PRELIMINARY STUDY ON CRASH PULSE INFLUENCE FOR CHILD ATD RESPONSE IN CHILD RESTRAINT SYSTEMS

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ABSTRACT
Different vehicle crash scenarios may produce different crash pulses dependent on several variables. Crash simulations utilizing a sled system are more repeatable but may subject anthropomorphic test devices (ATDs) to different input pulse levels depending on sled type and its settings. Those different input pulses may influence the test device's response. The goal of this current study examined different sled pulse inputs and their influence on child ATDs. ATDs were secured in child restraint systems on the proposed updated frontal test bench for Federal Motor Vehicle Safety Standard (FMVSS) No. 213 and subjected to three input pulses with the same target delta-v (48 kph). All three input pulses were within the FMVSS No. 213 boundaries. Hybrid III 10-year-old and 6-year-old test devices were tested using four belt positioning booster seats and one forward facing harness seat. Head, chest, neck, and belt load metrics were examined for coefficient of variation, trends related to input pulse acceleration increases, and any significant differences. Examination of the study results indicate that increased acceleration pulse inputs had the most influence on head accelerations, chest accelerations, and neck tensions but had little effect on chest deflections or head and knee excursions.

INTRODUCTION
Many studies have examined the effect of crash pulses on occupant kinetics. Crash pulse shape or characteristics can be influenced by vehicle structure, crash delta-v, and collision partner [1], [2], [3], [4], [5]. One study examining full scale crash tests of 1998 to 2002 model year vehicles including full frontal (48 kph), offset deformable (40 kph), and underride guard barriers (64 kph) demonstrated distinctly different pulse shape during the vehicle deceleration from the three types of crash tests [1].

Results from a study by Locey [2] examining crash pulse characteristics (CPC) of model year vehicles from 1997 - 2010 found new vehicle crashes within the same vehicle group and under the same conditions have become more homogenous as the model years increased. They found the range of peak acceleration was greater for the model year 1997 – 1999 vehicle group (17.1 G) vs the model year 2009 – 2010 vehicle group (10.7 G) [2]. That study also noted the same trend for maximum acceleration time and pulse duration was reduced from group to group for the newer model year vehicles. One conclusion was that as crash pulses become more homogenous, restraint system design can become more universal.

Most studies have examined effects to adult occupants or adult sized anthropomorphic test devices (ATD). One modelling study using a HIII 50th male ATD model found different pulses with the same delta-v affected HIC and chest Gs [6]. Another adult modelling study concluded that the CPC influences the occupant kinematics and the timing of potential injury [7]. One study of four different sled pulse effects on the spine response of a HIII 50th% found changes to the thoracic load, lumbar load, pelvis accelerations, and belt loads [8]. One field study reported that long-term neck injury consequences were more likely as mean and peak accelerations increased [9].

A previous NHTSA study reviewing test procedures found that drivers subjected to a stiffer pulse had a higher frequency and risk of serious to fatal injuries [3]. The NHTSA study defined a "soft pulse" as having a duration longer than 125 ms and a peak deceleration below 20 Gs, while a "stiff pulse" was defined as short duration, under 110 ms, and higher peak decelerations, approximately 25 Gs.

One recent study examined pulse effects on a Hybrid III 6-year-old, Hybrid III 5th% adult female, Hybrid III 95th% adult male, and THOR ATDs [4]. In that study the authors used a soft and severe pulse based on NHTSA New Car Assessment Program (NCAP) crash tests conducted at 56 kph. The severe pulse had an approximate peak of 57 Gs and duration of 85ms while the soft pulse had an approximate peak of 28 Gs and 110 ms duration and same delta-v. This study [4] concluded sled pulse changes from a soft to severe pulse resulted in increased HIC, neck tension, chest Gs, and chest deflection.

Most of the pulse influence studies have used adult ATDs and models. There is limited data for the influence of pulse on child occupants and child occupants using add-on child restraint systems. It is known that adult and child injuries differ in vehicle crashes [10], [11]. This study's goal was to analyze any response differences of child ATDs in child restraint systems (CRS) secured on the Federal Motor Vehicle Safety Standard (FMVSS) No. 213 proposed updated frontal standard seat test bench [12] subjected to different sled acceleration pulse inputs. The current FMVSS No. 213 sled pulse peak acceleration corridor extends from 19 G to 25 G with a duration between 75 and 90 ms. The tests conducted in this research targeted three pulses peaking at 21.5 G, 23.0 G and 24.5 G and a pulse duration less than 90 ms [13].

METHODS AND DATA SOURCE

Four belt positioning booster (BPB) seats and one forward-facing CRS with an internal harness (FFH) (n = 5 CRS models) were tested using three different frontal acceleration pulses at a 48-kph change in velocity (delta-v) (n = 15 test configurations). The CRSs tested included no-back BPB (NB-BPB), high-back BPB (HB-BPB) and forward-facing harness CRS. Hybrid III 6-year-old (HIII-6YO) and Hybrid III 10-year-old (HIII-10YO) ATDs were used for testing on each type of BPB, and the FFH was tested only with a HIII-6YO.

The three pulses had target peak accelerations of 21.5 G, 23.0 G and 24.5 G (low, mid, and high-pulse respectively) while remaining within the FMVSS No. 213 frontal pulse upper and lower acceleration boundaries and targeting a 48 kph delta-v (Figures 1 & 2). Tests were conducted on a Seattle Safety accelerator type sled. FMVSS No. 213 research test setup procedures [13] were followed. ATDs and CRS were secured on the proposed updated frontal standard seat assembly per the FMVSS No. 213 research procedures [12], [13], [14] and the CRS manufacturer supplied instructions. A 3-point fixed belt was used for the BPB seats (tensioned at 2-4 pounds) and the LATCH was used for the FFH CRS (tensioned at 12-15 lbs).

All data were collected and post-processed according to SAE J211 [15] specifications. A total of 41 sled tests were used for analysis. Collected data included head and chest accelerations, chest deflections (not regulated in FMVSS No. 213), upper neck axial forces (not regulated in FMVSS No. 213), head and knee excursions, and seat belt webbing loads (not regulated in FMVSS No. 213). Additional data for two BPBs was downloaded from the NHTSA database for three low-pulse tests.

