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Assessing the performance of various restraints on ambulance patient compartment workers during crash events

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The inability of emergency medical service (EMS) workers to remain safely restrained while treating patients in the patient compartment of a moving ambulance has been identified as a key impediment to EMS worker safety in North America. It has been hypothesised that restraint systems designed to provide mobility while offering the ability to lock during an impact or sudden manoeuvre, could greatly enhance worker safety in the back of ambulances. Through a series of 33 sled and crash tests impacting the front, side, and rear of simulated and actual ambulance patient compartments, the National Institute for Occupational Safety and Health examined the biomechanical and kinematic effects of two-, four- and five-point restraints on 95th percentile male Hybrid III anthropomorphic test devices. Results indicate that the inclusion of restraint systems offering mobility have the potential to improve worker safety under many working conditions in this unique work environment.

Keywords: crash testing; restraints; neck injury; head injury criteria (HIC); side-facing seating

1. Introduction

Most ambulances built in North America prior to 2008 were outfitted with fixed, two-point lap belts on the primary work location – the side-facing bench seat. The geometry of this work environment, coupled with the assigned tasking, often compels the worker to remove the lap belt to attend to the patient. In an effort to improve this worker safety issue, the National Institute for Occupational Safety and Health (NIOSH) collaborated with the U.S. Army Tank-Automotive and Armaments Command and the Canadian Forces Health Services Group Headquarters to conduct a comprehensive evaluation of four commercial off-the-shelf mobility restraint systems. In total, a combined 33 front, side and rear impact sled and crash tests were conducted to evaluate the ability of each mobility restraint system to manage the energy generated by a restrained 95th percentile male anthropometric test device (ATD). Additionally, the team evaluated the impact each restraint had on the instrumented Hybrid III ATD. This study was conducted to determine if mobility restraints could improve emergency medical service (EMS) worker safety by providing patient compartment occupants with a higher level of protection than when unrestrained, while still allowing mobility to care for patients.

2. Background: design limitations

2.1. *Ambulance types and differences*

There are three ambulance types in the USA. A Type I ambulance mounts a patient compartment or box on a truck chassis. A Type II ambulance is a modified van with a narrow profile and raised roof to increase interior height. A Type III ambulance consists of a box mounted on a cutaway van chassis. The patient compartment geometry is similar in Type I and Type III ambulances, but significantly larger than Type IIs; thus, EMS workers are likely to require greater mobility when working in Type Is or IIIs than in Type IIs. Therefore, this project focused on improving crash protection in box-type patient compartments. Seating locations, illustrated in Figure 1, include a rear-facing attendant seat, a side-facing squad bench with provisions for three occupants, and occasionally a side-facing CPR seat.

Crash protection for driver's compartment occupants includes National Highway Traffic Safety Administration (NHTSA) mandated restraint systems installed by the chassis manufacturer [42]. Patient compartment occupants are not afforded the same protection especially when seated on the dual-purpose squad bench that doubles as a supine patient transport location. A two-point lap belt (e.g. a seat

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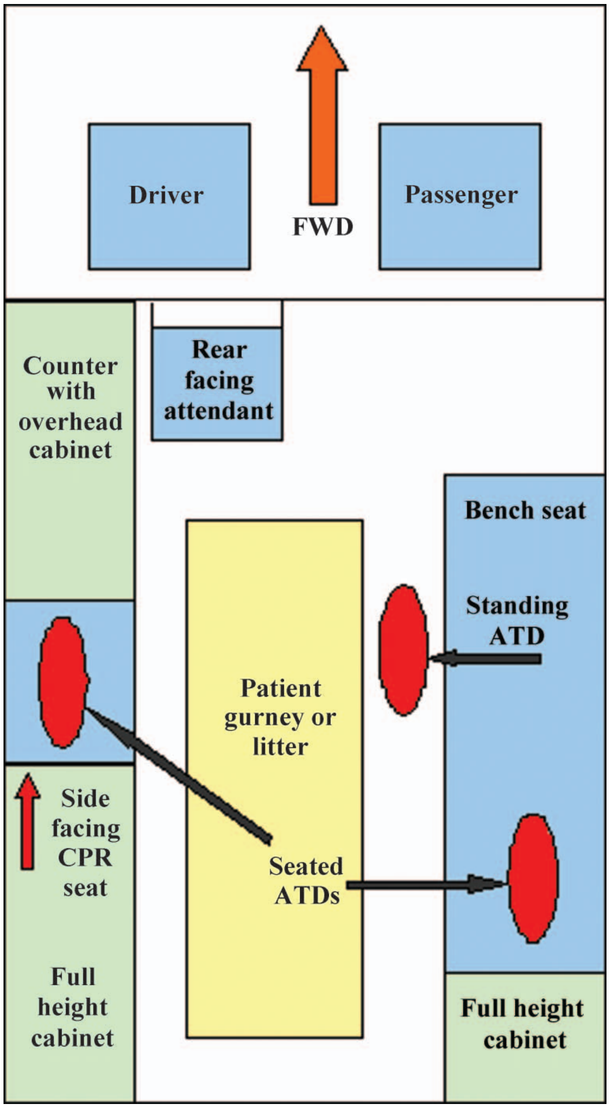


Figure 1. Patient compartment layout with ATD positions.

belt) is the crash protection commonly provided for squad bench occupants, as well as most other patient compartment seats. Proper use of a seat belt requires the occupant to be seated with his/her back against the seatback. This position prohibits moving to (1) the edge of the seat to access the patient, (2) reach across the compartment to access supplies and equipment, or (3) perform medical procedures that require standing or kneeling. Because of this, EMS workers routinely work unrestrained on the side-facing squad bench. Previous studies confirm that less than half of EMS workers use occupant restraints while in the patient compartment [22,31].

2.2. *Reach limitations*

To better understand the limitations experienced by workers seated on the bench seat, NIOSH evaluated the ability of a worker wearing a lap belt to reach two targets on the patient (left wrist and mouth) and three pieces of equipment typically found in an ambulance patient compartment (suction unit, radio and defibrillator) [15]. Simulations were conducted in an environment modelled from an ambulance purchased and manufactured in accordance with the Federal Specification for the Star-of-Life Ambulance [13]. Using simulated human forms representing the 5th percentile female and the 50th and 95th percentile male by height, findings indicated none of the human forms could reach the equipment while seated and restrained. Furthermore, only the 95th percentile male was able to reach both targets on the patient while remaining properly restrained. The 5th percentile female could not reach any of the five targets while restrained by the lap belt.

2.3. *Review of seat belt effectiveness in passenger vehicles*

Since the introduction of lap belts in automobiles, occupant restraint systems have evolved with the goal of improving occupant protection. Lap belts were introduced primarily to prevent occupant ejection. While successfully meeting this requirement, they did not provide upper torso restraint. As a result, in frontal crashes while the lower torso was restrained, the occupant's upper torso rotated violently forward about the lap belt in a 'jackknifing' manner. In some cases, the lap belt has been reported to have loaded the abdomen due to the slippage of the pelvis under the belt during crash (submarining), or being misplaced on the abdomen, resulting in severe injuries to the abdomen and the spine. This injury mechanism is commonly referred to as the 'seat belt syndrome' [12].

When combined with the jackknifing effect, the kinematics of a lap-belted occupant in a frontal crash introduced several injury mechanisms including head impact with the vehicle interior, abdominal organ injuries, lumbar spine injuries and cervical injuries [17,34,49,50]. Despite these hazards, the lap belt was more effective than being unbelted – the original design objective. Overall, the lap belt was 30%–40% effective in reducing abbreviated injury scale (AIS) 2–5 injuries and 25%–35% effective in reducing fatalities [14,27].

Accident data analysis indicated that improvements in restraint system effectiveness could be achieved by providing upper torso restraint that further reduced the excursion of the upper torso and head in the vehicle. This was accomplished by the introduction of three-point belts [29,37,40,51]. Continued refinement of the belt geometry, anchor locations, and seating surface geometry and stiffness have nearly eliminated submarining and unwanted neck loading by the shoulder belt [1,39]. In 1999, NHSTA reported that for rear seat occupants 'The change from lap to lap/shoulder belts has significantly enhanced occupant protection, especially in frontal crashes. . . . Lap/shoulder belts reduce the risk of both head and abdominal injuries in potentially fatal frontal crashes relative to lap belts only: head injuries by 47% and abdominal injuries by 52%' [28]. These findings are consistent with earlier statistics from General Motors researchers who indicated that the effectiveness of lap-belt-only restraints was less than half that of lap/shoulder belts (18% ± 9% compared to 41% ± 4%) [10]. Furthermore, NHTSA (1984) reported the lap/shoulder belt provided significantly greater effectiveness (40%–50% AIS 2–5 injuries and 45%–55% fatality) than a lap-belt-only restraint.

2.4. Belt-induced injuries: a review of NASS CDS data

When an occupant is seated side-facing relative to the crash acceleration vector with a shoulder strap restraining the upper torso on the shoulder towards the impact, the displacement of the upper torso is limited by the shoulder strap, while the head moves towards the impact. This causes the head to move laterally and rotate downward placing the neck into lateral bending, distraction and/or shear similar to the kinematics demonstrated in Figure 2, as the shoulder strap applies its restraining load at the base of the neck. While the reduction of injuries caused by impact with the interior achieved by such a restraint likely far outweighs the injuries caused by restraint loading and kinematics, further research into this neck loading condition is warranted. Military troops, who are routinely transported in aircraft side-facing seats while wearing five-point restraints, are subjected to this type of loading during a crash landing. Similarly, occupants in the centre rear seat of passenger cars equipped with lap/shoulder belts are subjected to a lateral impact loading to the neck when the vehicle is struck on the side on which the shoulder belt is located.

Figure 2. Seated side-facing ATD in five-point restraint at maximum excursion during 40-km/h frontal impact.

