Investigation of EvaRID Injury Risk Curves with ISO Technical Specifications

Z. Jerry Wang, Naveena Vijendra, Fuchun Zhu, Yun-Seok Kang

Abstract EvaRID, a scaled-down version of the BioRID finite element (FE) model developed in 2011, represents a 50th-percentile female for whiplash testing. This study developed injury risk curves for EvaRID virtual testing, planned as part of the China New Car Assessment Program (C-NCAP) roadmap. Eleven publications were selected for this study, comprising approximately 30 different load cases and a mix of volunteer and postmortem human subject (PMHS) data. Matched-pair simulations were conducted using EvaRID FE models. injury risk curves were generated for the predictors used in Euro NCAP, IIHS, and JNCAP rating thresholds. The Euro NCAP assessment thresholds were scaled down for EvaRID, and the corresponding injury risks were calculated using the IRCs developed in this study. The results showed that NIC, upper neck tension (Fz), and T1 acceleration correspond to lower injury risks (less than 8%). In contrast, Nkm, upper neck posterior shear force (Fx), upper neck moment (Myoc), lower neck shear force (Fx), and lower neck moment (My) all exhibited high injury risks (near or at 100%). Additionally, findings indicated that while upper neck posterior shear force, a required metric in NCAP assessments, has a weak statistical correlation with injuries, upper neck anterior shear force is strongly associated with injury outcomes.

Keywords 50th percentile female, EvaRID, neck, rear impact, whiplash

INTRODUCTION

Whiplash injuries remain a problem in real-world motor vehicle crashes (MVCs). While fatalities are rare, the economic costs associated with whiplash injuries are significant. The total annual cost of whiplash injuries in the United States, including both economic costs and quality-of-life impacts, has been estimated at approximately $2.7 billion [1]. In Europe, whiplash-related crashes have an estimated annual cost of around €10 billion, with a substantial impact on individuals and society [2]. Significant efforts have been made over the past few decades to improve safety measures. BioRID, a 50th percentile male dummy was developed to assess occupant safety in whiplash crash conditions. The research on female occupant whiplash injuries followed the pioneer work on males. Higher risk of whiplash injuries among females has drawn attention in the automotive industry. The higher injury risk for female could be caused by many factors, for example, the body geometry difference, shorter status, and headrest designed according to the males only in the current safety design requirement [3].

Accident data indicates that females have a greater risk of sustaining whiplash injuries than males [3]. Low-to-moderate speed rear impact experiments were predominantly conducted using a 50th percentile male anthropomorphic test device (ATD). The IIHS has used the 50th male rear impact ATD, BioRID II, to provide safety ratings for whiplash tests since 1995, but in 2023, the head restraint test rating was discontinued after nearly all vehicle models achieved good ratings [4]. Analysis suggested that a 24 km/h impact pulse is crucial for distinguishing meaningful differences between seat designs. However, the use of male ATD limits the development and assessment of female whiplash protection systems. A study in [5] showed that a scaled-down male FE model is not adequate to simulate female responses accurately. Moreover, male and female necks are not geometrically similar [6], which suggests female-specific ATDs are necessary. Differences in kinematic responses between males and females in rear impacts have been reported in volunteer studies [5][7]. In these previous studies, females tend to exhibit higher horizontal head accelerations, smaller head-to-T1 rearward displacements, and greater rebound motions than males. These kinematic differences between males and females highlight that the current male rear impact ATD could not represent female occupant responses and

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potential injuries in rear impacts. Given only male rear impact ATD exists, current seats, head restraints, and whiplash protection systems are not likely optimized for female occupants, which could contribute to the fact that whiplash protection systems often exhibit better effectiveness for males than for females in rear-world MVCs [10].

Research in [11] examined the initial spine curvatures of males and females in automotive seating at 20° and 25° seatback angles, as well as in standing and supine postures. The study found that seatback angles influence cervical and thoracic spine curvatures. Additionally, males and females exhibit different spinal curvatures in automotive seating, with variations across different seatback angles. However, the study scaled the spine curvature to 1.0 for comparison, and a full-scale size for the average male and female spine curvature was not provided. Such data would be valuable for ATD design, particularly for ATDs developed explicitly for female whiplash injury assessment.