The ATDs' triaxial head and chest accelerometers, chest potentiometer, upper neck force transducer, and belt webbing load cells were collected at 20,000 Hz. utilizing a DTS (Seal Beach, CA) data acquisition system. Seat belt webbing loads were measured with Denton seat belt load cells (Denton, Michigan, United States) installed on the shoulder and lap belt, and when possible, on the lower anchors and top tethers. Head and knee excursion data were calculated from video analysis using TEMA (Specialised Imaging, United Kingdom) software using the 4-point target method [13].

Final ATD measurements were done with a coordinate measuring machine (CMM). Generally, head, thorax, belt and CRS positions were within 15 mm of each test for same CRS and ATD set up. Belt tensions were set per the FMVSS No. 213 research procedures [13]. Comparisons between corresponding peak metrics for each ATD-CRS combination and the acceleration inputs were analyzed. When data for more than one test was available, the results were averaged in each pulse group and a coefficient of variation calculated. Generally, each individual CRS pulse group had good repeatability for the metrics in this research.

Regression analysis was conducted to determine trends related to the changing pulses. R^2 greater than 0.8 was considered an excellent fit, and an R^2 greater than 0.7 was considered a good fit. Single factor ANOVA was computed for each CRS-ATD group to determine if there was a significant difference between the pulse groups. P-values less than 0.05 were considered as being significantly different. Post hoc Tukey tests were performed to determine if a specific group in each metric was related to causing the difference. Coefficient of variation (CV) was calculated for the metrics and pulses for each CRS-ATD-Pulse group when more than one sled test was available, 10% or less was considered a low degree of variation and good test repeatability. For each metric with a CV, the team calculated the substantiveness of the variation relative to the IARV or performance limit. Sigma-to-Limit (StL, σ_L) (Equation 1) results above 2.0, would indicate at least two standard deviations between the average response and the IARV or performance limit.

$$Sigma - to - Limit\ (STL, \sigma_L) = \frac{(Limit - \bar{x})}{\sigma} \qquad \textbf{\textit{Equation 1}}$$

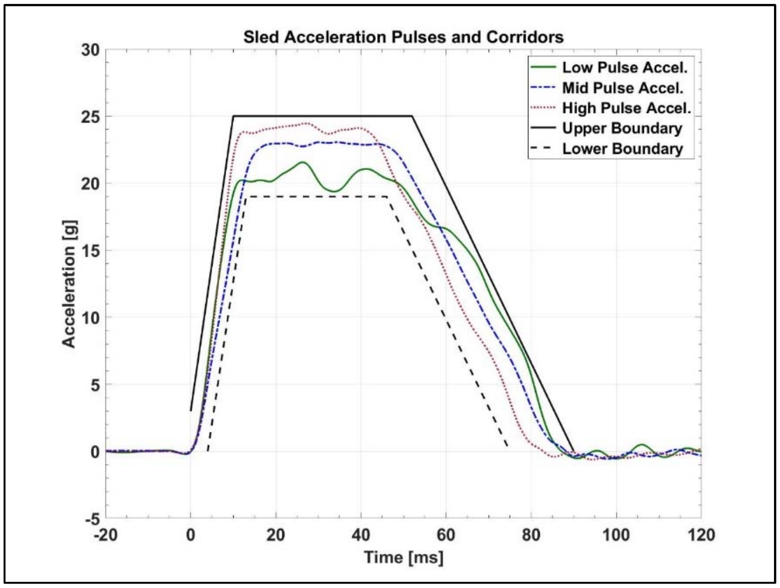

Figure 1. Average acceleration vs time with FMVSS No. 213 upper and lower boundaries

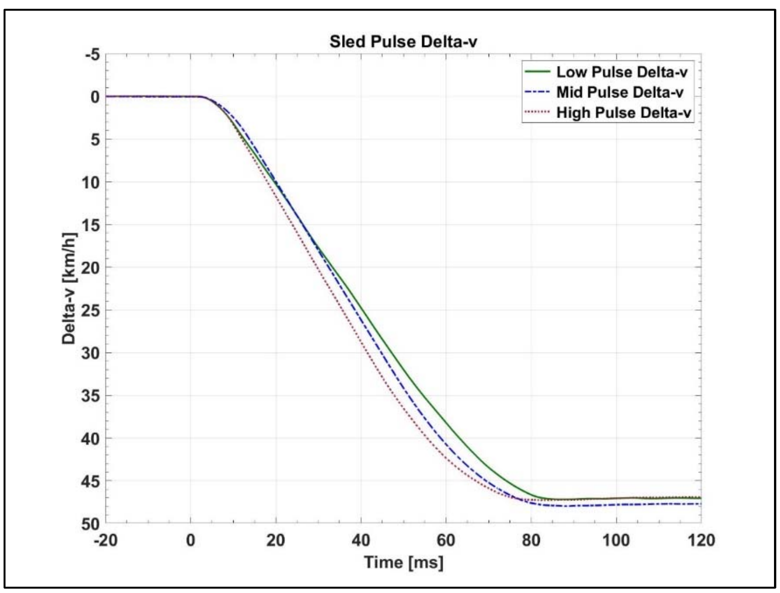

Figure 2. Average delta-v vs time of three input pulses.

RESULTS

The five different CRS models were evaluated comparing each CRS results from each pulse group. Results include analysis of HIC36, 3ms chest acceleration, chest deflection, neck tension, head excursion, knee excursion and belt loads. Averages were calculated from each pulse group and used to determine any correlation of increases or decreases between pulse inputs. Single factor ANVOA statistical results were calculated for each metric of the whole CRS group and pulses, and regression analysis was done comparing each metric relative to the pulse acceleration change. For this study p-values less than 0.05 are significant and R^2 values over 0.8 are considered excellent correlation. Tukey post hoc test were conducted for any metric found to be significantly different. Sigma-to-limit was calculated for each CRS / ATD group, and responses greater than two identify “good” levels of variation that are unlikely to cross the IARV or performance limit. Detailed results are presented in tables 1, 2, 3 and 4. IARV and performance values are included for reference only, some metrics are not regulated by the FMVSS No. 213 such as neck tension, chest deflection, belt loads and HIC36 for the HIII 10YO.