Each year NHTSA's National Automotive Sampling System (NASS) Crashworthiness Data System (CDS) investigates several thousand tow-away crashes. Review of the 1997–2007 NASS CDS database for side-impact crashes involving adult occupants restrained by lap/shoulder belts at the rear centre occupant position revealed that in only six cases was the shoulder belt positioned on the shoulder towards the side of the impact. In the five cases in which the delta-V was calculated, the delta-V was less than 24 kilometers per hour (km/h). While none of the occupants sustained a cervical injury, the data were insufficient to assess the potential for a serious shoulder-belt-induced cervical injury under this loading condition. However, 25 cases involving children 1–6 years of age restrained in forward-facing child safety seats with dual shoulder straps, during side impacts of 31 km/h delta-V or greater, were identified in the 1997–2007 NASS CDS database. The lateral delta-V for the cases ranged from 31 to 56 km/h, with a mean of 39 km/h. Since these child safety seats have shoulder straps on both shoulders, impacts to either side of the vehicle would create the head/neck kinematics described above. While it is widely recognised that young children have a lower threshold to cervical injury than adult occupants [2,41], of the children identified, none sustained serious cervical injury during these relatively high lateral delta-V crashes. Similar conclusions were reported in a review of German and Australian crash events involving restrained children covering the years 1996–2000 [11]. These data, while limited, suggest that occupants oriented side-facing to a crash vector, restrained by dual shoulder straps, are not likely to sustain serious cervical injury.

3. Discussion: performance requirements

The performance of five different restraint systems was evaluated using electronic data, standard injury analysis methods and a gross kinematic analysis of the visual data. Specific data channels collected varied by seating position due to instrumentation limitations. In general, head, chest, and pelvis accelerations and upper neck and lumbar loads were collected during each of the tests. Following each test, the post-test position of the occupants and condition of the occupant restraints were assessed and documented. Through a systematic analysis of the collected data, restraint performance and compartment design concerns were identified and evaluated. Specific test methods and results are discussed in Sections 4–7. However, prior to discussing the detailed test set-up and results, this paper will discuss the fundamental issue identified during this effort – that is, the determination of suitable injury assessment reference values (IARVs) for the various data collected.

While Federal Motor Vehicle Safety Standards (FMVSS) provide regulatory limits which vehicle manufacturers selling in the United States must meet, several concerns were identified that limited their application to the data collected under this effort. The FMVSS requirements focus on the performance of a 50th percentile male Hybrid III ATD in a near-frontal crash [42]. They also focus on the head injury criteria (HIC), chest acceleration, chest compression and femur loads. More recently, the FMVSS have been updated to include the 5th percentile female Hybrid III and assessment of neck injury using N_{ij}. N_{ij} is an assessment tool used to evaluate the combined effect of fore/aft bending and tension/compression on the upper neck. Additionally, the FMVSS 214 addresses side-impact requirements and testing using side-impact ATDs. At the time this testing was performed, the only ATD included in FMVSS 214 was the 50th percentile male side-impact ATD and performance was evaluated using thoracic trauma index (TTI) and lateral pelvis acceleration. More recently, the FMVSS have been expanded to include ATDs of other designs (5th percentile female) and additional regulatory limits for HIC, rib deflection, pubic symphysis, and abdominal and acetabular forces [44]. An excellent summary of past research as it relates to regulatory limits, IARV, and research value determinations for in-position and out-of-position frontal and side-impact occupants was provided by Mertz et al. in 2003 [26]. However, given the unique seating positions found in this environment, the rationale for each IARV or limit chosen is provided by body segment.

3.1. Anthropometric test device selection

Testing in each seating position evaluated was performed using 95th percentile male Hybrid IIIs. Though none of the occupant positions evaluated was a forward-facing seat, by virtue of testing side and rear impacts in addition to frontals, all seating positions were tested as if facing forward relative to the crash vector at some point in the test programme. ATDs representing the 95th percentile were selected since the goals of this programme were to investigate the ability of each restraint to protect the most challenging occupant size, to verify the structural integrity of the systems, to assess the space available for occupant ridedown and to compare each restraint system's impact on ATD-measured parameters. As was true in the testing completed by the Australian Department of Defence [32] and more recently by AmSafe [16], side-impact ATDs were not used for a variety of reasons: availability, the need to collect upper neck loading, the expectation the occupants would not remain aligned lateral to the impact, the desire to have arms on the ATDs and the need to have at least one of the ATDs in a standing position.

3.2. Head injury criteria

Generally speaking, most research related to head injury uses HIC to evaluate the injury potential from impact accelerations of the head. Both HIC and its predecessor, the Wayne State Tolerance Curve, are related to impact to the frontal portion of the non-helmeted head. In this test programme, head contacts occurred to various regions of the head. While HIC can still be calculated and is useful for comparison purposes, its correlation to injury is not as clear as when the contacts are to the frontal region of the head only. The FMVSS 208 now requires calculation of HIC using a maximum time interval of both 36 milliseconds (ms) and 15 ms and requires the calculated value to be less than 1000 and 700 for these intervals, respectively. For this programme, HIC was calculated for three different allowable time intervals: 36 ms, 15 ms and unlimited. However, HIC36 and HIC15 were the focus of the analysis.

Mertz provided a curve correlating HIC to the probability of life-threatening brain injury [25]. According to this curve, an HIC of 1000 equates to a 15% chance of life-threatening brain injury for impact situations only.

Kleinberger (1998) provides a formula equating HIC36 with the probability of skull fracture (AIS $\geq$ 2) [20]. Kleinberger's formula is

$$p(\text{fracture}) = N\left(\frac{\ln(\text{HIC}) - \mu}{\sigma}\right). \tag{1}$$

Again, the correlation of HIC36 to injury pertains to skull fractures and is most applicable in instances of head contact or impact situations.

Hodgson found that if a head impact does not contain a critical HIC interval of less than 15 ms, it should be considered safe as far as cerebral concussion is concerned [18]. This would imply that HIC intervals longer than 15 ms would not indicate a chance of concussion, though they may be useful in predicting skull fracture. In addition to looking at HIC, the peak head acceleration was also considered. While FMVSS 208 does not address peak head accelerations, FMVSS 201 (Occupant Protection in Interior Impact) and FMVSS 218 (Motorcycle Helmets) do place limits on head acceleration. FMVSS 201 provides the more conservative limit, requiring that head acceleration not exceed 80 g for more than 3 ms; this will be referred to as a 3-ms clip value.

3.3. Neck injury criteria

A variety of neck injury analytical methods were used to assess injury potential. Individual neck loads and moments were compared to the corresponding regulatory limit, accepted IARV or research value available in the literature. In addition, N_{ij} was calculated for all cases, but since N_{ij} considers fore/aft bending combined with tension/compression of the neck, its usefulness as an evaluation tool is limited to instances where neck loading occurs in the sagittal plane.

Table 1. IARVs for the large male neck.

Peak tension N (Ibs)	Peak compression N (Ibs)	Peak fore/ aft shear N (Ibs)	Peak flexion Nm (in-ib)	Peak extension Nm (in-ib)	Reference
5030 (1131)	4830 (1086)		415 (3673)	179 (1584)	NHTSA $N_{\rm ij}$ Ver. 10 Software
3300 (742)	4000 (900)	3100 (697)	190 (1680)	57.9 (504)	NHTSA (1985)
5030 (1131)	4830 (1086)		415 (3673)	179 (1584)	Eppinger (2000)

One of the primary foci for this programme was the protection of occupants seated on the side-facing crew bench because of its reported high occupancy rate during patient transport [3]. For bench-seated occupants, ambulance frontal crashes are lateral impacts and in many cases resulted in lateral bending of the neck. The available injury criteria for lateral bending of the neck are somewhat limited. However, a good overview, including proposed limits for the 50th percentile male, is provided by Soltis [35,36].

In addition to proposing lateral neck criteria, Soltis also proposed an adaptation of the neck injury criteria ($N_{\rm ij}$) used by NHTSA. What was proposed was a lateral $N_{\rm ij}$, which evaluates the combined effect of lateral bending and tension/compression of the neck. The Federal Aviation Administration later used this approach to conduct a preliminary injury assessment of side-facing aircraft seats, citing the similar biomechanical basis between lateral $N_{\rm ij}$ and the 'fore/aft $N_{\rm ij}$ criteria' [5].

3.3.1. Neck force and moment criteria

The values for the various other IARVs for the large male neck are summarised in Table 1. The IARVs provided by Eppinger were used to assess the peak upper neck flexion, extension, tension and compression as these are the same limits adopted for Version 10 of $N_{\rm ij}$ [8,9]. Fore/aft shear forces were assessed using the limits provided in the FMVSS 208 Docket [4].

3.3.2. N_{ij} analysis

Additionally, $N_{\rm ij}$ was calculated for each occupant using the following formula:

$$N_{\rm ij} = (F_z / F_{\rm z-int}) + (M_y / M_{\rm y-int}). \tag{2}$$

Intercept values for the 95th percentile male Hybrid III ATD are provided in Table 2. The values provided by Eppinger were used for analysis, as these are the same values used by NHTSA online in Version 10 of the $N_{\rm ij}$ calculator.

Kleinberger recommends using an $N_{\rm ij}$ critical value of 1.4 for all size ATDs, whereas Eppinger's 1999 and 2000 reports and FMVSS 208 all use a critical value of 1. Eppinger (1999) and Kleinberger do provide injury risk curves to relate $N_{\rm ij}$ to the probability of AIS 3 or greater and 5 or greater level neck injuries, whereas no such formulae are provided by Eppinger (2000). Due to the apparent discrepancy in allowable $N_{\rm ij}$ and the fact that $N_{\rm ij}$ values for the CPR-seated, bench-seated and standing occupants were virtually all well below 1.0, no attempt was made to equate $N_{\rm ij}$ to probability of injuries.