EvaRID, a FE model of a 50th-percentile female rear-impact dummy, was developed in 2011 [8][12]. Data from the AGY Zurich and Folksam databases, which record occupant injuries and related insurance claims in Switzerland and Sweden, respectively, indicate that the most frequently injured females have a stature of 160-169 cm with a mean of 164.5 cm and mass of 60-64 kg, which are close to the average female size[12]. EvaRID was designed with the same design concept from BioRID, a 50th-percentile male rear-impact dummy developed in [13], but in accordance with the average female anthropometry EvaRID has a stature of 161.8 cm and a mass of 62.3 kg, representing a 50th-percentile female. Its external dimensions and body segment weights are documented in [8], while validation of the EvaRID FE model against biofidelity corridors from volunteer tests is provided in [12]. Injury thresholds for EvaRID were established in [12]. An analysis presented in [14] estimated that the risk of whiplash injuries for females is 20% higher than for males based on field data. Since males and females share similar anatomical structures, it was proposed that EvaRID injury thresholds should be estimated with a 20% reduction from the corresponding BioRID injury thresholds [12].

The objective of this study is to generate injury risk curves (IRCs) in accordance with ISO TS18506 for whiplash-related neck injuries, focusing the AIS1+ as the injury thresholds. These curves would provide injury assessment reference values (IARVs) at any injury risk level, offering a more comprehensive assessment compared to the previously applied simple reductions in IRAV values.

METHODS

This study followed ISO TS18506, a technical specification outlining procedures for constructing injury risk curves to evaluate road user protection in crash tests. Goodman-Kruskal Gamma (GKG) and Area Under the Receiver Operating Characteristic (AUROC) curves were used to assist in the final selection of survival fit models.

To develop IRCs, literature review was conducted, and the load cases (same test condition for each load case) are identified to be used in this study. FE simulations were conducted to generate response data from the EvaRID FE model in matched-pair tests with post-mortem human subjects (PMHS) or volunteers. No physical tests were conducted because the EvaRID physical dummy was not built yet. The data were then analyzed and correlated with injurious and non-injurious conditions observed in PMHS and volunteer studies.

Load Case Selection

A literature review was conducted to identify the maximum number of load cases for this study. The selected load cases include volunteer (non-injurious) and PMHS (injurious and non-injurious) cases [3][9][15], as outlined in Table I. A total of 30 load cases from 11 publications were identified and selected for analysis (see Table I for a summary). Of these, nine load cases involve female subjects, consisting of 31 female volunteers and four PMHS specimens. Due to the limited number of female data samples for injurious cases, male samples were included in the IRC construction. Publications that did not document boundary conditions were excluded from this study, as EvaRID simulations could not be reliably performed under those test conditions.

FE Simulation

FE simulations were conducted to generate data for this investigation. The loading environments, boundary conditions, and pulses were reconstructed based on information from the publications. The FE time history outputs from EvaRID model were then compared with biomechanical corridors published from volunteer and PMHS test data to ensure the results were consistent and did not produce any unusual outputs. The boundary conditions are generally adjusted with engineering judgement to prevent any egregious differences between the

FE output and the corridors. In general, the EvaRID results are lower than the mean of the corridors, but not far from the unscaled male corridors, considering the average male mass and stature of 76 kg and 175 cm and the average female mass and stature of 62.3 kg and 161.8 cm for EvaRID. The experimental seat and production seats were validated in previous projects [31][32]. The seat used in [22], Honda accord seat was downloaded from NHTSA file downloads [33]. For the foam material model used in[20], the seat was based on UNECE R44 and it was geometrically similar with the increased height to support the shoulder (590 mm height). The seatback was padded with stiff polyurethane (PU) foam with 70 mm thick and 30 kg/m^3 density. The seat base was padded with high density closed cell PU foam EV30) manufactured by Zotefoam™.

