Simulating the Influence of Lap-belt Load Limiting on Injury Risks for Upright and Reclined Occupants with the SAFER Human Body Model

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I. INTRODUCTION

Pelvic wing fractures, fractures between the anterior superior iliac spine (ASIS) and the anterior inferior iliac spine (AIIS), have been reported in 50 km/h frontal sled tests with belted Post-Mortem Human Subject (PMHS) in both upright [1-2] and reclined seated postures [3]. A belt system including double lap-belt load limiters was proposed as an important addition to a restraint system to prevent pelvic wing fractures [2][4]. However, that proposed belt system was separated at the buckle into a shoulder belt and a lap belt; it was not a continuous 3-point belt.

The objective of this study was to evaluate the risk of pelvic wing fracture for a continuous 3-point belt system with various lap-belt load-limiting configurations: (1) a lap-belt load limiter at the end bracket; (2) a lap-belt load limiter at the buckle; and (3) two lap-belt load limiters – one at the end bracket and one at the buckle. A 3-point belt system without lap-belt load limiter served as reference.

II. METHODS

Frontal 50 km/h sled simulations were conducted using SAFER Human Body Model (HBM) v9 [5] seated on a generic semi-rigid seat [1][6] and restrained by a seat-integrated 3-point belt system. Two occupant postures were evaluated: upright (the seatback at 23° to the vertical axis); and reclined (the seatback at 45°). In both postures the seat pan was at 15° to horizontal.

Upright and reclined simulations were performed with the reference 3-point belt system without lap-belt load limiter. This reference system consisted of double 2 kN lap-belt pretensioners, a crash locking tongue and a shoulder-belt retractor with 4 kN load limiter and 2 kN pretensioners. The buckle pretensioner was activated at 3 ms and the shoulder and end bracket pretensioners at 9 ms. This belt system was developed and verified to avoid submarining in a reclined seated posture [3][7]. However, two out of five PMHS sustained pelvic wing fractures [3]. Simulations of the three variants of lap-belt load limiters were performed, keeping the 4 kN shoulder-belt loader limiter and the same activation times of the pretensioners, see Table I.

TABLE I
BELT SYSTEMS VARIANTS

Belt system	Shoulder belt		End bracket (right)		Buckle (left)	
	Load limiter	Pretensioner	Load limiter	Pretensioner	Load limiter	Pretensioner
No lap-belt load limiter (reference)	4 kN	2 kN	No load limiter	2 kN	No load limiter	2 kN
Lap-belt load limiter at the end bracket	4 kN	2 kN	5 kN	2 kN	No load limiter	2 kN
Lap-belt load limiter at the buckle	4 kN	2 kN	No load limiter	2 kN	7 kN	2 kN
Two lap-belt load limiters	4 kN	2 kN	5 kN	2 kN	7 kN	2 kN

As there is no injury risk function available for iliac wing fractures under frontal crash loading conditions, the pelvic wing forces (left and right) were evaluated. The forces were measured using a cross-section force in the area of the ASIS and AIIS of the HBM [6]. In addition to belt and buckle forces, the risk of rib fractures (AIS2+) for a 65-year-old occupant [8-9], risk of concussion (AIS2+) [10-11], and compression force at the first lumbar vertebra, L1 [6], were calculated and evaluated. The occupant head and pelvis kinematics were visualized by forward displacement plots in the x-z plane.

III. INITIAL FINDINGS

No submarining occurred in any simulation; however, pelvis displacement increased with the application of any lap-belt load limiter. Head-to-thigh contact was observed in the upright posture, but was avoided with two lap-belt load limiters. Head and pelvis forward trajectories are visualized in Fig. 1. Without the lap-belt load limiter, the belt forces reached approximately 9 kN and 13 kN at the end bracket and the buckle, respectively. (The buckle force sums up the lap-belt force and the shoulder-belt force.) The end bracket and buckle forces were

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reduced to 7.5 kN and 7 kN with the load limiter at the buckle, and to 5 kN and 9.5 kN with the load limiter at the end bracket. With two lap-belt load limiters the forces were reduced to 5 kN and 7 kN at the end bracket and at the buckle, respectively. The reduced lap-belt forces also substantially reduced the ASIS forces, indicating lowest pelvic wing loading with two lap-belt load limiters, see Table II. The highest rib fracture and concussion risks were indicated for the upright posture with the load limiter at the end bracket. There was no clear influence on the L1 compression forces, more than it was increased for the reclined posture.