Head Acceleration

Thirty-six millisecond head injury criteria (HIC36) were calculated for each test. All five CRS had increased HIC values as the peak sled pulse acceleration was increased (R^2 = 0.88-0.96). Three of the CRS had significant HIC36 differences between the low- and high-pulses (all p<0.004). The NB-BPB with the HIII-10YO had the largest increase from the low to high peak pulse ranging from 502 to 831, and there was significant difference between pulses (P = 0.004). The FFH CRS with the HIII-6Y0 had the greatest significant difference (P = 0.0001) with average HIC36 values ranging from 389 to 625. Generally, the CV for HIC values within CRS pulse groups demonstrated good repeatability. Calculated CVs across all three pulses demonstrated high variation and ranged from 6.2% to 26.7%. Only the HB-BPB with HII-6YO had a CV less than 10% across the three pulses. The HIC values did not exceed IARV or performance limits for any test (Figure 3). The NB-BPB HIII-10YO sigma-to-limit calculation was 1.8, all other combinations were over 2.

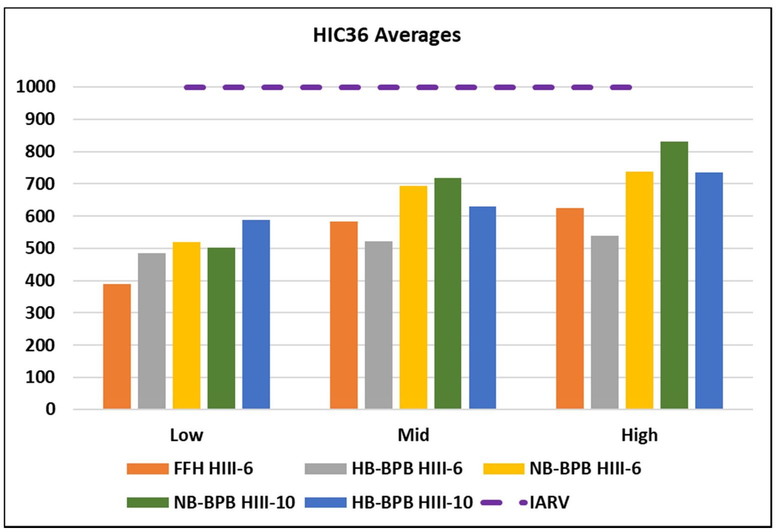

Figure 3. HIC 36 averages across pulses and IARV [16].

Chest/Thorax

ATD peak chest deflections (not regulated) were measured for each test. The HB-BPB with the HIII-6YO chest deflection recorded the largest increasing difference (8.2 mm) from low- to high-pulse (R^2 = 0.85) and was significantly different between pulses (P = 0.001). The HB-BPB with the HIII-6YO chest deflection ranged from -37 to -45 mm. The HB-BPB with a HIII-10YO also had significant differences between pulses (P = 0.03) but did not have any correlation nor trend related to pulse inputs and only recorded a 2 mm difference between the three pulses. Post hoc test of the results found that the significance differences were between the low- and mid-pulses. Other CRS chest deflections only varied about 1.0 mm. The chest deflections exceeded IARV values for the HIII-6YO in BPB seats for some tests in low-, mid- and high-pulse inputs [16].

Three millisecond chest accelerations (CLP3) were calculated from the thoracic resultant acceleration. All the CRS tested had increases of the calculated CLP3 with increasing pulse severity. Four of the CRS CLP3 calculations correlated well to the pulse increases ($R^2 = 0.82$ - 0.99) and one correlated good ($R^2 = 0.77$). The NB-BPB with HIII-10YO, had the lower correlation ($R^2 = 0.77$) due to the calculated average CLP3 for the mid and high-pulses being almost identical (47.9 G and 48.0 G respectively) while the low-pulse average was 42.7 G. The two BPBs with the HIII-10YO did not have significant CLP3 differences between the pulse inputs (P = 0.06 and 0.07), although the HB-BPB with HIII-10YO demonstrated a good correlation to pulse increases ($R^2 = 0.99$). All CRS tested with the HIII-6YO had CLP3 values that were significantly different between pulses (P = 0.0001 – 0.001). Post hoc analysis indicated that all the CRS with the HIII-6YO were significantly different between the low-pulse and both the mid and high inputs. The 6YO in BPBs had average CLP3 values near the maximum performance values (59.4 and 59.5 G) both during the high-pulse.

Both chest deflection and CLP3 demonstrated good repeatability and low variation across each CRS pulse group and across all the CRS pulse groups, ranging from 2.7% to 6.6% for chest deflection and 5.2% to 9.3% for CLP3. All sigma-to-limit values for chest deflections were over two. CLP3 sigma-to-limit calculations were under two for both BPBs with the HII 6YO. Average results are presented in Figures 4 and 5.

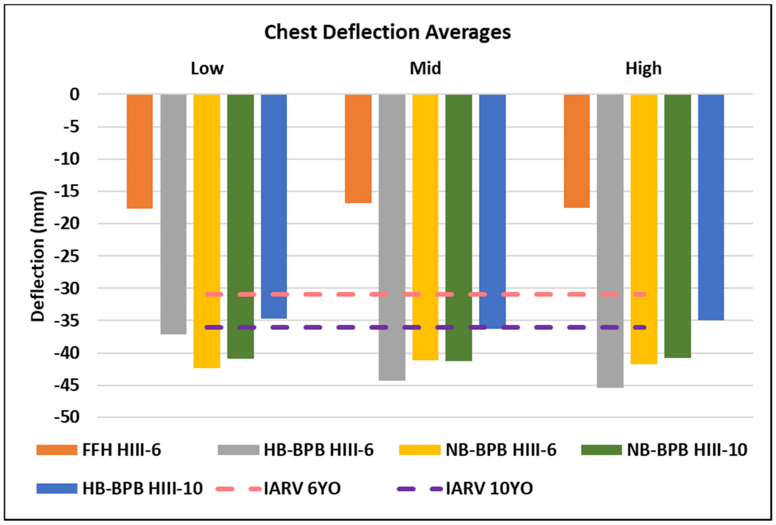

Figure 4. ATD Chest Deflection averages across pulses and IARV [16].