3.3.3. Lateral N_{ij} analysis

This effort adopted the lateral $N_{\rm ij}$ analysis method proposed by Soltis [35] to add another comparative parameter to the evaluation of each restraint option. The lateral $N_{\rm ij}$ component parameters for the 50th percentile male, developed for side-facing aircraft seats, are provided by Soltis in Table 3. They offer the same peak and intercept values presented by Eppinger [9] for mid-sized male neck tension and compression. Soltis added a 60 Nm limit for lateral neck bending (Mx).

Recognising all testing discussed in this report used the 95th percentile male, Soltis' lateral bending limit was scaled for the larger occupant. Eppinger [9] presented flexion–extension limits for both the mid-sized and large males. As presented in Table 4, the limits for the large male are approximately 33% greater than those for the mid-sized male. This scaling factor was applied to the lateral neck-bending limit proposed by Soltis, resulting in a lateral bending limit of 80 Nm for the large male. It should be noted that the flexion–extension limits used herein agree with those published by Mertz (2003); however, the authors chose the conservative value of 80 Nm for lateral bending (Soltis-scaled) versus 178 Nm (Mertz 2003) given the limited test data to support the higher limit. The values in the last column of Table 4

Table 2. Intercept values for the 95th percentile Hybrid III male ATD.

Axis	Intercept value [9]
Neck compression (Fz)	7440 N (1673 Ibs)
Neck tension (Fz)	8216 N (1847 Ibs)
Neck flexion (My)	415 Nm (3673 in-lbs)
Neck extension (My)	179 Nm (1584 in-lbs)

Table 3. Soltis [36] proposed values for lateral N_{ij} (50th percentile male).

Axis	Intercept value	Peak limits
Neck compression (Fz)	6160 N (1385 lbs)	4000 N (900 lbs)
Neck tension (Fz)	6806 N (1530 lbs)	4170 N (940 lbs)
Lateral neck bending (Mx)	60 Nm (531 in-lbs)	

were used to calculate a lateral N_{ij} for all tests. In acknowledgement of the scaled neck-bending limit and the limited information related to the use of lateral N_{ij}, a critical value was not used, rather the calculated lateral N_{ij} is presented for comparison purposes only. It is expected that a higher lateral N_{ij} will equate to an increased risk of injury.

3.4. *Thorax acceleration criteria*

IARVs exist for the thorax acceleration, sternal deflection and TTI. While TTI could have been a useful evaluation tool for some occupant test configurations, testing was conducted with Hybrid III ATDs; thus, TTI could not be used. Additionally, the expectation that chest deflection would provide little insight into restraint performance led the team to select acceleration-based criteria for the evaluation of thorax response. Thus, this evaluation used the FMVSS 208 criteria, as influenced by Mertz and Gadd [24] and Stapp [38], which requires the thorax acceleration not to exceed 60 g for intervals whose cumulative duration exceeds 3 ms.

3.5. *Pelvis injury criteria*

Injury assessment of pelvic injuries is also very limited. FMVSS 208 does not include pelvic acceleration as a measured parameter. FMVSS 214 (Side Impact Protection) includes a limit of 130 g on lateral pelvic acceleration but does not place a limit on resultant acceleration. While this limit is applicable to side-facing occupants in frontal and rear crashes and the rear-facing occupant in a lateral impact, there is still a question as to what limits should be used for the pelvic acceleration when there is a substantial fore/aft component. Therefore, for this analysis, the only IARV applied was for lateral acceleration.

3.6. *Lumbar loads*

Published IARVs regarding the lumbar spine address compressive loading only. Recognising the paucity of data available, the authors have used a limit published specifically for loading measured on the 95th percentile Hybrid III ATD of 11,272 N (2534 pounds) [21,33,45,46]. Given the seating configurations, restraints and impact directions tested, no significant lumbar compression loads were expected.

Little data are available addressing the lumbar spine's tolerance to lateral bending. Though it is recognised that IARVs do not exist for distraction and lateral bending, data were collected for all seated ATDs for comparative purposes only. Data were not collected for the standing ATD as it was not equipped to measure such loading.

4. Sled test methodology

4.1. *Restraints evaluated*

A total of five different commercial off-the-shelf restraint systems were tested including a standard lap belt. Of these, only two were specifically built for use in an ambulance. Recognising that the four mobility restraints tested have been modified based on the results of this testing; manufacturers will not be disclosed.

4.1.1. *Lap belt only*

A type 1 lap belt restraint was tested in two reduced-severity frontal tests, in one seating position. For the purpose of this testing, a lap belt is described as having two anchorage points and fits over the upper thighs/hips. Its primary function is to hold the occupant in the vehicle; it does not provide upper body restraint. The lap belt was positioned below the iliac spines of the pelvis. The lap belt system was a traditional belt-based vehicle restraint incorporating an automatic locking retractor.

Table 4. Proposed values for large male lateral N_{ij}.

Axis	Proposed values mid male (Soltis [36])	Mid male (Eppinger [9])	Large male (Eppinger [9])	Proposed values 95th male
Neck compression intercept (Fz)	6160N (1385 tbs)	6160N (1385 Ibs)	7440 N (1673 Ibs)	7440 N (1673 Ibs)
Neck tension intercept (Fz)	6806 N (1530 Ibs)	6806 N (1530 Ibs)	8216 N (1847 Ibs)	8216N (1847 Ibs)
Literal neck bending (Mx)	60 Nm (531in-lb)			80 Nm (706 in-lb)
Peak neck compression (Fz)	4000 N (900 Ibs)	4000 N (900 Ibs)	4830 N (1086 Ibs)	4830 N (1086 Ibs)
Peak neck tension (Fz)	4170 N (940 Ibs)	4170 N (940 Ibs)	5030 N (1131 Ibs)	5030 N (1131 Ibs)
Neck flexion (+My)		310Nm (2744 in-lb)	415 Nm (3673 in-tb)	
Neck extension (−My)		135 Nm (1195 in-lb)	179 Nm (1584in-lb)	

Figure 3. System A, three ambulance mounted tethers and retractors with lap belt (pretest).

Figure 4. System B (pretest), four-tether harness.

4.1.2. System A

System A (Figure 3) used a vest with three tethers attached to the ambulance via retractors. Each tether was attached to the vest with a clip and D-ring. One tether was hooked to a D-ring on the front of the vest (left or right side depending upon the occupant position), a second was hooked to a D-ring located on the upper middle of the back, and a third tether was hooked to a D-ring located on the lower middle of the back. The other end of each tether was attached to the sled buck by its own web-sensing retractor. The chest-linked tether retractor was mounted in front, and to the aft, of the assigned occupant seating position while the other two retractors were mounted in a vertical fashion behind the centreline of the occupant position. The manufacturer's instructions for this restraint required that, in addition to the vest, an occupant wear the available lap belt at each occupant position when they were seated. For each of the seated occupants in this test series, a lap belt (as described in Section 4.1.1) was used in addition to the vest restraint. The standing ATD was restrained with the restraint systems' tethers and retractors only.

4.1.3. System B

System B (Figure 4) used a harness that incorporated four tether/straps. Two of the tethers were attached near the back of each shoulder and the other two were attached near the right and left iliac crest (pelvis). Each tether was attached to its own retractor and all retractors were locked via a centrally mounted vehicle acceleration-sensing device. The four retractors were mounted to the sled buck behind the occupant. The lower retractors were mounted below the

Figure 5. System C (pretest), three-tether harness with web- and vehicle-sensing retractors.

level of the seating surface near the seat bight. Each upper tether strap was routed through a bulkhead-mounted D-ring and then to a retractor above head height. The D-rings were mounted slightly above, and in-line with, the ATD's shoulders.

4.1.4. System C

System C (Figure 5) used a dedicated harness with three tethers attached to the vehicle. One tether was sewn in place to the middle of the upper back of the harness in relation to the occupant and went to a retractor mounted to the outside of the sled buck approximately behind the location of the tether-to-harness attachment. The two lower tethers attached directly to the harness in the area of each iliac crest. The lower retractors were mounted to the vehicle, below the seating surface and at the outside edge of each occupant position. The retractors were off-the-shelf retractors used in aviation applications with both web- and vehicle-sensing systems set to a 5.0 g limit – much higher than the 0.7 g limit required of emergency-locking retractors used in the automotive industry [43,19]. The retractors were also equipped with a manual override capability.

4.1.5. System D

System D (Figure 6) also used a dedicated harness with three tethers attached to the vehicle. The harness and tether-to-vehicle geometry were very similar to that of System C. The most significant difference between Systems C and D was retractor type as the retractors used with System D were web sensing only. The lower retractors also had a manual over-ride feature that would allow an occupant to lock the retractor, if desired. Like System C, the lower retractors were off-the-shelf retractors used in various aviation applications. As such, the web-sensing system was set to a 5.0 g limit.

4.2. ***Data acquisition system***

The data acquisition system (DAS) collected up to 93 channels of data from the test buck and ATDs. Two Hybrid III ATDs representing the 95th percentile male, by seated height and weight, were used at each seating position evaluated. A third 95th percentile Hybrid III ATD, equipped with an articulating hip, was positioned at the forward end of the bench seat in a near-standing position between the chest of the patient on the gurney and the bench seat. This was done to simulate a worst-case, real-world scenario when a worker perceives the need to stand to attend to the patient or access equipment. (Note: The authors do not advocate a worker stand in the patient compartment of a moving ambulance. However, previous research has indicated that this situation frequently occurs and is perceived by many care providers as a job requirement in patient care in a moving ambulance [23,30].) Given comparative, rather than compliance,

Figure 6. System D (pretest), three-tether harness with web-sensing retractors.

testing was the goal, the same ATD always occupied the same seating position for each test to minimise ATD-induced variability.