TABLE I
LIST OF THE LOAD CASES IDENTIFIED IN THIS STUDY

No.	Reference	Subject	Sex	# of Subject	Seat	Delta V (km/h)
1	Mertz et al. 1967 [15]	PMHS	M	2	Rigid Seat	16.0
		PMHS	M	2	Rigid Seat	24.0
		Volunteer	M	1	Rigid Seat	16.0
2	Ono et al. 2006 [9]	Volunteer	F	2	Rigid	5.8
		Volunteer	M	4	Rigid	5.8
3	Sato et al. 2014 [16] (test series 1)	Volunteer	F	8	Rigid	6.2
		Volunteer	F	3	Rigid	7.9
		Volunteer	M	12	Rigid	6.2
		Volunteer	M	10	Rigid	7.9
	test series 2, also in Ono et al. 2006 [9]	Volunteer	F	2	rigid	5.8
		Volunteer	M	4	rigid	5.8
4	Davidsson et al. 1998 [17]	Volunteer	M	8	Lab seat V	7.0
		Volunteer	M	7	Lab seat ES	7.0
5	Bertholon et al. 2000 [18]	PMHS	M	3	Rigid	10.8
6	Yoganandan et al. 2000 [19]	PMHS	M	1	Rigid	24.0
		PMHS	F	2	Rigid	24.0
		PMHS	F	2	Rigid	15.0
7	Roberts et al. 2002 [20]	Volunteer	M	10	Experimental Seat	7.0
8	Kang et al. 2012 [30]	PMHS	M	7	Experimental Seat	17.0
		PMHS	M	7	Experimental Seat	24.0
9	Kang et al. 2014 [21]	PMHS	M	2	Production Seat B	17.5 (JNCAP)
		PMHS	M	2	Production Seat B	24.0
10	Carlsson et al. 2021 [3]	Volunteer	F	8	Experimental Seat	7.0(HRC10C)
		Volunteer	F	6	Experimental Seat	7.0 (HRC15C)
		Volunteer	F	2	Experimental Seat	7.0 (HRC15NC)
11	Kang et al. 2021 [22]	PMHS	M	3	Production Seat	16 (FMVSS202a)
		PMHS	M	3	Production Seat	18.2 (JNCAP)
		PMHS	M	3	Production Seat	24.0

Scaling Method

EvaRID data were scaled to individual female subjects to directly use female injury outcomes in the IRCS. Although numerous scaling methods have been proposed in previous studies, in this study, the method defined in [23] was used when neck circumference and weight were available. If only weight and stature were provided, the method defined in [24] was applied instead. The scaling ratios, as defined in [23], are summarized below.

Mass: $\lambda_M = M_s/M_e$
Dimension x: $\lambda_x = C_s/C_e$
Dimension z: $\lambda_z = \lambda_M/\lambda_x^2$

Displacement x: λ_z^2/λ_x
Displacement z: λ_z
Neck force: λ_x^2

Stiffness: $\lambda_K = \lambda_x^4/\lambda_z^3$

Acceleration: $\sqrt{\lambda_K/\lambda_M}$

Angular velocity: $\sqrt{\lambda_K/\lambda_M}$

Neck moment: λ_x^3

Rotation: λ_z/λ_x

The scaling ratios according to the method defined in [24] are summarized below.

Mass: $\lambda_M = M_s/M_e$

Acceleration: ${\lambda_M}^{-\frac{1}{3}}$

Angular velocity: ${\lambda_M}^{-\frac{1}{3}}$

Neck force: ${\lambda_M}^{\frac{2}{3}}$

Neck moment: λ_M

Rotation: 1

Where

λ is the scaling factor,

Ms is the mass of the subject; Me is the mass for EvaRID of 62.3 kg,

Cs is the neck circumference of the subject; Ce is the neck circumference of the EvaRID

The scaling was applied only to the female test data. No scaling of the EvaRID data was performed on male volunteers or PMHS specimens due to the limited accuracy of scaling data from males to females. In such cases, all male tests under the same test conditions were counted as a single sample (data point) for injurious specimens and another data point for non-injurious specimens. Repeated PMHS tests with different specimens could result in varying Abbreviated Injury Scale (AIS) levels [25]. However, these tests were not distinguished individually but were categorized into either the injurious group (AIS 1+) or the non-injurious group (AIS 0). For the load cases that involved male subjects, EvaRID data were not scaled but were instead used directly for the corresponding female load conditions.