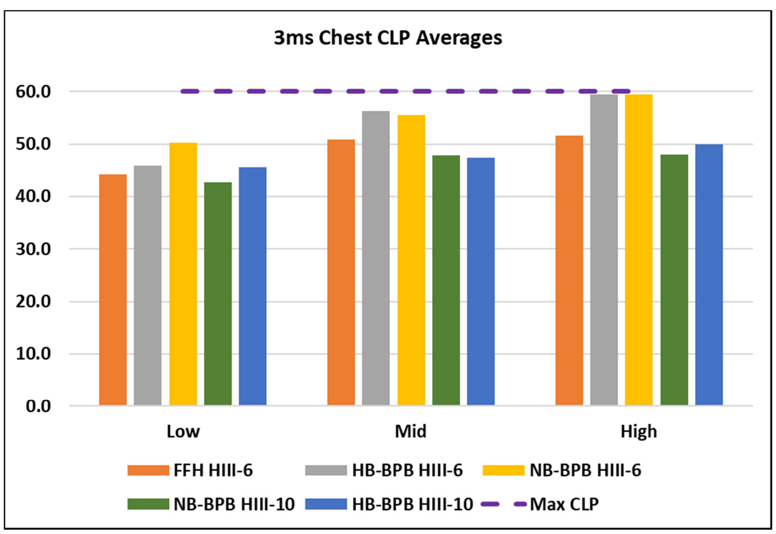

Figure 5.ATD Chest 3ms CLP averages across pulses and performance limit.

Neck Tension

Neck tension values (not regulated) were measured for each test and CRS-ATD combination. Four of the CRS had increasing neck tension values as the pulses were increased, three of which correlated well ($R^2 = 0.81 - 0.99$) and the fourth was good ($R^2 = 0.77$). One CRS did not have increasing neck tension with the increasing pulse, HB-BPB with HIII-10YO ($R^2 = 0.08$). The HB-BPB with HIII-10YO had a lower average peak neck tension during the three mid-pulse tests (2.7 kN) compared to the low-pulse (3.1 kN) and high-pulse (3.3 kN). Only three of the CRS had significant differences between the pulse inputs; HB-BPB with HIII-10YO, NB-BPB with HIII-6YO, and FFH with HIII-6YO (all $P \leq .04$). Calculated CVs within the CRS pulse groups varied from 1.8% for a HB-BPB with HIII-10YO during the high G pulse to 17.7% for the NB-BPB with HIII-6YO during the high G pulse. Only the HB-BPB with HIII-6YO had a CV less than 10% across all tests, the other CRS ranged from 11.2% to 20.4% indicating variations between the lower and higher pulses. Neck tension calculated sigma-to-limit values were less than two for all CRS except the HB-BPB with the HIII 6YO. Average neck tension results are presented in Figures 6.

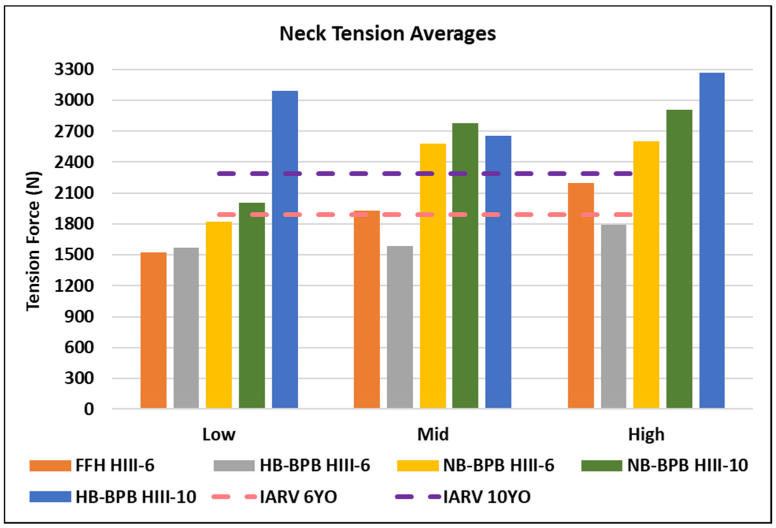

Figure 6. ATD Neck Tension averages across pulses and IARV [16].

Head / Knee Excursion

Measured head excursions between pulse groups were not significantly different for any CRS ($P > 0.05$). Both HB-BPB with HIII-10YO and HIII-6YO had increased head excursion which correlated with the increased pulse inputs ($R^2 = 0.97$ and 0.96 respectively). The measured increases were small, HB-BPB with HIII-10YO was 9 mm difference, and the HB-BPB with HIII-6YO was 18 mm difference from low to high-pulses. The FFH with HIII-6 had the greatest excursions (628 to 671 mm). The lowest excursions were the NB-BPB with the HIII-6YO (453 to 472 mm) There was a reduction of head excursion for the FFH CRS of 23 mm ($R^2 = 0.95$). There was no correlation of pulse inputs to changes of head excursion for the NB-BPB HIII-10YO and HIII-6YO ($R^2 = 0.01$ and 0.22 respectively).

Three CRS did not have any correlation to the pulse differences ($R^2 < 0.8$) for the measured average peak knee excursions for HB-BPB HIII-10YO, NB-BPB HIII-10YO and NB-BPB HIII-6YO. The NB-BPB with HIII-6YO knee excursion did demonstrate a significant difference ($P = .01$) between the pulse inputs which ranged from 567 to 598 mm ($R^2 = 0.7$). The average knee excursion during the mid-pulse test was only 2 mm larger than the high-pulse but was 31 mm larger than the low-pulse. The FFH HIII-6YO had a decreasing trend for average peak knee excursion ($R^2 = 0.75$) and resulted in 2 mm differences, between the low- to high-pulses. The greatest excursions occurred with the FFH with HIII-6YO (779 to 801 mm) while the lowest excursions were with the NB-BPB with HIII-6YO (551 to 607 mm).

Head and knee excursions were very repeatable and had low variation across the individual pulse groups and across the three pulse groups. CVs were very low across the three CRS pulse groups (< 3%). No excursion performance requirements were exceeded for any test (head < 720 mm and knee < 813 mm) (Figures 7 and 8). Sigma-to-limit calculations were over two for both head and knee excursions for all CRS groups.