4.3. *Patient compartment layout*

The layout of the patient compartment sled buck was based upon a small survey of ambulances operating at 12 sites in and around Philadelphia, PA, and Morgantown, WV. As such, the buck does not copy any single manufacturer's design. Rather, it is an amalgam of those found in service. Interior views are provided as Figures 7 and 8. It should be noted, this test did not make an effort to mitigate the effects of surface type on the magnitude of contact loading. Though some benefit could be expected and has been reported in previous research [7], 0.125 aluminum sheet over extruded aluminum tubing was used throughout.

4.4. *Sled test pulse definition*

Sled testing was conducted in a single horizontal plane at three attitudes simulating lateral, rear and frontal impacts. Table 5 provides a review of the sled test programme target pulses while a more detailed description of the origins of each is provided below. Figure 9 provides a graphical depiction of each pulse shape (acceleration versus time) for each impact attitude.

A 2002 review of NHTSA's database of available crash tests failed to reveal any side- or rear-impact tests with vehicles larger than a full-size pickup or van. While it was recognised the gross vehicle weight for these vehicles is half that of a Type I or Type III ambulance, with no better data to use, data from these vehicles were used to approximate the side- and rear-impact crash pulses for the ambulance. The side pulse was adapted from a crash test of a 1987 Chevy C1500 Pick-up, NHTSA Test 2102 [47]. The rear impact crash pulse chosen was adapted from NHTSA Test CP5111 for 1993 Toyota T-100 Pick-up [48].

The high-severity frontal impact testing was designed to produce an acceleration pulse similar to that found during

Table 5. Range of actual test pulse.

Impact attitude	Delta V (km/h)	Acceleration (peak values of g)
Frontal (higher)	49.9–51.8	29.3–3.1.8
Frontal (reduced)	38.0–42.5	23.4–26.4
Rearward	28.2–30.4	22.8–25.8
Lateral	26.5–27.7	19.7–20.5

Figure 7. Interior of patient compartment buck looking in from rear doorway, street side.

an FMVSS 208 frontal barrier test conducted at a nominal 49.5 km/h, but representative of that which might be found when testing a large frame-based vehicle. The best available data found were from a frontal barrier crash test of a Type III ambulance mounted on a 1990 Ford RV chassis. This test was conducted in Canada in 1991 [6]. In addition, moderate–severity frontal impact tests were conducted at approximately 40 km/h delta-V and at acceleration levels of 25 g using the same curve shape as found in the high-severity frontal impact testing.

Figure 8. Interior of patient compartment buck looking in from rear doorway, bench seat and curb side.

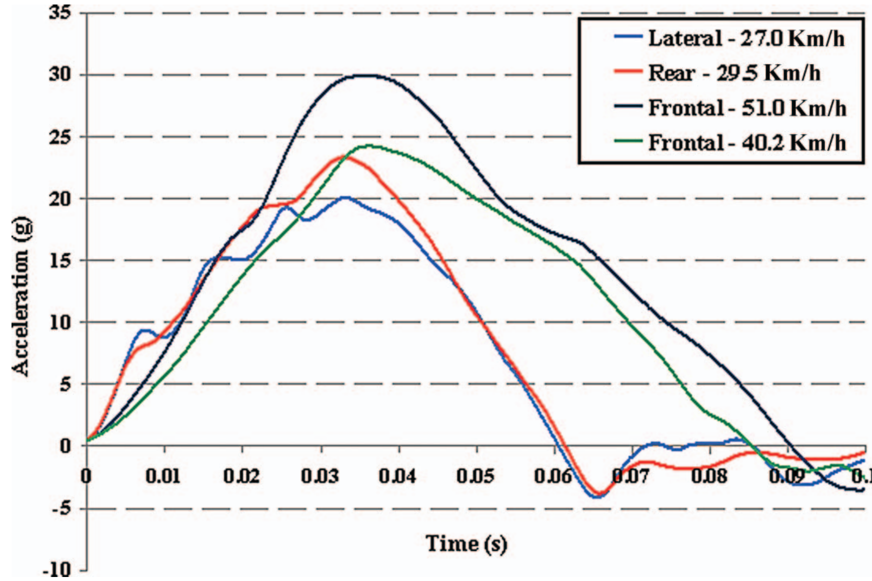


Figure 9. Average sled test pulses.

5. Results: sled test programme by impact attitude and seating position

5.1. Sled test matrix

A detailed summary of the full sled test matrix, including selected measured and calculated ATD response data, is contained in Tables 6–8. Data are sorted by seating location, impact attitude and restraint type.

5.2. Seated, opposite-side impact

A seated ATD was located on the rear seating position of the bench seat on the curb side of the vehicle. This test impact attitude was received as a true frontal impact for this side-facing seating position. Sled test impact velocity values ranged from 26.5 to 27.7 km/h, with accelerations ranging from 19.7 to 21.2 g, for each of the eight test events. All calculated and measured values for this occupant position, for all four mobility restraints systems tested, remained within published and accepted IARVs as delineated in Table 6. No structural failures were noted with any of the four mobility systems tested in either seating position.

5.3. Standing, opposite-side impact

The standing ATD was located near the forward seating position of the bench seat on the curb side of the vehicle, adjacent to and facing the chest of the patient as if preparing to deliver CPR. This test impact attitude was received as a true frontal impact for this side-facing standing position. Sled test impact velocity values ranged from 26.5 to 27.7 km/h, with accelerations ranging from 19.7 to 20.5 g, for each of the eight test events. All calculated and measured values for this occupant position, for all four mobility restraints tested, remained within published and accepted IARVs as delineated in Table 7. However, the knees of the standing ATD contacted the gurney and ATD representing the patient. The standing ATD did not carry femur load cells; therefore, no assessment of the impact severity can be provided, though one would not expect such an impact to result in a life-threatening injury for the restrained worker.

5.4. Seated, near-side impact

An impact to the street side of the ambulance is received as a rear impact for the ATD located in this occupant position (CPR seat). In this orientation, the seating system provided the primary means of restraint during the initial impact while the restraint systems supported the ATD during rebound. There were no significant issues or failures noted with any of the four mobility restraints at this occupant position. However, during each test the ATD's head contacted the wall of the test buck behind the seating position.

Two parameters exceeded the accepted IARVs during this test series: upper neck extension moment and peak lumbar compressive force. (A complete summary of measured and calculated values can be found in Table 8.) As seen in Figure 10, the ATD's head hyperextended over the low seatback, allowing head contact. Despite this contact, head acceleration and HIC values remained well below IARVs for all lateral impact tests. This is attributed to the lack of significant supporting structure directly behind the head of the ATD and the relatively short head travel distance

Table 6(a). Bench-seated ATD, sled test data.

Parameter definition	Limit	Attitude / Acceleration / Delta V (km/h) / Test ID / Restraint															
		Attitude															
		Acceleration	Lat	Lat	Lat	Lat	Lat	Lat	Lat	Lat	Rear	Rear	Rear	Rear	Rear	Rear	Rear	Rear
		Delta V (km/h)	21.2	20.2	20.5	19.8	19.8	19.7	19.9	20.1	25.8	22.8	23.0	23.3	22.9	23.1	23.2	23.0
			26.6	26.7	26.9	27.4	27.7	27.2	27.0	27.4	30.4	28.2	29.3	29.6	29.1	29.9	29.3	29.5
		Test ID	5013	5014	5015	5016	5017	5018	5019	5020	5021	5022	5023	5024	5025	5026	5027	5028
Parameter definition	Limit	Restraint	A	A	A	B	B	B	C	D	A	A	A	B	B	B	D	C
Res head acceleration (G with 3 ms clip)	80		24	32	27	39	45	54	38	21	63	58	76	102	97	51	63	95
Head acceleration – HIC15	700		42	83	54	193	526	514	121	30	368	164	273	728	1165	212	232	559
Head acceleration – HIC36	1000		60	164	97	193	526	514	121	59	380	301	350	728	1165	218	248	559
Res chest acceleration (G with 3 ms clip)	60		27	23	26	27	33	34	25	20	49	62	64	78	83	73	70	61
Peak lateral pelvis acceleration (G)	130		ND	ND	ND	ND	ND	ND	ND	13	68	86	72	72	99	75	34	103
95th male peak upper neck tension [+Fz] (N)	5030**		1417	1359	1299	2007	2042	1971	1404	1188	2432	2186	2317	2435	2453	2468	3185	2489
95th male peak upper neck comp [−Fz] (N)	4830**		337	275	257	670	824	1407	537	386	725	611	924	1970	2584	465	1143	1128
Peak upper neck lateral bending [+Mx] (Nm)	80**		14	12	9	12	12	20	14	20	104	94	99	95	96	113	101	82
Peak upper neck lateral bending [−Mx] (N m)	80**		24	15	12	25	8	16	19	23	31	74	49	70	78	55	62	40
Peak upper neck flexion [+My] (Nm)	415		61	98	71	46	58	61	57	34	15	21	27	41	56	36	67	29
Peak upper neck flexion [−My] (Nm)	179		36	53	49	30	52	60	17	22	18	51	38	40	49	24	26	27
Maximum standard N_{ij}	1.00		0.30	0.42	0.39	0.30	0.40	0.33	0.19	0.22	0.33	0.39	0.29	0.44	0.62	0.32	0.41	0.33
Peak lumbar comp [−Fz] (N)	11,272**		910	1296	1166	5050	4766	2555	3696	2262	1897	1249	1243	1628	1637	1312	1036	1068
Peak lumbar tension [+Fz] (N)	NI		5382	4703	4639	1187	1555	1896	1020	2300	1159	2071	3515	2837	2660	1757	1964	2615
Peak lumbar flexion [+My] (Nm)	NI		910	714	685	75	196	274	49	244	61	71	56	31	111	42	64	67
Max lateral N_{ij}	NI		0.34	0.34	0.22	0.36	0.34	0.32	0.26	0.34	1.37	1.29	1.37	1.26	1.46	1.46	1.47	1.46

Note: NI, no accepted injury assessment reference value (IARV); ND, no data available; **, research value.

Table 6(*b*). Bench-seated ATD, sled test data.