Calculations of Upper Neck Moment at Occipital Condyle, NIC and Nkm

The upper neck moments were transferred to the occipital condyle (OC). The lower neck x- and y-moment were not transferred to the C7/T1 joint, consistent with Euro NCAP.

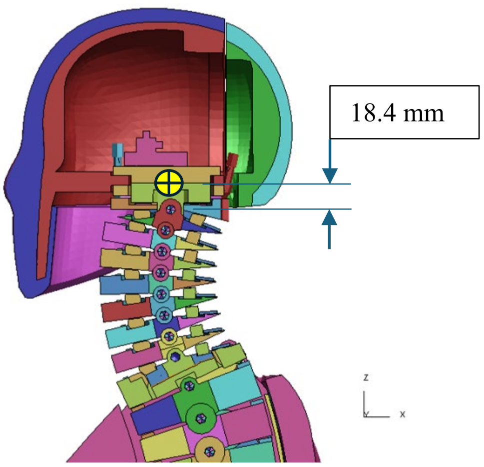

Fig. 1. Illustration of neck load cell and the OC joint distance (EvaRID FE model v1.5.1)

The Formula to calculate the moment at OC joint:

Mxoc = Mx + 0.0184*Fy (1)

Myoc = My – 0.0184*Fx (2)

Where

Mx and My are moments from the upper neck load cell measurements

NIC is calculated with the following formula [26]:

NIC(t) = 0.2*(relative x-acceleration(t)) + (relative velocity(t))2 (3)

Where

relative x-acceleration(t) = T1 x-acceleration (t) – head x-acceleration(t)

relative velocity(t) = numerical integration of relative x-acceleration(t)

Nkm is calculated with the following formula [27]:

$$\text{Nkm} = \frac{Fx(t)}{Fint} + \frac{My(t)}{Mint} \quad (4)$$

Where

Fx is shear force

My: flexion or extension moment

Fint = 507 for shear force [12]

Mint = 29 Nm for extension moment, 53 Nm for flexion moment [12]

Survival Model Fit

The survival model development in this study followed the ISO TS 18506 specifications [28]. The 10-step process outlined in the specification is as follows: 1) collect the relevant data; 2) assign the censoring status; 3) check for a single injury mechanism; 4) estimate the coefficients; 5) identify overly influential observations 6) check the distribution assumption; 7) choose the best distribution; 8) validate the prediction; 9) calculate 95% confidence intervals and assess their relative size; 10) check the quality index.

Additionally, the Akaike Information Criterion (AIC), Goodman-Kruskal Gamma (GKG), and Area Under the Receiver Operating Characteristic Curve (AUROC) were used to evaluate the survival model fits. Based on our findings, the Weibull survival model provided the best overall fit. Therefore, only the results from the Weibull model are presented in this study.

RESULTS

The analysis results were categorized into two groups. The first group includes parameters used in the NCAP and IIHS whiplash seat rating (prior to 2023), namely NIC, Nkm, shear force (Fx), tension (Fz), extension and flexion of the y-moment at the occipital condyle of the upper neck, shear force (Fx), tension (Fz), extension and flexion of the y-moment, and T1 acceleration (see Table II for details).

The second group consists of parameters investigated in previous research, including accelerations, displacements, y-rotation, and y-angular velocity of the head and T1, as well as neck-to-T1 relative displacements (NDCx and NDCz) and rotation (NDCr).