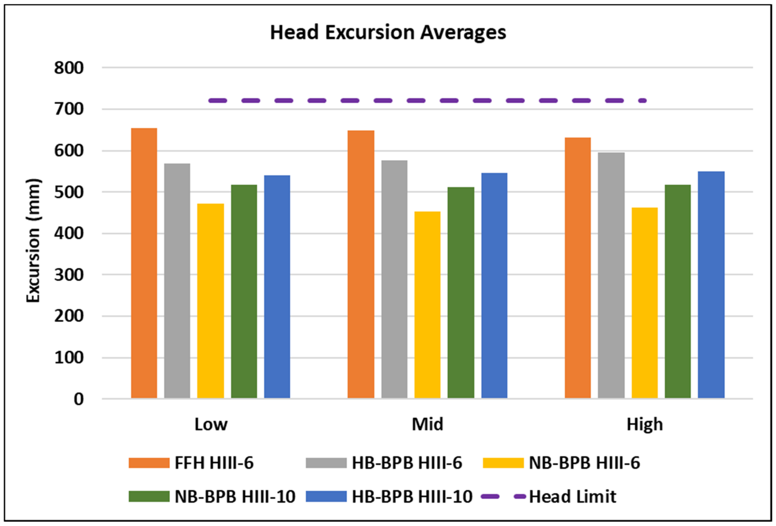

Figure 7. ATD Head Excursion averages across pulses and lower performance limit.

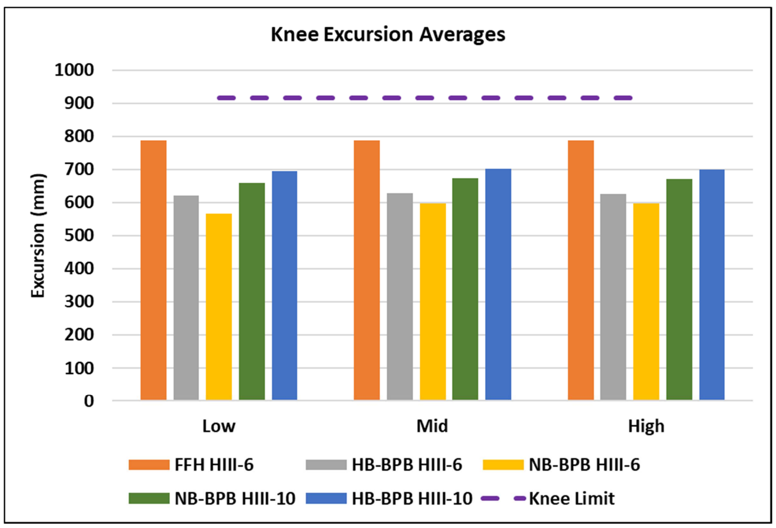

Figure 8. ATD Knee Excursion averages across pulses and performance limit.

Belt Webbing Forces

Measured peak shoulder belt loads (not regulated) for the HB-BPB and NB-BPB increased as the pulses increased ($R^2 > 0.8$) (Figure 9). Shoulder belt loads had significant differences for three of the BPB ($P = 0.001$ to 0.05), while the NB-BPB HIII-10YO calculated significant difference was $P = 0.06$.

The peak lap belt loads for the HB-BPB and NB-BPB increased as the pulses increased. Three of the CRS had an excellent correlation ($R^2 > 0.80$) and the HB-BPB with HIII-10YO had a good correlation ($R^2 = 0.76$) (Figure 10). Only the NB-BPB with HIII-6YO had a significant difference ($P = 0.001$) for the lap belt load with a 1.1 kN difference from the low to high-pulses. The HB-BPB with HIII-10YO lap belt loads was significantly different and while only presenting a 0.3 kN difference from the low- to high-pulse, while the mid- and high- pulse demonstrated nearly the same values (4.0 kN) for the lap belt loads ($P = 0.05$, $R^2 = 0.76$). Post hoc tests of this group indicated most significant differences were due to the low-pulse tests as compared to both the mid- and high-pulse tests.

The FFH CRS with the HIII-6YO was secured with lower anchor and tether webbing. The lower anchor and tether webbing peak loads did not have an increasing or decreasing trend as the pulse acceleration peaks became higher (R^2 = 0.57 and 0.05 respectively). The lower anchor and tether loads were significantly different across the pulse inputs (P = 0.01 and 0.0001 respectively). The lower anchor loads recorded a difference of 1.2 kN between the pulses with the highest average peak measured during the mid-pulse (5.3 kN). The tether only recorded a 0.4 kN peak difference between the pulses, while the low-pulse had the highest average peak load (3.1 kN).

Within each CRS pulse group, the belt webbing forces calculated CVs demonstrated good repeatability, CV<10% for all except one group which was 11.1%. For the combined three groups shoulder belt force CVs ranged from 3.3% to 10.6% and the lap belt force differences ranged from 3.8% to 16.5% for the BPBs. The LATCH lower anchors and tether forces CVs were 12.0% and 6.1% respectively across the three pulses for the FHH.

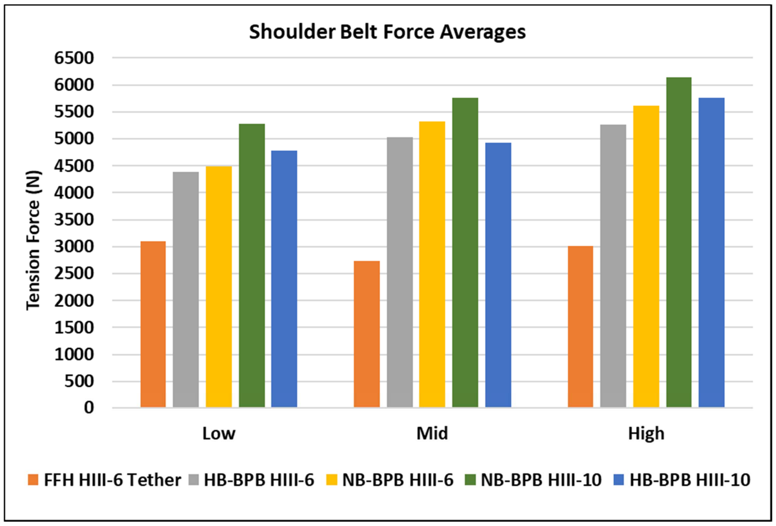

Figure 9. Shoulder/Tether Belt Webbing Force averages across pulses.