Parameter definition	Limit														
		Attitude	High	High	High	High	High	High	High	Low	Low	Low	Low	Low	Low
		Acceleration	front	front	front	front	front	front	front	front	front	front	front	front	front
		Delta V (km/h)	31.8	29.9	30.7	29.3	29.3	30.5	29.3	24.8	26.6	24.6	21.9	25.1	23.4
			51.8	50.9	50.7	50.7	49.9	51.8	50.9	40.2	42.5	39.8	41.8	39.3	38.0
		Test ID	5102	5103	5104	5105	5106	5107	5108	5109	5110	5111	5112	5113	5114
		Restraint	A	A	A	D	C	B	B	B	D	A	C	LBO	LBO
Res head acceleration (G with 3 ms clip)	80		58	73	62	36	48	69	79	41	49	58	40	69	86
Head acceleration – HIC15	700		368	821	363	498	3519	511	298	68	1339	90	203	343	350
Head acceleration – HIC36	1000		764	821	599	498	3519	511	507	89	1339	486	203	553	1108
Res chest acceleration (G with 3 ms clip)	60		44	44	48	ND	46	45	40	30	29	33	34	22	24
Peak lateral pelvis acceleration (G)	130		67	96	106	180	60	53	51	48	88	43	42	53	48
95th male peak upper neck tension [+Fz] (N)	5030**		3515	4098	3116	1403	2338	2125	2861	1277	2665	2986	1945	2598	2975
95th male peak upper neck comp [−Fz] (N)	4830**		526	460	288	3088	1058	1808	2373	647	1936	820	2058	1610	1439
Peak upper neck lateral bending [+Mx] (Nm)	80**		40	30	33	31	46	25	73	87	42	28	12	39	35
Peak upper neck lateral bending [−Mx] (N m	80**		119	112	92	46	131	115	116	22	77	106	96	43	80
Peak upper neck flexion [+My] (Nm)	415		47	18	53	134	80	17	28	22	43	32	9	13	10
Peak upper neck flexion [−My] (Nm)	179		38	65	47	66	50	72	106	26	42	38	30	39	99
Maximum standard N_{ij}	1.00		0.58	0.73	0.56	0.65	0.37	0.39	0.81	0.22	0.45	0.49	0.39	0.49	0.66
Peak lumbar comp [−Fz] (N)	11,272**		879	877	924	1931	1319	636	800	1222	2562	964	3780	792	933
Peak lumbar tension [+Fz] (N)	NI		6969	6483	6261	4123	5490	6577	4990	3243	3019	3780	1005	7161	3164
Peak lumbar flexion [+My] (Nm)	NI		548	402	467	277	251	280	433	96	140	326	45	472	401
Max lateral N_{ij}	NI		1.73	1.59	1.26	0.62	1.73	1.54	1.57	1.12	0.98	1.47	1.28	0.73	1.19

Note: NI, no accepted injury assessment reference value (IARV); ND, no data available; **, research value.

Table 7(*a*). Bench standing ATD, sled test data.

		Attitude	Lat	Lat	Lat	Lat	Lat	Lat	Lat	Lat	Rear	Rear	Rear	Rear	Rear	Rear	Rear	Rear
		Acceleration	21.2	20.2	20.5	19.8	19.8	19.7	19.9	20.1	25.8	22.8	23.0	23.3	22.9	23.1	23.2	23.0
		Delta V (km/h)	26.6	26.6	26.7	26.9	27.4	27.7	27.2	27.0	27.4	30.4	28.2	29.3	29.6	29.1	29.9	29.3
		Test ID	5013	5014	5015	5016	5017	5018	5019	5020	5021	5022	5023	5024	5025	5026	5027	5028
Parameter definition	Limit	Restraint	A	A	A	B	B	B	C	D	A	A	A	B	B	B	D	C
Res head acceleration (G with 3 ms clip)	80		24	23	26	22	20	23	21	19	24	27	24	13	12	12	21	19
Head acceleration – HIC15	700		38	34	45	30	27	35	29	22	35	27	24	9	3	6	23	22
Head acceleration – HIC36	1000		69	66	78	49	51	54	45	29	55	33	45	15	17	11	38	45
Res chest acceleration (G with 3 ms clip)	60		19	17	19	19	20	34	23	16	17	203*	18	26	ND	ND	ND	13
Peak lateral pelvis acceleration (G)	130		80	30	55	36	11	21	24	74	18	15	25	26	38	19	20	16
95th male peak upper neck tension [+Fz] (N)	5030**		1716	1649	1299	1147	1151	1247	1226	1408	396	1020	774	623	451	602	804	851
95th male peak upper neck comp [−Fz] (N)	4830**		235	257	257	465	616	200	316	420	343	267	348	560	389	732	318	288
Peak upper neck lateral bending [+Mx] (Nm)	80**		21	19	26	15	16	16	22	26	20	30	20	33	34	33	55	43
Peak upper neck lateral bending [−Mx] (N m	80**		11	11	12	17	20	17	18	27	37	27	23	28	11	19	14	13
Peak upper neck flexion [+My] (Nm)	415		33	34	38	32	32	43	31	28	33	26	31	17	31	23	21	30
Peak upper neck flexion [−My] (Nm)	179		25	34	30	20	21	23	23	27	36	46	36	18	14	21	27	23
Maximum standard N_{ij}	1.00		0.27	0.28	0.24	0.15	0.19	0.17	0.17	0.21	0.28	0.36	0.25	0.14	0.12	0.16	0.21	0.18
Peak lumbar comp [−Fz] (N)	11,272**		ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND
Peak lumbar tension [+Fz] (N)	NI		ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND
Peak lumbar flexion [+My] (Nm)	NI		ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND
Max lateral N_{ij}	NI		0.28	0.31	0.38	0.27	0.28	0.25	0.29	0.38	0.5	0.41	0.33	0.45	0.45	0.45	0.72	0.56

Note: NI, no accepted injury assessment reference value (IARV); ND, no data available; **, research value.

Table 7(*b*). Bench standing ATD, sled test data.

Parameter definition	Limit	Attitude / Acceleration / Delta V (km/h) / Test ID / Restraint	High front	High front	High front	High front	High front	High front	High front	Low front	Low front	Low front	Low front	Low front	Low front
		Acceleration	31.8	29.9	30.7	29.3	29.3	30.5	29.3	24.8	26.6	24.6	21.9	25.1	23.4
		Delta V (km/h)	51.8	50.9	50.7	50.7	49.9	51.8	50.9	40.2	42.5	39.8	41.8	39.3	38.0
		Test ID	5102	5103	5104	5105	5106	5107	5108	5109	5110	5111	5112	5113	5114
		Restraint	A	A	A	D	C	B	B	B	D	A	C	None	None
Res head acceleration (G with 3 ms clip)	80		82	60	55	118	147	69	112	35	90	41	124		
Head acceleration – HIC15	700		1065	2237	615	3782	5139	1272	1075	71	2467	155	1303		
Head acceleration – HIC36	1000		1065	2237	615	3782	5139	1272	1001	111	2467	288	1303		
Res chest acceleration (G with 3 ms clip)	60		105	93	105	67	93	59	85	33	74	25	43		
Peak lateral pelvis acceleration (G)	130		220	207	191	ND	63	227	109	24	ND	46	24		
95th male peak upper neck tension [+Fz] (N)	5030**		3475	3278	2120	9225	4304	2865	5147	1945	3802	1869	2349		
95th male peak upper neck comp [−Fz] (N)	4830**		624	379	1450	3634	2801	626	1377	1287	2904	445	3032		
Peak upper neck lateral bending [+Mx] (Nm)	80**		110	60	32	265	250	25	45	72	223	34	155		
Peak upper neck lateral bending [−Mx] (N m	80**		51	98	64	72	88	121	185	40	36	79	52		
Peak upper neck flexion [+My] (Nm)	415		60	53	45	62	102	77	101	40	36	53	48		
Peak upper neck flexion [−My] (Nm)	179		41	48	39	79	126	59	90	29	42	39	41		
Maximum standard N_{ij}	1.00		0.47	0.45	0.33	1.14	1.07	0.52	0.84	0.31	0.53	0.34	0.59		
Peak lumbar comp [−Fz] (N)	11,272**		ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND		
Peak lumbar tension [+Fz] (N)	NI		ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND		
Peak lumbar flexion [+My] (Nm)	NI		ND	ND	ND	ND	ND	ND	ND	ND	ND	ND	ND		
Max lateral N_{ij}	NI		1.57	1.43	0.93	3.75	3.27	1.55	2.37	0.96	2.88	1.1	2.21		

Note: NI, no accepted injury assessment reference value (IARV); ND, no data available; **, research value.

Table 8(*a*). CPR-seated ATD, sled test data.