TABLE II

PREDICTORS USED IN IIHS WHIPLASH SAFETY RATING (PRIOR 2023) AND NCAPS

Injury Predictors	NIC	Nkm	Upper Neck				Lower Neck				T1
			Fx+	Fz+	Myoc Ext.	Myoc Flx.	Fx+	Fz+	My Ext.	My Flx.	Acc.
Euro NCAP	X	X	X	X	X	X	X		X	X	X
JNCAP	X		X	X	X	X	X	X	X	X	X
C-NCAP	X		X	X	X	X	X	X	X	X	X
IIHS			X	X							X

Results of the Predictors used in IIHS (prior 2023) and NCAPs

Injury risk functions for NIC, Nkm, upper neck compression and tension force, flexion and extension moment at occipital condyle (OC), lower neck anterior/posterior shear force, compression and tension, flexion and extension moment, and T1 acceleration are summarized in Table III. A few selected IRC plots for indictors with strong statistical significance are outlined here, including NIC, Npf, Npe, upper neck anterior shear (Fx-), upper neck extension moment (My-), lower neck posterior shear (Fx+), lower neck extension moment (My-), T1 rearward and upward accelerations (Ax- and Az-), see Fig. 1–10. The rest are attached in the Appendix for reference (Figure A1–A12).

The IRCs for NIC, Npf and Npe are shown in Fig. 2–4. The NIC IARVs for 5%, 25%, and 50% injury risk are 22.71, 34.25, and 42.23, respectively. For Npf, the IARVs for 5%, 25%, and 50% injury risk are 0.18, 0.28, and 0.36, respectively, while the Npe IARVs for 5%, 25%, and 50% injury risk are 0.25, 0.54, and 0.80, respectively.

The IRCs for the upper neck anterior shear (Fx-) and extension moment (My-) are shown in Fig. 5–6. The upper neck anterior shear (Fx-) has IARVs of 63.1 N, 83.5 N, and 96.4 N for 5%, 25%, and 50% injury risk, respectively. The upper neck extension moment (My-) has IARVs of 4.59 Nm, 6.69 Nm, and 8.12 Nm for 5%, 25%, and 50% injury risk, respectively.

TABLE III

IRC RESULT SUMMARY FOR NIC, NKM, NECK LOADS, AND T1 ACCELERATION AND HEAD TO CHEST RELATIVE MOTION

Location	Predictor	Description	Shape	Scale	GKG	AUROC	Qual. Index	IARVs 5%	IARVs 25%	IARVs 50%
NIC	NIC	neck injury	4.20	46.08	1.00	1.00	0.163	22.71	34.25	42.23
Nkm	Naf	anterior	12.88	0.20	0.67	0.83	0.106	0.16	0.18	0.19
	Nae	anterior	2.84	0.50	0.20	0.60	0.228	0.18	0.32	0.44
	Npf	posterior	3.69	0.39	1.00	1.00	0.200	0.18	0.28	0.36
	Npe	posterior	2.22	0.95	1.00	1.00	0.320	0.25	0.54	0.80
Upper Neck	Fx(+)	posterior	1.92	364.70	1.00	1.00	0.377	77.7	190.6	301.3
	Fx(-)	anterior	6.15	102.28	0.04	0.52	0.256	63.1	83.5	96.4
	Fz(-)	compression	5.11	191.69	1.00	1.00	0.159	107.2	150.2	178.4
	Fz(+)	tension	2.38	1345.42	1.00	1.00	0.295	387.3	798.0	1153.8
	My(-)	extension	4.57	8.79	0.94	0.97	0.168	4.59	6.69	8.12
	My(+)	flexion	6.21	4.46	1.00	1.00	0.111	2.76	3.65	4.20
Lower Neck	Fx(+)	posterior	2.22	549.57	1.00	1.00	0.323	143.8	313.2	465.8
	Fx(-)	anterior	2.92	158.97	1.00	1.00	0.213	57.58	103.83	140.25
	Fz(-)	compression	7.78	397.99	0.98	0.99	0.104	271.7	339.1	379.7
	Fz(+)	tension	1.97	641.07	1.00	1.00	0.394	142.3	340.9	532.4
	My(-)	extension	4.68	14.37	1.00	1.00	0.161	7.62	11.01	13.29
	My(+)	flexion	3.28	8.81	1.00	1.00	0.217	3.57	6.03	7.88
T1	Ax(-)	rearward	4.63	13.98	1.00	1.00	0.160	7.4	10.7	12.9
	Az(-)	upward	3.97	11.59	1.00	1.00	0.163	5.5	8.5	10.6
	Az(+)	downward	4.25	18.60	1.00	1.00	0.167	9.3	13.9	17.1
	Ar	resultant	3.59	30.22	1.00	1.00	0.186	13.2	21.4	27.3