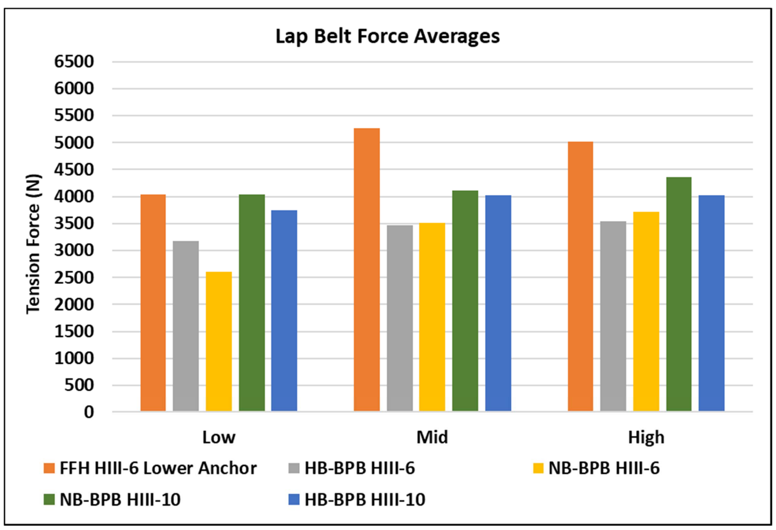

Figure 10. Lap/Lower Anchor Belt Webbing Force averages across pulses.

Table 1. ATD metric increases correlation to sled acceleration increases.

R2 values									
Type	ATD	HIC36	Chest Deflec	Chest CLP	Neck Tension	Head Excursion	Knee Excursion	Shoulder / Tether Belt Force	Lap / LA Belt Force
HB-BPB	HIII-10	0.94	0.02	0.99	0.08	0.97	0.35	0.86	0.76
NB-BPB	HIII-10	0.97	0.08	0.77	0.85	0.01	0.64	0.99	0.91
NB-BPB	HIII-6	0.89	0.24	0.99	0.77	0.22	0.70	0.92	0.88
HB-BPB	HIII-6	0.96	0.85	0.91	0.81	0.96	0.39	0.93	0.90
FFH	HIII-6	0.88	0.02	0.82	0.99	0.95	0.75	0.05	0.57

Table 2. ATD metric increase or decrease trend related to input pulse.

Trend Direction									
		HIC36	Chest Deflec	Chest CLP	Neck Tension	Head Excursion	Knee Excursion	Shoulder / Tether Belt Force	Lap / LA Belt Force
Type	ATD								
HB-BPB	HIII-10	increase	none	increase	none	increase	none	increase	increase
NB-BPB	HIII-10	increase	none	increase	increase	none	none	increase	increase
NB-BPB	HIII-6	increase	none	increase	increase	none	none	increase	increase
HB-BPB	HIII-6	increase	increase	increase	increase	increase	none	increase	increase
FFH	HIII-6	increase	none	increase	increase	decrease	decrease	none	none

Table 3. ATD metric p-values related to differences across all pulses.

P values (F - value)									
Type	ATD	HIC36	Chest Deflec	Chest CLP	Neck Tension	Head Excursion	Knee Excursion	Shoulder / Tether Belt Force	Lap / LA Belt Force
HB-BPB	HIII-10	0.002	0.03	0.06	0.04	0.36	0.70	0.05	0.24
(F-value)		22.163	6.38	4.687	5.54	1.23	0.31	6.03	1.94
NB-BPB	HIII-10	0.020	0.98	0.07	0.13	0.92	0.22	0.004	0.54
(F-value)		11.570	0.02	5.45	4.35	0.08	2.29	30.16	0.71
NB-BPB	HIII-6	0.004	0.69	0.001	0.04	0.06	0.01	0.001	0.001
(F-value)		16.642	0.40	24.506	5.68	4.65	11.41	23.372	23.541
HB-BPB	HIII-6	0.07	0.001	0.0001	0.13	0.26	0.41	0.009	0.20
(F-value)		4.199	81.214	29.810	3.54	1.92	1.11	19.430	2.43
FFH	HIII-6	0.0001	0.67	0.0005	0.00001	0.18	1.00	0.0001	0.0078
(F-value)		55.896	0.427	36.022	133.66	2.27	0.04	73.791	12.15

Table 4. Sled and ATD minimum, maximum, min-max differences, average, standard deviation, coefficient of variation, and sigma-to-limit across all three pulse groups and performance limits and/or IARV [16].

CRS Type	ATD		Peak Accel	HIC36 (10 YO*)	Chest Deflec. (mm) *	Chest CLP (Gs)	Neck Tension (N) *	Head Excursion (mm)	Knee Excursion (mm)	Shoulder Belt Force (N) *	Lap Belt Force (N) *
IARV or Performance Criteria (10YO / 6YO)			n/a	1000	belt only (36/31)	60	(2290/ 1890)	720 /813	915	n/a	n/a
HB-BPB	10	Min	21.1	589	-36	45.5	2658	541	695	4785	3749
		Max	24.5	734	-35	50.0	3266	549	702	5761	4024
		Diff	3.4	146	2	4.5	608	9	7	976	275
		Avg.	22.9	651	-35	47.6	3004	548	699	5206	3954
		SD	1.5	69	1	2.5	336	7	11	551	191
		CV	6.4%	10.6%	2.7%	5.2%	11.2%	1.3%	1.5%	10.6%	4.8%
		StL, σ_L		5.0	78.9	5.0	-2.1	37.3	20.6		
NB-BPB	10	Min	21.1	502	-41	42.7	2003	511	658	5280	4042
		Max	24.1	831	-41	48.0	2904	518	674	6136	4363
		Diff	3.0	330.0	1	5.3	901	7	16	856	321
		Avg.	22.6	674	-41	45.7	2583	516	666	5716	4190
		SD	1.5	180	2	3.3	524	12	9	443	381
		CV	6.9%	26.7%	5.1%	7.3%	20.3%	2.3%	1.6%	7.7%	9.1%
		StL, σ_L		1.8	36.6	4.3	-0.6	25.6	26.8		
NB-BPB	6	Min	21.2	518	-42	50.2	1821	453	567	4485	2611
		Max	24.6	737	-41	59.4	2603	472	598	5611	3713
		Diff	3.3	218	1	9.2	782	19	32	1126	1103
		Avg.	23.0	650	-42	55.0	2334	462	587	5141	3277
		SD	1.5	109	2	4.2	476	11	17	539	539
		CV	6.3%	16.8%	3.7%	7.7%	20.4%	2.3%	2.9%	10.5%	16.5%
		StL, σ_L		3.2	49.8	1.2	-0.1	33.2	19.0		
HB-BPB	6	Min	21.2	485	-45	45.9	1570	576	626	4385	3176
		Max	24.6	538	-37	59.5	1794	594	629	5256	3540
		Diff	3.4	53	8	13.7	225	18	3	871	364
		Avg.	23.4	524	-44	56.2	1674	583	626	5032	3455
		SD	1.3	32	3	5.2	141	16	5	324	174
		CV	5.5%	6.2%	3.0%	9.3%	8.4%	2.7%	0.9%	6.4%	5.0%
		StL, σ_L		14.9	26.6	0.7	4.4	14.4	56.7		
										Tether (N)	Lower Anchor (N)
FFH	6	Min	21.3	389	-18	44.2	1526	632	786	2738	4041
		Max	24.6	625	-17	51.6	2200	655	788	3094	5264
		Diff	3.3	236	1	7.4	674	23	2	355	1223
		Avg.	23.0	532	-17	48.9	1885	645	787	2949	4775
		SD	1.4	112	1	3.7	297	15	8	180	571
		CV	6.3%	21.0%	6.6%	7.5%	15.8%	2.4%	1.0%	6.1%	12.0%
		StL, σ_L		4.2	43.9	3.0	0.02	4.9	16.8		