Parameter definition	Limit	Attitude / Acceleration Delta V (km/h) / Test ID / Restraint															
		Attitude	Lat	Lat	Lat	Lat	Lat	Lat	Lat	Lat	Rear	Rear	Rear	Rear	Rear	Rear	Rear	Rear
		Acceleration	21.2	20.2	20.5	19.8	19.8	19.7	19.9	20.1	25.8	22.8	23.0	23.3	22.9	23.1	23.2	23.0
		Delta V (km/h)	27.5	27.7	27.8	28.3	28.7	28.2	28.0	28.3	31.5	29.2	30.3	30.7	30.2	31.0	30.3	30.5
		Test ID	5013	5014	5015	5016	5017	5018	5019	5020	5021	5022	5023	5024	5025	5026	5027	5028
		Restraint	A	A	A	B	B	B	C	D	A	A	A	B	B	B	D	C
Res head acceleration (G with 3 ms clip)	80		40	44	36	41	33	34	68	35	58	55	51	50	59	42	67	57
Head acceleration – HIC15	700		103	152	70	108	51	89	207	75	175	317	400	137	194	97	276	287
Head acceleration – HIC36	1000		147	222	114	142	100	132	344	118	212	317	400	166	207	142	276	287
Res chest acceleration (G with 3 ms clip)	60		53	58	51	60	52	59	60	57	86	97	86	77	89	65	109	92
Peak lateral pelvis acceleration (G)	130		9	5	9	6	7	7	7	6	90	71	89	109	123	109	76	128
95th male peak upper neck tension [+Fz] (N)	5030**		1986	2291	1832	2129	1641	1769	2539	1842	2658	2740	2577	2287	3053	2054	3376	2977
95th male peak upper neck comp [−Fz] (N)	4830**		1332	1472	2040	1790	2043	1705	1076	2061	1143	625	787	297	429	272	613	324
Peak upper neck lateral bending [+Mx] (Nm)	80**		15	9	7	13	7	5	11	4	114	133	112	100	121	77	143	123
Peak upper neck lateral bending [−Mx] (N m)	80**		5	9	4	11	9	7	7	6	135	.40	130	108	132	92	144	135
Peak upper neck flexion [+My] (Nm)	415		99	98	96	73	88	79	94	75	30	28	26	20	23	24	24	23
Peak upper neck flexion [−My] (Nm)	179		122	105	93	102	91	95	91	91	30	25	24	23	33	17	32	31
Maximum standard N_{ij}	1.00		0.76	0.63	0.58	0.72	0.58	0.65	0.65	0.62	0.43	0.39	0.38	0.36	0.43	0.29	0.43	0.45
Peak lumbar comp [−Fz] (N)	11,272**		4909	8370	11,905	12,857	12,244	12,609	13,636	12,948	1463	565	1210	1232	1439	1247	1872	1518
Peak lumbar tension [+Fz] (N)	NI		985	1214	3655	2739	486	740	2730	2187	3121	1582	3197	3260	3877	2626	4043	4215
Peak lumbar flexion [+My] (Nm)	NI		333	340	293	357	319	406	396	344	71	424	66	72	61	106	45	51
Max lateral N_{ij}	NI		0.28	0.37	0.31	0.40	0.30	0.29	0.35	0.30	1.34	1.99	1.71	1.53	1.85	1.25	2.24	1.96

Note: NI, no accepted injury assessment reference value (IARV); ND, no data available; **, research value.

Table 8(*b*). CPR-seated ATD, sled test data.

Parameter definition	Limit														
		Attitude	High	High	High	High	High	High	High	Low	Low	Low	Low	Low	Low
		Acceleration	front	front	front	front	front	front	front	front	front	front	front	front	front
		Delta V (km/h)	31.8	29.9	30.7	29.3	29.3	30.5	29.3	24.8	26.6	24.6	21.9	25.1	23.4
			51.8	50.9	50.7	50.7	49.9	51.8	50.9	40.2	42.5	39.8	41.8	39.3	38.0
		Test ID	5102	5103	5104	5105	5106	5107	5108	5109	5110	5111	5112	5113	5114
		Restraint	A	A	A	D	C	B	B	None	LBO	LBO	None	None	None
Res head acceleration (G with 3 ms clip)	80		87	118	113	94	127	88	ND		91	61			
Head acceleration – HIC15	700		2184	4392	4463	6366	2354	2088	ND		1855	2722			
Head acceleration – HIC36	1000		2184	4392	4463	4463	2354	2088	ND		1855	2722			
Res chest acceleration (G with 3 ms clip)	60		68	51	ND	53	46	50	66		37	55			
Peak lateral pelvis acceleration (G)	130		176	135	150	129	110	126	177		89	148			
95th male peak upper neck tension [+Fz] (N)	5030**		4467	3917	3917	4373	2343	2206	ND		2363	2472			
95th male peak upper neck comp [−Fz] (N)	4830**		4549	3597	3597	4077	4195	4156	ND		4359	2832			
Peak upper neck lateral bending [+Mx] (Nm)	80**		141	83	82	163	61	135	ND		62	86			
Peak upper neck lateral bending [−Mx] (N m)	80**		163	143	150	137	106	107	ND		117	159			
Peak upper neck flexion [+My] (Nm)	415		22	24	19	29	34	29	ND		17	22			
Peak upper neck flexion [−My] (Nm)	179		40	42	45	45	41	52	ND		49	33			
Maximum standard N_{ij}	1.00		0.65	0.56	0.52	0.59	0.64	0.62	ND		0.66	0.57			
peak lumbar comp [−Fz] (N)	11,272**		1534	529	494	729	602	1483	1127		225	771			
Peak lumbar tension [+Fz] (N)	NI		4804	3829	4352	3946	4148	5579	4256		4108	3443			
Peak lumbar flexion [+My] (Nm)	NI		39	106	55	31	57	88	18		59	13			
Max lateral N_{ij}	NI		2.17	1.98	2	2.07	1.48	1.71	1.53		1.71	2.24			

Note: NI, no accepted injury assessment reference value (IARV); ND, no data available; **, research value.

Figure 10. CPR seated ATD viewed from the right side of the patient compartment, side impact test with head-to-wall contact.

before impact. During the hyperextension of the neck, upper neck extension moment (My) values of 91–122 Nm were recorded, exceeding the IARV of 57 Nm.

Additionally, the peak lumbar compressive force, which ranged from 4909 to 13,636 N, exceeded the IARV of 11,272 N in six of the eight tests. A review of the video shows that this may have resulted from a gap between the horizontal seat pad and the bulkhead-mounted seatback. This gap appeared to allow the ATD's buttock to 'wedge' into this gap.

Finally, there was some concern that restraint System A could induce increased flexion moments as the occupant moved rearward due to the front tether strap. However, this was not seen for the occupant located in the CPR seat during any of the three tests of this restraint.

5.5. *Seated, rear impact*

Eight rear impact tests were conducted at delta-V values ranging from 29.2 to 31.5 km/h. A rear impact to the vehicle is a side impact relative to the two seated positions studied: the CPR seat and the rear bench seat. Thus, data for a total of 16 test events were recorded. A full, unpadded, seat-to-ceiling wall, representing storage cabinets, was located immediately aft of both seating positions. In 15 of the 16 events, the chest acceleration IARV of 60 g was exceeded with values ranging from 49 to 83 g for the bench seat and 65 to 109 g for the CPR seat. The higher values recorded in the CPR seat were attributed to the greater travel distance found from ATD to rear wall in this seating position. Additionally, head accelerations for two of three tests of System B, on the bench seat, had measured values above the IARV of 80 g. A review of video data and restraint systems post-test did not identify anything extraordinary with these tests in comparison to the other 13 of this series. Coincidentally, the recorded head acceleration value for the third test of System B was the lowest of the eight tests performed at this seating location. The variability in these data is attributed to small differences in the initial ATD position which affected head impact location.

5.6. *Standing, rear impact*

A rear impact to the vehicle is a side impact relative to the standing ATD position. No restraint system failures were noted in any of the eight tests. All four mobility restraints performed admirably in these tests with all but one measured and/or calculated value falling well below accepted IARVs for each test. The lone exception was the resultant chest acceleration for the second of the three tests conducted with System A. Tests 1 and 3 recorded values of 17 and 18 g, respectively, while test 2 produced a value of 213 g. However, the video does not show any abnormal or substantial impact and appears similar to the other tests. The data channel that led to this high level, chest x, was only recording values in the ± 8 g range in the prior test (test 5021) and recorded no data in the next test (test 5027). It was later replaced. The x direction is also a direction from which high accelerations would not normally be expected with this impact input. Therefore, the data are considered suspect but are reported for completeness.

5.7. *Seated, high-severity frontal impact*

Seven high-severity frontal impact tests were conducted at delta-V values ranging from 49.9 to 51.8 km/h and Gx ranging from 29.3 to 31.8, with a mean of 30.1. A frontal impact to the vehicle is a side impact relative to the ATDs seated on the bench or CPR seat. In each event, and in both seating positions, damage to mobility restraints was found, generally in the form of webbing or stitching tears.

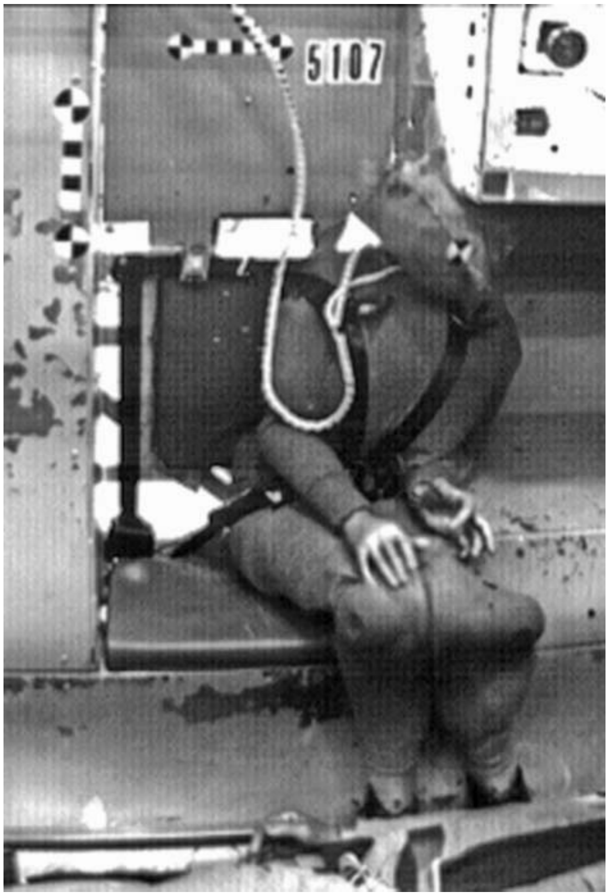


Figure 11. CPR ATD viewed from the right side of the patient compartment, frontal impact test with head-to-cabinet impact.