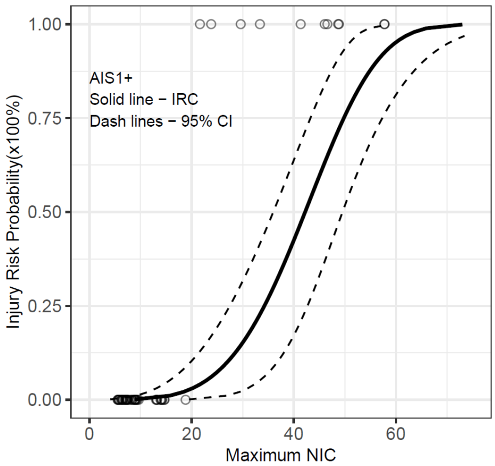

Fig. 2. IRC for neck injury criterion (NIC)

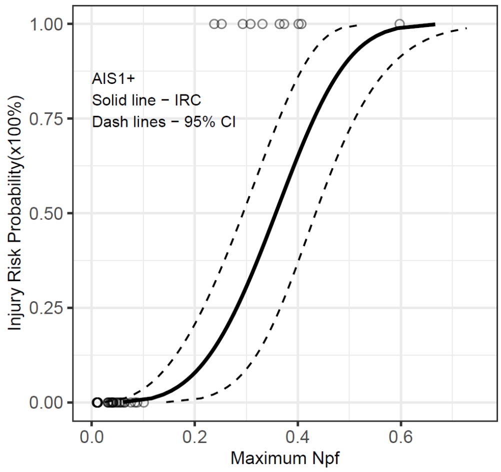

Fig. 3. IRC for neck posterior shear and flexion (Npf).

The IRCs for the lower neck posterior shear (Fx+) and extension moment (My-) are shown in Fig. 7–8. The lower neck posterior shear force (Fx+) has IARVs of 143.8 N, 313.2 N, and 465.8 N for 5%, 25%, and 50% injury risk, respectively, while the lower neck extension moment (My-) has IARVs of 7.62 Nm, 11.01 Nm, and 13.29 Nm for 5%, 25%, and 50% injury risk, respectively.

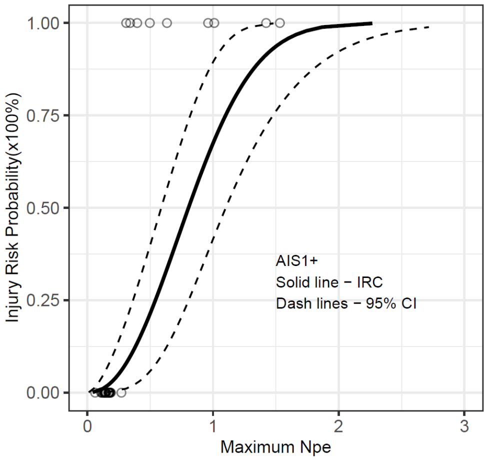

Fig. 4. IRC for neck posterior shear and extension (Npe).

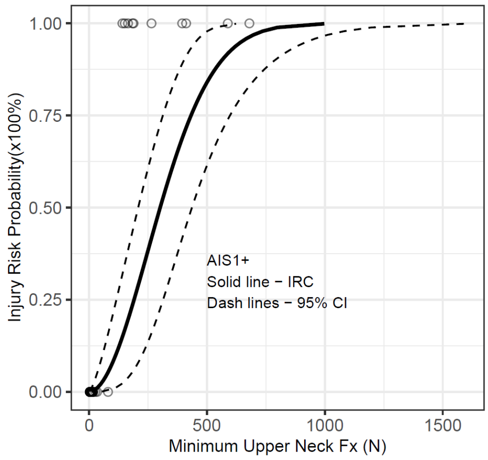

Fig. 5. IRC for upper neck anterior forward shear Fx.