TABLE II

ASIS PEAK FORCES, RISK OF RIB FRACTURES, RISK OF CONCUSSION AND L1 COMPRESSION PEAK FORCES

Belt system	Seated posture	Left ASIS peak force (kN)	Right ASIS peak force (kN)	Risk of rib fractures 65yo (AIS2+)	Risk of concussion (AIS2+)	L1 compression peak force (kN)
No lap-belt load limiter	Upright	4.9	3.4	3%	15%	2.2
Lap-belt load limiter at the end bracket	Upright	3.4	1.7	21%	45%	2.0
Lap-belt load limiter at the buckle	Upright	3.0	2.3	9%	9%	2.0
Two lap-belt load limiters	Upright	2.3	1.7	1%	6%	2.6
No lap-belt load limiter	Reclined	4.9	5.2	1%	30%	4.0
Lap-belt load limiter at the end bracket	Reclined	3.5	3.0	0%	11%	3.4
Lap-belt load limiter at the buckle	Reclined	3.1	3.2	0%	9%	3.7
Two lap-belt load limiters	Reclined	2.7	2.4	0%	7%	3.6

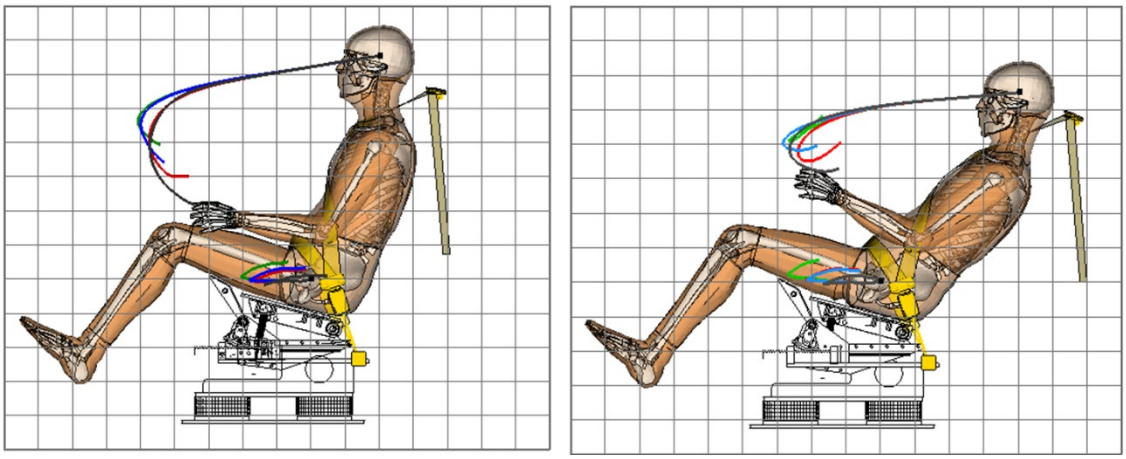

Fig. 1. Left: SAFER HBM in upright seated posture. Right: SAFER HBM in reclined seated posture. Both figures show head and pelvis forward trajectories for the four different belt systems. Black for no lap-belt load limiter, red for end bracket load limiter, blue for buckle lap-belt load limiter and green for two lap-belt load limiters.

IV. DISCUSSION

The belt system with two lap-belt load limiters effectively reduced the forces induced in the pelvis wings. One lap-belt load limiter, either at the buckle or at the end bracket, also reduced the ASIS forces, but not as effectively. Notably, use of only one lap-belt load limiter increased the risk of rib fractures for the upright occupant. However, longer pelvis forward displacement obtained with the two lap-belt load limiters can increase the risk of submarining or sliding off the seat. To explore this further, it is recommended that future work includes a wider range of HBM sizes, including larger occupants, different HBM postures (e.g. slouched seated postures) and more severe crash pulses. The results obtained in this preliminary study will guide the development of a mechanical 3-point belt system prototype to be used in reclined PMHS tests for the Enable New Occupant Position (ENOP) project [12]. Those forthcoming tests will further verify the protective capabilities of the proposed belt system with two lap-belt load limiters.

V. REFERENCES

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