As illustrated in Figure 11, ATDs seated on the CPR seat were also exposed to a secondary impact hazard with an overhead cabinet at head height just forward of the seating position. In each event, the ATD impacted the cabinet prior to completely loading the restraint system. This resulted in extremely high HIC15 values ranging from 2088 to 6366. However, the bench-seated occupant fared much better with HIC15 values ranging from 298 to 821 for six of the seven tests. The lone exception was test 5106 using System C, with a value of 3519. During this test, the restraint system experienced tearing of the harness and a failure of the retractor mounting bolts which allowed the head of the ATD to strike the curb-side wall.

A review of measured and calculated data for the remainder of the parameters for each of the seven tests, for both seating positions, shows all remained below the established IARVs with the exception of those values calculated as a part of the proposed lateral N_{ij} analysis. For these seating positions and test conditions, lateral N_{ij} ranged from 0.62 to 1.73 for the bench seat and 1.48 to 2.17 for the CPR-seated occupant with head impact.

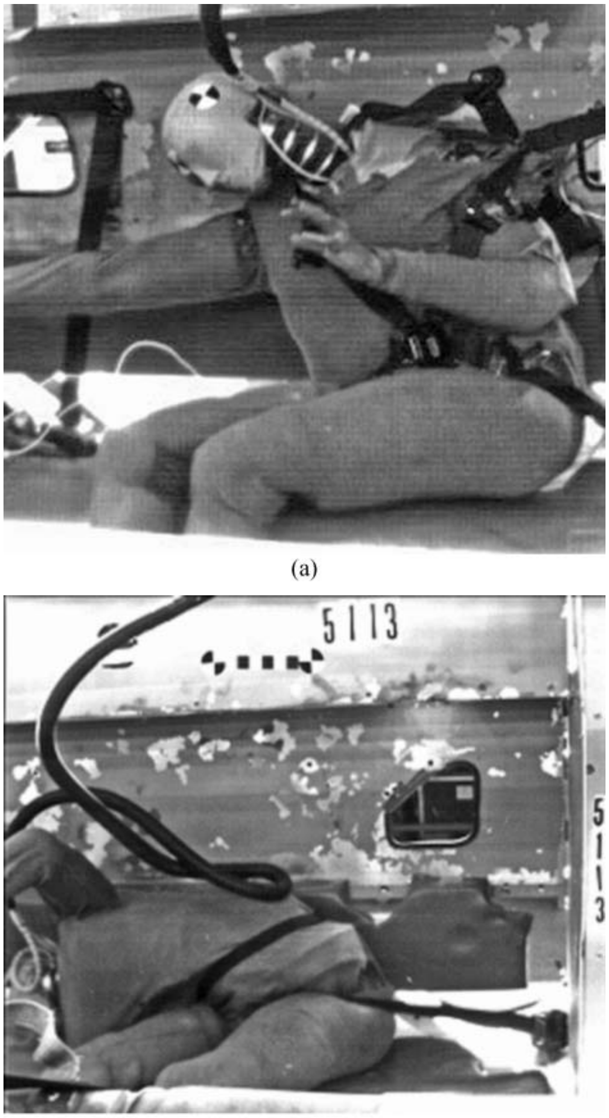
5.8. *Standing, high-severity frontal impact*

During the high-severity frontal impact tests, the restraint systems experienced several forms of hardware failure. These system failures resulted in some loss of restraint, which generally increased ATD excursions, and led to impact with the ambulance interior structure. Due to these hardware failures, little was learned from these tests regarding the crash performance of the restraint systems. However, it is clear the restraints, even with noted structural failures, attenuated a significant amount of crash energy, allowing many measured or calculated values for some of the systems to fall below the identified IARVs. Additionally, the hardware failures did identify areas for improvement for each manufacturer to consider in future designs.

5.9. *Seated, moderate–severity frontal impact*

Six moderate–severity frontal impact tests were conducted at delta-V values ranging from 38.0 to 42.5 km/h and Gx ranging from 21.9 to 26.6, with a mean of 24.4. Of these, two tests were run with ATDs outfitted with lap belts only on both the bench and CPR seats. For those on the CPR seat, excessive head strikes occurred with the cabinet just forward of the seating position, resulting in HIC15 values of 1855 and 2722, far in excess of the IARV of 700.

Data from the bench-seated ATD offered considerably more useful data as four tests were run with mobility restraints (one for each mobility restraint type) and two with lap belts only. Of these, both lap belts and three of the four mobility restraints performed as expected without significant structural failure. The lone failure resulted in head contact with the interior wall, generating an HIC15 of 1339. Measured and calculated values for the remaining three mobility restraints were found to be lower than the lap belt for head acceleration, HIC15 and peak lateral pelvis acceleration. Likewise, measured values for lumbar tension and lateral flexion, when wearing a lap belt only, were generally double those measured on ATDs wearing any of the four mobility restraint systems. As expected, with the torso restrained, measured values for the neck (peak upper neck tension, compression, lateral flexion and lateral extension) trended higher for the mobility restraints when compared to the lap-belt-only tests; however, all were well within currently published limits. Figure 12 illustrates the differences in kinematics found between these two restraint options. Finally, in an effort to better understand and compare the effects of lateral bending on the ATDs using these distinctly different options, a lateral N_{ij} calculation was performed. Again, as a result of upper torso restraint, the mean values for the proposed lateral N_{ij} calculation were 26% higher for the mobility restraints as compared to the lap-belt-only condition.



(a)

(b)

Figure 12. Comparison of kinematic response of ATDs in mobility restraint and a lap belt only, during moderate–severity frontal impact sled test events at maximum manikin excursion.

5.10. Standing, moderate–severity frontal impact

During attenuation of the moderate–severity frontal impacts, the mobility restraints remained essentially intact despite the presence of minor webbing tears on some of the harnesses. As a result, two of the four mobility restraints (Systems A and B) prevented the standing occupant from impacting the forward bulkhead of the test buck, with all measured and calculated ATD parameters falling below currently accepted IARVs. Though Systems C and D each experienced only minor webbing tears, they did not prevent contact with the forward bulkhead, resulting in high head acceleration, HIC15 and lateral N_{ij} values. It has been hypothesised that the higher retractor lock settings coupled with the single upper back tether may have contributed to increased ATD lateral excursions.

6. Crash test methodology

Four full ambulance crash tests were performed: three frontal and one side impact. Specific details regarding individual tests, vehicle attributes, and acquired vehicle chassis and patient compartment accelerations were reported in an earlier publication [3]. Limited details are presented here.

6.1. Anthropometric test device

The crash test programme used the same Hybrid III ATDs representing the 95th percentile male, by seated height and weight, placed in the same seating locations, as described in the sled test programme.

6.2. Restraints evaluated

The crash test programme evaluated only three of the five systems described in the sled test programme: the standard lap belt, System A and System B. When an ATD was seated and wearing System A, a lap belt was added as per the manufacturer's instructions.

6.3. Crash test programme DAS

The data acquisition system used in the crash test programme largely mirrored that of the sled test programme. Sampling rates and ranges for each measured parameter were identical.

6.4. Test vehicle description

Three of the four ambulances tested were retrofitted with additional structure to support one or both of the mobility restraint systems to be tested. Detailed descriptions of each test vehicle used in this effort are provided in Current et al. [3].

7. Results: crash testing by impact attitude and seating position

7.1. Standing, opposite-side impact

A single side-impact test was conducted. In this test, a 3351 kg 1984 GMC Sierra C2500 crew cab dump truck 'bullet vehicle' struck the street side of the target ambulance at approximately 64 km/h, generating a lateral delta-V of 25.8 km/h for the ambulance. The ambulance was equipped with two ATDs, standing side by side between the bench seat and gurney, one wearing the System A restraint while the other wore a System B restraint. Both mobility restraints remained largely intact through the test event and prevented each ATD from contacting interior wall and cabinet surfaces. A review of the video revealed that both ATDs experienced lower leg contact with the gurney structure. However, the authors are unable to comment on the likelihood of lower extremity injury as neither ATD was instrumented to collect femur loading. As reported in Table 9, all measured and calculated values for the head, neck, torso and pelvis were far below currently published IARVs during this test event for both standing ATDs. This indicates a high probability of occupant crash survivability, even when standing, if wearing either of these mobility restraint systems during a left-side impact.

7.2. Seated, frontal impact

Three full vehicle frontal impact tests were conducted with a target barrier approach speed of 48 km/h. Results from this testing produced delta-V values ranging from 49.2 to 52.3 km/h; values very much in line with the higher severity frontal impact sled testing. However, given that the vehicles were free to attenuate loading in both the longitudinal and vertical planes, measured values for Gx ranged from 23.6 to 28.2, with a mean of 26.4 (Figure 13). This more closely mirrored the mean Gx values measured during the moderate–severity frontal impact sled testing. As a result, measured and calculated test results for ATDs wearing Systems A, B and the lap-belt-only were very much in line with those found during the moderate–severity frontal impact sled testing. A review of data in Table 9 shows mobility restraint Systems A and B produced head acceleration, HIC, lateral pelvis, upper neck tension, flexion and extension values, as well as lumbar tension below those measured on the same ATD when wearing the lap belt only. In virtually all cases, the measured and calculated values for Systems A and B fell well below IARVs as well. The exceptions were head acceleration (90 g) and HIC36 (1001) and the proposed lateral N_{ij} (1.02) for System A. However, when compared to values for the lap-belt-only case, where head acceleration (107 g), HIC15 (1810) and HIC36 (2146) far exceeded accepted IARVs, System A could be judged to be a considerable improvement.

7.3. Standing, frontal impact

A standing ATD was tested wearing both Systems A and B coincident with the seated ATD testing of same. During the third frontal test, an unrestrained ATD was seated at the forward end of the bench. A full floor-to-ceiling web system was installed adjacent to the forward end of the bench.

