curve yields 9% risk of injury, not the observed 34% in reality.
- DRI-Injury Rate Relationship is only valid for misalignments of the seat with respect to the catapult direction < 5 degrees, which was not true for the F-4 seat
- In a survey of 223 ejections by British aircraft pilots over 1968-83, Anton [8] found a poor agreement between the incidence of spinal fracture and the DRI for ejections from 5 out of 6 ejector seats and concluded that predictors such as DRI "**have no apparent practical utility**" [11]



Spinal Injury Risk Calculated from Laboratory and Operational Data valid for AIS 2+ Injuries [3,5] .

Ejector Seat Data raises doubts as to suitability of the DRI as an injury measure [11]

DRI – Other Known Issues

- The curved lumbar spine is incorporated in H-III ATD to replicate typical seated automotive occupant positions in military vehicle applications. This results in a misalignment of the accelerometer axes and the lumbar spine by about 21^0. The DRI, which by definition, assumes a straight lumbar spine, deviates in this case from its intent to be an indicator of lumbar force.
- The DRI represents a whole body motion criterion which represents a load criterion instead of an injury criterion. Load criteria are based on physical parameters which specify an external load on the human body (e.g. footplate intrusion), whereas injury criteria are established with physical parameters which describe the biomechanical response of the human body or its surrogate [**14**]. Neck and Lumbar loads are examples of injury criteria.
- The DRI model is based on unconstrained motion of a single constant reaction lumped mass. Restraint systems impede the vertical motion, especially for mine blast seats which extend the loading duration. The reaction mass is increasingly constrained during the duration of blast response. Because the model treats the whole body as a lumped mass, the seat geometry and restraints used in the test data are critical to achieve the same results [**9**].
- As noted in [**13**], the physical parameter which affects fracture is always force. Using a model which is based on another physical parameter causes less accuracy and can lead to contradictory results.
- Even though the DRI model is based on single-degree-of-freedom vibration, it has been found [**11**] that *even for continuous vibration*, at frequencies > 8.4 Hz, the response tends to decrease in proportion to $freq^2$, so the predicted stress on the spine decreases at 12 dB per octave. Consequently, when the DRI model is used for continuous sinusoidal motion, it erroneously indicates that excessively high accelerations are permissible at high frequencies.
- The DRI model lacks fidelity in regards to gender, weight, anthropometrics and age.

DRI – Other Known Issues (contd)

- In several mine protection trials, seat acceleration data have shown to have a high variation and a lack of reproducibility [13]. Although the DRI SDOF system is comparable to a filter and tends to smooth out the input acceleration, the variation of the seat acceleration input has a negative impact on the reproducibility of the DRI data. An apparent advantage that the DRI measure has, in that it can still be computed when ONLY the seat acceleration data is available, tends to get neutralized by the above finding.
- Additional helmeted and vest masses may cause the natural frequency and damping characteristics of the human to change, invalidating the model [9]
- The assumption of linearity of the DRI model is highly unrealistic. It has been shown [16] that the frequency characteristics of the upper human body are *distinctly* different at low and high amplitude accelerations. Furthermore, the same paper also points out that *in vitro* compression testing of L1-L2 spinal units have indicated a non-linear force-displacement curve. Such non-linear characteristics have been and can be easily incorporated into the ATD models and hardware for determination of more accurate lumbar loads.
- The simplified assumption of a single mass, stiffness, and damping value, and reliance on pelvic or seat acceleration over the full time duration, leads to the undesired behavior of the DRI being affected by late peaks and valleys in the input acceleration, significantly after the effect of the blast load has already occurred.
- The DRI, by the nature of its very definition, has limited number of variables that can be changed to account for any new research findings on lumbar spine behavior. In contrast, the continued development of end-to-end, full system underbody blast tools [15] and the determination of the LL from an detailed ATD provides a much better "upgrade path" to accommodate new emerging data and predict lumbar spine injuries.

Summary / Conclusions

- DRI has the attraction of being an apparently simple, tangible model which clearly had high utility before the advent of detailed ATDs that could produce reproducible results in controlled testing.
- While DRI can be calculated when only seat accelerations are available and indeed may be the only injury measure that can be calculated in such a case (like when an ATD is not used), the consistency and usefulness of such data is highly questionable due to the variability in the seat accelerations.
- Human responses are highly nonlinear, and to expect a simple linear model such as DRI to be capable of responding accurately to a wide range of shock amplitudes is highly unrealistic.
- DRI and LL responses are both dynamic, and the peak values may even be in the ball-park, but DRI lags far behind as to when the peak occurs due to the use of only one frequency characteristic. This can lead to unrealistic consequences where later changes in pelvic acceleration can affect the DRI.
- There is a lack of any kind of overall general correlation between DRI and LL.
- Requiring pelvic accelerations for accurate DRI calculations means ATD is required. In which case, the lumbar load can be directly measured and compared against its IARV.
- Calculating DRI for encumbered occupants can be tricky in that while it is clear that the increased mass increases the lumbar load, what factor to use on the actual physical mass is still not clear. Also, it is recommended that if DRI is used at all, that it be determined using the standard normalization constant so that the familiar DRI values are still preserved.
- The availability of force-based IARV injury criteria on direct measurements such as lumbar load, makes them highly attractive as candidates for incapacitation assessment for the lumbar region.

Too simple, Too many assumptions, Too many questions…. DRI had an important role 50 years ago in the evolutionary timeline, but has the since largely outlived its utility

….. Time to move to a more “direct” injury measure (Lumbar Load from detailed ATDs)

AIS	Abbreviated Injury Scale
AM50	American Male 50th Percentile
APG	Aberdeen Proving Grounds, Maryland
ARL	Army Research Laboratory
ATD	Anthropomorphic Test Device
ATEC	Army Test and Evaluation Center
COTS	Commercial-off-the-Shelf
DOF	Degree-of-Freedom
FEA/FEM	Finite Element Analysis/Method
g	acceleration due to gravity
H-II / H-III	Hybrid-II or Hybrid-III ATD
kg	kilogram, unit of mass; 1kg ~ 2.204 lb
lb/lbf	pounds, pounds of force; 1lbf ~ 4.45 N
IARV	Injury Assessment Reference Value
LL	Lumbar Load
LSDYNA	COTS structural dynamics software from LSTC, CA
LSTC	Livermore Software Technology Corporation, CA
ms	msec, milliseconds, unit of time (1 ms = 0.001 second)
M&S	Modeling & Simulation
MADYMO	MAthematical DYnamic MOdels (COTS software from TNO)
N	Newtons, unit of force, 1 N ~ 0.22472 lbf
OCP	Occupant-Centric Platform
PMHS	Post-mortem Human Specimens
R&D	Research & Development
RDECOM	Research, Development and Engineering Command
RI	Relative Injury Index = Injury Value / IARV
SimBRS	Simulation-Based Reliability and Safety
SDOF	Single Degree-of-Freedom
SLAD	Survivability and Lethality Analysis Directorate in ARL
TARDEC	Tank Automotive Research, Development and Engineering Center
T&E	Test & Evaluation
UBM	Underbody Blast Modeling/Methodology
WMRD	Weapons and Materials Research Directorate in ARL

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DISCLAIMER

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