Notes: Three tests from another other lab were included for low pulse tests; data includes HIC 36 and Chest CLP, Chest Deflection, Belt Forces and excursions NB-BPB with HIII10YO, and HB-BPB with HIII 6YO.

* HIC36 for HIII-10, chest deflection, neck tension, and belt webbing forces are not regulated by FMVSS No. 213.

DISCUSSION

This preliminary study examined different sled pulse inputs influence on child ATD biomechanics secured using CRS on the proposed FMVSS No. 213 updated frontal standard seat assembly test bench fixture using three different peak acceleration pulses (targeting peak 21.5 G, 23.0 G, and 24.5 G) utilizing the same delta-v (48 kph). This research evaluated the measured metrics for trends, repeatability, and significant differences across the different pulses. The test set up during this research included the same CRS and belt configurations in each group with only the acceleration pulse changing.

In this research, while the peak sled pulse acceleration values, approximately 21.5 G to 24.5 G, from the three test groups demonstrated a CV less than 7%, ATD metrics had higher CV values, increasing trends, some variations, and differences. While the delta-v remained consistent with less than 0.8 kph difference across the differing pulses, some occupant metrics were influenced by the pulse changes. This research found the small increases in peak pulse had the most influence on the head accelerations, chest acceleration, neck tensions and shoulder belt webbing tension. Results indicated only small increasing or decreasing trends for the chest deflections, head excursions and knee excursions. Chest deflections, head excursions and knee excursions generally did not demonstrate significant differences, any large variations, and appear be affected little by pulse changes.

Test device data differences may be attributed to lab set up processes, test fixtures, material types, restraint material characteristics, CRS design, CRS variability, and/or ATD instrumentation or construction variances. Sled systems input pulses are controlled with several methods such as computer-controlled braking, air pressure modulation, or hydraulic or mechanical deceleration control. Each producing their own CPC. Repeatability and reproducibility testing can account for differences in test methods, test labs, test pulses (within the specified boundary conditions) and ATDs with a calculated CV.

Vehicle CPC vary from one vehicle model to another model. Objects impacted can also change a CPC. As vehicle structure and CPC may influence injury it is important to examine the standard test methods to evaluate the different influences on the ATDs. This investigation is important to ensure the test methods, ATDs and crash pulses are robust enough to capture different crash scenarios. Additionally, as lab differences can be encountered, this research can identify CRS and ATD sensitivities related to the acceleration changes.

Other studies have examined the crash pulse influence on occupant kinetics. Most studies were conducted examining the influence on adult occupants, or supplemental restraint systems ([1], [3], [4], [6], [8], [17], [18]. Research has also been conducted to analyze the vehicle structure for crashworthiness [5] and studies found an increase of thoracolumbar spine fractures increased with newer model year vehicles [17], [18]. Pintar [17] suggested that the possibility of stiffer vehicle structure may influence injury patterns, and that study also concluded the struck object had an influence on the crash pulse and subsequent injuries.

Locey's [2] research found that the newer vehicle fleets, including all vehicle types, have a more homogenous pulse. Their research reported that the latest vehicle group in their dataset (2009 – 2010) only had a peak acceleration difference of about 11 Gs and pulse duration difference of about 13 ms. Across all three pulse variations, the average sled acceleration peaks in our research had a 3 G difference and about a 10 ms difference in pulse duration (Figure 1).

As vehicle CPC become more homogenous it will be advantageous for CRS design and regulatory performance requirements. The FMVSS No. 213 pulse used in this research allows for a range peak acceleration from 19 G to 25 G and a 75 to 90 ms duration. The current FMVSS No. 213 small pulse variations presented in this study are more homogenous than the current vehicle fleet but did present some differences of some measured metrics.

The small peak acceleration changes in pulse during this preliminary study found minimal effect on the ATDs chest deflections, head excursions and knee excursions. Our study used the fixed 3-point belt system specified in the FMVSS No. 213 procedures for the BPB. The FFH CRS used the child restraint manufacturer lower belt webbing and a tether attached to the LATCH anchors. The fixed belt and LATCH webbing likely limited the head and knee motion during the different pulses.

There were strong correlations of increased head acceleration, chest acceleration, and neck tension as the peak pulse acceleration increases for most of the CRS. The head injury and chest injury values are measured with accelerometers; it would be expected that there would be an increasing trend since pulse acceleration was increased.

Force is based on mass times acceleration and therefore the neck force trend increase would also be expected. Of those which had a good correlation, only three CRS in each metric demonstrated significant differences between pulses.

Seat belts and CRS belts with load limiters may have more influence on head and knee excursions and chest deflections, particularly as ATD or occupants become heavier. Load limited seatbelts and CRS or other advanced restraints may decrease the head and chest accelerations and neck tensions during increased accelerations as described in another study [4]. The Hu study [4] results are similar to our study except for the chest deflection. That study [4] used a production rear vehicle seat and belt system, and other advanced restraint systems.

Limitations in this study include that only two ATDs, HIII-10YO and -6YO, were examined. Only five models of CRS were tested which included NB-BPB, HB-BPB, and FFH CRS types. Two groups of data had only one test and three tests were data collected at a different facility (Table 5). Although these limitations exist the results are similar to other research conducted.