During these tests, Systems A and B were able to restrain the ATD without structural failure and in a manner that prevented the ATD from striking the forward bulkhead. As a result, all measured and calculated parameters for Systems A and B fell below published IARVs, with the exception of the proposed calculation for lateral N_{ij} for System A (1.85).

Finally, testing of the unrestrained but seated ATD generated several findings. First, the web system alone was not able to restrain the ATD and prevent it from striking the forward bulkhead. Second, this resulted in head acceleration (136), HIC15 (1555), and HIC36 (1555) values that exceeded accepted IARVs. Third, these values exceeded those measured and calculated for Systems A and B, even though the ATDs started in a standing position when wearing Systems A and B.

8. Limitations

The ATDs used in this test programme were not specifically designed for side-impact testing. However, the Hybrid III was considered the most biofidelic ATD available to meet the range of testing performed. Though testing was conducted in 2003, the test community still lacks a qualified and calibrated biofidelic neck suitable for far-side lateral impact testing. Therefore, the ATD response data acquired from this testing when the acceleration vector was oriented laterally with respect to the ATD, while useful for comparative purposes, should not be used in an attempt to predict injury. The NHTSA-approved neck injury criterion, N_{ij}, developed to assess the potential for cervical injury, considers flexion and extension combined with tension and compression and is appropriate to be used when the crash acceleration vector is frontal or rear relative to the occupant. While the authors of this paper and others have suggested and used a lateral 'N_{ij} like' parameter in data analysis, a similar validated criterion that includes lateral bending moments is needed to fully assess this restraint condition.

9. Strengths and opportunities for improvement

While each of the systems clearly offered benefits over the lap belt, the testing did reveal some noticeable differences, as well as opportunities for improvements in performance.

9.1. Strengths

Each of the mobility restraint systems provided improvements in upper body restraint, thus reducing head excursion. This will be extremely important as the industry continues to look for opportunities to relocate equipment closer to the worker to reduce the need to stand. Also, each of the mobility restraints offered the opportunity for varying amounts of protection when not seated against the seat back, while the lap belt is rendered useless in these positions. When reviewing the results of specific restraint systems, the use of a chest tether on System A in addition to those on the shoulders reduced the likelihood of impact with interior surfaces regardless of impact direction or ATD positioning. Likewise, the full vest provided better fit and allowed the

Table 9. Crash test data.

Parameter definition	Limit											
		Seat	Bench seat	Bench seat	CPR seat	Bench seat	Bench std	CPR seat	Bench seat	Bench std	Bench std	Bench std
		Attitude	Front	Front	Front	Front	Front	Front	Front	Front	Side	Side
		Acceleration	23.6	23.6	23.6	27.5	27.5	27.5	28.2	28.2	19.7	19.7
		Delta V (km/h)	50.4	50.4	50.4	49.2	49.2	49.2	52.3	52.3	24.9	24.9
		Test ID	Test 1	Test 1	Test 1	Test 2	Test 2	Test 2	Test 3	Test 3	Test 4	Test 4
		Restraint	B	B	B	A	A	A	Lap belt	No belt	A	B
Res head acceleration (G with 3 ms clip)	80		48	53	31	90	50	67	107	136	21	15
Head acceleration – HIC15	700		206	193	77	472	258	483	1810	1555	26	4
Head acceleration – HIC36	1000		337	279	163	1001	491	604	2146	1555	41	6
Res chest acceleration (G with 3 ms clip)	60		32	43	47	30	31	28	22	53	18	22
Peak lateral pelvis acceleration (G)	130		30	40	55	33	69	68	40	58	19	38
95th male peak upper neck tension [+Fz] (N)	5030**		2145	1101	984	3034	2602	2792	3935	567	787	492
95th male peak upper neck comp [−Fz] (N)	4830**		126	847	38	998	75	669	115	1510	1034	1415
Peak upper neck lateral bending [+Mx] (Nm)	80**		13	16	54	25	33	76	51	73	8	10
Peak upper neck lateral bending [−Mx] (N m)	80**		59	75	12	70	136	67	31	31	10	7
Peak upper neck flexion [+My] (Nm)	415		12	71	18	8	5	26	23	42	14	19
Peak upper neck flexion [−My] (Nm)	179		48	26	24	38	42	41	69	35	29	55
Maximum standard N_{ij}	1.00		0.35	0.25	0.27	0.51	0.39	0.48	0.63	0.22	0.27	0.42
peak lumbar comp [−Fz] (N)	11,272**		ND	ND	1254	ND	ND	1670	1255	ND	861	3579
Peak lumbar tension [+Fz] (N)	NI		7580	ND	2729	ND	ND	2661	10,840	ND	1397	292
Peak lumbar flexion [+My] (Nm)	NI		59	ND	41	83	ND	88	188	ND	55	58
Max lateral N_{ij}	NI		0.85	0.95	1.01	1.02	1.85	1.33	0.71	0.95	0.17	0.24

Note: NI, no accepted injury assessment reference value (IARV); ND, no data available; **, research value.

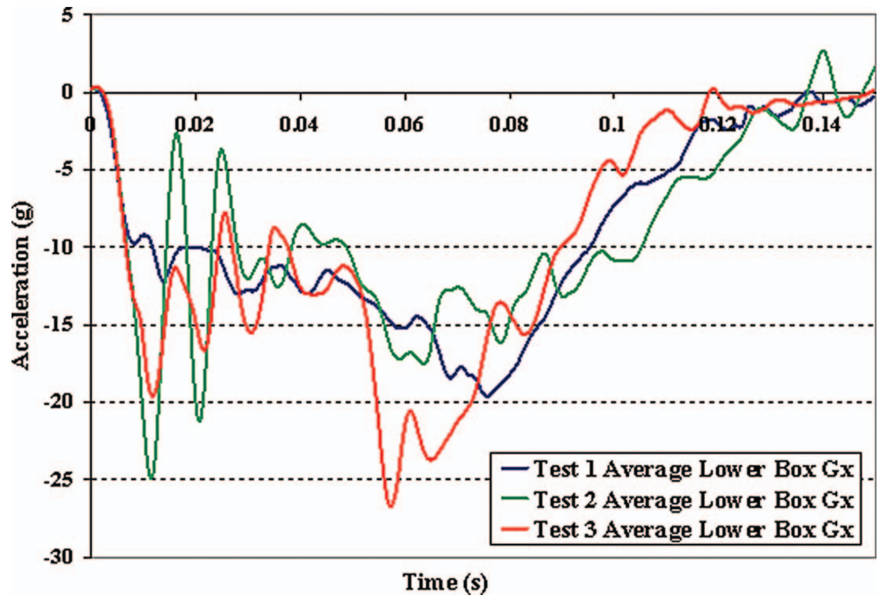


Figure 13. Averaged lower box frontal crash pulses in longitudinal axis.

restraint to remain better coupled with the ATD during ride down. The use of both shoulder and hip retractors was a plus on Systems B–D. These systems provided better distribution of loading across the bony structure of the hips compared to System A.

9.2. *Opportunities for improvement*

System A's reliance on the use of the lap belt while seated worked well when the ATD was seated. However, once removed to permit the occupant to stand, the lack of pelvic restraint reduces load distribution across the skeletal structure. This may be improved with the use of integrated pelvic restraints on retractors similar to those found on Systems B–D. While Systems B–D differed in geometry, each utilised a lighter weight harness-like restraint system. During load attenuation, each restraint experienced slippage in the shoulder area when side facing with respect to the impact direction. However, the yoke system employed by Systems C and D provided improvement in this condition compared to the geometry of System B. The addition of a lateral web or modified yoke member should be considered to improve harness fit and the likelihood the shoulder harnesses will remain more closely coupled with each shoulder.

Given the goal of this testing was to identify opportunities to reduce the likelihood of injury during a crash, thoughtful consideration must be given to the location of retractors from a strike hazard perspective. Flush mounting of retractors should be considered a primary design goal during the build process. Finally, strong consideration must be given to retractor design. While each restraint system used a different model retractor, vehicle-sensing retractors offered benefits over web-sensing-only retractors. However, differences in retractor lock-up limits will influence their performance and should be optimised to fit work operational requirements. To enable the occupant to stand, all of the systems incorporated long retractable tethers. When the occupant is seated, a large amount of the tether is retracted into the retractor. In a crash, when the restraint is loaded the large amount of webbing wrapped around the retractor spool will tighten down and stretch, introducing slack to the tether. This slack increases body excursion and degrades crash ride down, increasing the potential for injury. To eliminate this phenomenon retractors incorporating web clamps or pretensioners should be considered.

10. Conclusions

In the 1960s and 1970s, the use of lap belts and later lap/shoulder belts represented incremental improvements to the safety of an occupant in a passenger vehicle. While today's ambulances are equipped with lap belts, which this research confirms provide some benefit in certain crash loading events, peers have reported the lap belt is not used by most workers when transporting a patient as they impede the worker's ability to satisfactorily provide patient care.

Recognising the industry trend is for ever larger vehicles carrying a wider array of medical devices, the need to move from the primary seating position appears to be growing. The combination of larger vehicles equipped with severe mobility-limiting restraint systems (lap belts), coupled with a demand for increasing patient care task performance, has led to a potentially dangerous work environment.

The lessons learned and advances made in occupant crash protection in automobiles and aircraft have direct application to ambulances. While technologies such as inflatables, pretensioners and the crash sensing systems needed to initiate those systems may not be feasible to incorporate in the short term, more effective restraints incorporating upper torso restraint will likely significantly increase crash protection. The unique need to provide mobility for ambulance medical personnel creates a challenge that may be addressed by retrofitable mobility restraints. As discussed above, all restraints have limits of protection and introduce their own unique unintended risks or consequences, but their benefits typically far outweigh those risks. Tests and analyses of the retrofitable mobility restraints indicate a high potential for the reduction of injury risk to ambulance patient compartment workers.

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Disclaimer

The findings and conclusions in this report are those of the authors and do not necessarily represent the views of the National Institute for Occupational Safety and Health.

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