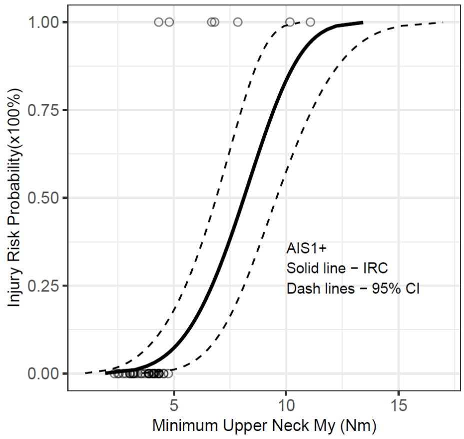

Fig. 6. IRC for upper neck extension moment My.

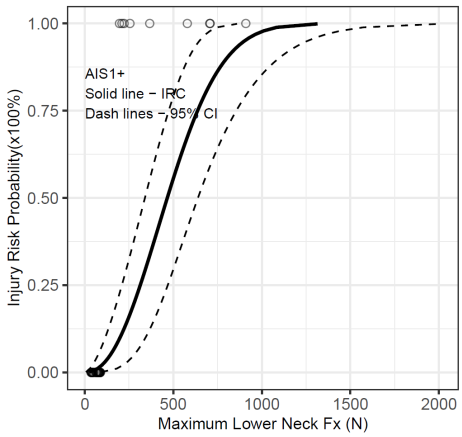

Fig. 7. IRC for lower neck posterior shear Fx.

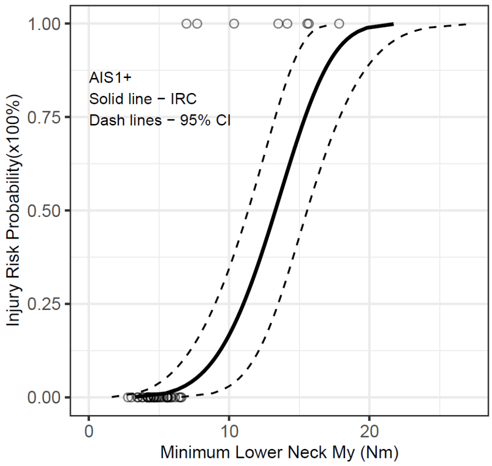

Fig. 8. IRC for lower neck extension moment My.

The IRCs for T1 accelerations are shown in Fig. 9 – 10. The T1 rearward acceleration (Ax-) has IARVs of 7.4 g, 10.7 g, and 12.9 g for 5%, 25%, and 50% injury risk, respectively. The T1 upward acceleration (Az-) IARVs are 5.5 g, 8.5 g, and 10.6 g for 5%, 25%, and 50% injury risk, respectively.

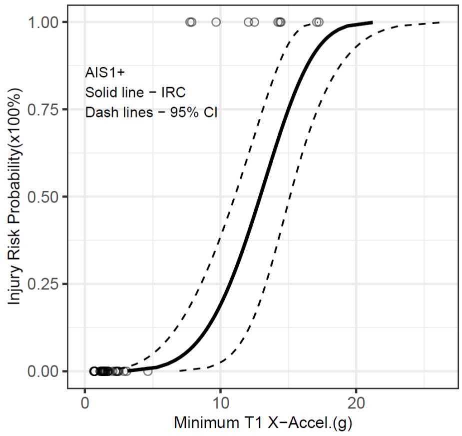

Fig. 9. IRC for T1 rearward acceleration Ax.

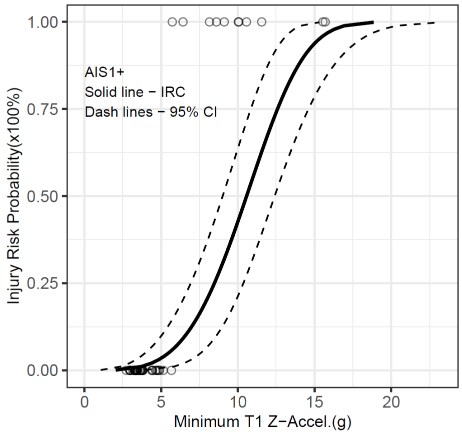

Fig. 10. IRC for T1 upward acceleration Az.