CONCLUSIONS

This preliminary study concluded that a small increase in peak sled acceleration, while maintaining constant delta-v, influenced ATD head acceleration, chest acceleration and neck tension values most. The change in pulse did not significantly affect chest deflection, and head and knee excursion of ATDs restrained in most CRS, although there may have been slight changes. Many metrics increased with the pulse changes and resulted in differences, but they only had small variations, which included belt webbing forces, chest accelerations, and other individual CRS-ATD metrics. Generally, CV results and ANOVA calculations both demonstrated the pulse influence for head accelerations and neck tensions, while other metrics varied across both analysis methods. This study includes new research related to child ATDs and CRS sensitivity during different test conditions.

Table 5. CRS and ATD test notes

Model	Type	ATD	Number Low Pulse Tests	Number Mid Pulse Tests	Number High Pulse Tests	Notes
Evenflo Amp	**HB-BPB**	**HIII-10**	**3**	**3**	**3**	
Chicco GoFit	**NB-BPB**	**HIII-10**	**3 ***	**1**	**3**	Limited data, only 1 test 23 G pulse tests
Harmony Youth	**NB-BPB**	**HIII-6**	**3**	**3**	**3**	
Graco Turbobooster	**HB-BPB**	**HIII-6**	**1****	**3**	**3**	Limited data, only 1 test 21.5 G pulse tests
Evenflo SureRide	**FFH**	**HIII-6**	**3**	**3**	**3**	Tether and lower anchor forces

* Two tests from the NHTSA database ** One test from the NHTSA database

Low-pulse = peak approximately 21.5 G / 48 kph
Mid-pulse = peak approximately 23.0 G / 48 kph
High-pulse = peak approximately 24.5 G / 48 kph

ABBREVIATIONS

ATD - Anthropomorphic Test Devices
CPC - Crash Pulse Characteristics
CRS – Child Restraint System
CV - Coefficient of Variation (CV)
FMVSS – Federal Motor Vehicle Safety Standard
HB-BPB – high-back belt positioning booster seat
NB-BPB – no-back belt positioning booster seat
HIC – Head Injury Criteria
LATCH – Lower Anchors and Tether for Children
NHTSA – National Highway Traffic Safety Administration
FFH – Forward-facing-harness child restraint system
HIII-10YO – Hybrid III 10-year-old ATD
HIII-6YO - Hybrid III 6-year-old ATD

REFERENCES

[1] Comeau, J. L., German, A., & Floyd, D. (2004). Comparison of Crash Pulse Data from Motor Vehicle Event Data Recorders and Laboratory Instrumentation. *Proc. CMRSC-XIV*, 27-30.

[2] Locey, C. M., Garcia-Espana, J. F., Toh, A., Belwadi, A., Arbogast, K. B., & Maltese, M. R. (2012, October). Homogenization of vehicle fleet frontal crash pulses from 2000–2010. In *Annals of advances in automotive medicine/annual scientific conference* (Vol. 56, p. 299). Association for the Advancement of Automotive Medicine.

[3] Hollowell, W. T., Gabler, H. C., Stucki, S. L., Summers, S., & Hackney, J. R. (1999). Updated review of potential test procedures for FMVSS No. 208. *NHTSA Docket*, 6407-6.

[4] Hu, J., Fischer, K., Lange, P., & Adler, A. (2015). *Effects of crash pulse, impact angle, occupant size, front seat location, and restraint system on rear seat occupant protection* (No. 2015-01-1453). SAE Technical Paper.

[5] Gu, L., Tyan, T., Li, G., & Yang, R. J. (2004, January). Vehicle structure optimization for crash pulse. In *International Design Engineering Technical Conferences and Computers and Information in Engineering Conference* (Vol. 46946, pp. 945-951).

[6] Mark, S. (2003). Effect of frontal crash pulse variations on occupant injuries. In *Proceedings: International Technical Conference on the Enhanced Safety of Vehicles* (Vol. 2003, pp. 7-p). National Highway Traffic Safety Administration.

[7] Grimes, W. D., & Lee, F. D. (2000). *The effect of crash pulse shape on occupant simulations* (No. 2000-01-0460). SAE Technical Paper.

[8] Hauschild, H. W., Halloway, D., & Pintar, F. A. (2015). ATD spine response as a function of crash pulse input. *Traffic Injury Prevention*, *16*, S237-S240.

[9] Kullgren, A., Krafft, M., Nygren, Å., & Tingvall, C. (2000). Neck injuries in frontal impacts: influence of crash pulse characteristics on injury risk. *Accident Analysis & Prevention*, *32*(2), 197-205.

[10] Rao, R. D., Berry, C. A., Yoganandan, N., & Agarwal, A. (2014). Occupant and crash characteristics in thoracic and lumbar spine injuries resulting from motor vehicle collisions. *The Spine Journal*, *14*(10), 2355-2365.

[11] Bilston, L. E., Clarke, E. C., & Brown, J. (2011). Spinal injury in car crashes: crash factors and the effects of occupant age. *Injury Prevention*, *17*(4), 228-232.

[12] NHTSA. (2019b). NHTSA Standard Seat Assembly; FMVSS No. 213, NHTSA-213-2016 Drawings. Child Frontal Impact Sled – V2. May 2019. Docket No. NHTSA-2020-0093-0004

[13] NHTSA. (2020). NHTSA Research Procedure for the Proposed FMVSS No. 213 Frontal Impact Test. Nov 2020. Docket No. NHTSA-2020-0093-0016

[14] NHTSA. (2019a). Federal Motor Vehicle Safety Standard (FMVSS) No. 213 Updated Frontal Standard Seat Assembly. Solicitation no. 693JJ919R000042. July 19, 2019

[15] SAE International Surface Vehicle Recommended Practice. Instrumentation for Impact Test, SAE J211. 2014.

[16] Mertz, H. J., Irwin, A. L., & Prasad, P. (2003). Biomechanical and scaling bases for frontal and side impact injury assessment reference values (No. 2003-22-0009). *SAE Technical Paper*.

[17] Pintar, F. A., Yoganandan, N., Maiman, D. J., Scarboro, M., & Rudd, R. W. (2012, October). Thoracolumbar spine fractures in frontal impact crashes. In *Annals of Advances in Automotive Medicine/Annual Scientific Conference* (Vol. 56, p. 277). Association for the Advancement of Automotive Medicine.

[18] Doud, A. N., Weaver, A. A., Talton, J. W., Barnard, R. T., Meredith, J. W., Stitzel, J. D., ... & Miller, A. N. (2015). Has the incidence of thoracolumbar spine injuries increased in the United States from 1998 to 2011?. *Clinical Orthopedics and Related Research®*, *473*(1), 297-304.