Results of the Second Group of Indicators

The IRCs and IARVs of head acceleration, head and T1 displacements, y-rotation, y-angular velocity and head-to-T1 relative displacements and rotations are summarized in Table IV. The IRC plots for the predictors outlined in Table IV are shown in Figure A13-24 in the appendix.

TABLE IV
SUMMARY OF HEAD AND HEAD-TO-T1 IRCS AND IARVS

Location	Description	Predictor	Shape	Scale	AIC	GKG	AU ROC	Qual. Index	IARVs		
									5%	25%	50%
Head	Head x-accel. min	Ax (-)	3.7978	9.3156	50	1.00	1.00	0.20	4.3	6.7	8.5
	Head x-accel. max	Ax (+)	1.6459	43.8815	106	1.00	1.00	0.38	7.2	20.6	35.1
	Head z-accel. min	Az (+)	3.7148	8.0422	44	0.88	0.94	0.19	3.6	5.8	7.3
	Head z-accel. max	Az (-)	2.9020	19.6673	71	1.00	1.00	0.24	7.1	12.8	17.3
	Head x-displ. min	Dx (-)	1.0287	272.9854	31	0.45	0.73	2.33	15.2	81.3	191.2
	Head x-displ. max	Dx (+)	5.0560	379.6040	111	0.99	0.99	0.14	211	297	353
	Head x-displ. min	Dz (-)	4.2869	109.9216	99	1.00	1.00	0.16	55.0	82.2	100.9
	Head z-displ. max	Dz(+)	2.1339	27.6576	69	0.94	0.97	0.34	6.9	15.4	23.3
	Head y-rotation min	Ry (-)	1.3385	13.6156	74	1.00	1.00	0.50	1.5	5.4	10.4
	Head y-rotation max	Ry (+)	7.4297	82.1795	66	0.82	0.91	0.10	55.1	69.5	78.2
	Head y-ang. Vel. min	ωx (-)	4.5972	1405.8294	133	1.00	1.00	0.17	737	1072	1298
	Head y-ang. Vel. max	ωx (+)	6.1280	1866.2151	107	0.99	0.99	0.14	1149	1523	1758
Head to T1 Displacement	x-displacement min	NDCx(-)	4.2920	201.3855	61	-0.22	0.39	0.37	101	151	185
	x-displacement max	NDCx(+)	1.6392	46.0927	95	1.00	1.00	0.46	7.5	21.6	36.9
	z-displacement min	NDCz(-)	0.7147	7.2247	12	0.15	0.58	3.61	0.1	1.3	4.3
	z-displacement max	NDCz(+)	5.2055	138.1173	80	1.00	1.00	0.16	78.1	108.7	128.7
	y-rotation min	NDCr(-)	3.2611	15.3787	52	1.00	1.00	0.24	6.2	10.5	13.7
	y-rotation max	NDCr(+)	1.7581	77.8173	131	0.16	0.58	0.42	14.4	38.3	63.2

DISCUSSION

Whiplash injury prediction is a challenging task due to the difficulty in accurately detecting physical injuries. The field data rely heavily on clinical data, which are not supported by objective diagnostic tools such as CT scans, MRIs, and other imaging methods, but instead are based on patients' subjective feelings. Quantifying injury severity is challenging, and whiplash injuries typically use AIS 1+ as the threshold for analysis. This research is the first to explore a method for creating whiplash IRCs(for low-to-moderate speeds) using techniques commonly employed for high-severity injuries in frontal, side, or oblique MVCs. While many challenges were encountered during the analysis, several interesting findings were discovered.

In the data analysis, ISO TS18506 was followed [28]. One of the steps involves removing highly influential points. The technique used for this analysis is detecting influential points in regression using dfbetas, with a threshold of $2/\sqrt{n}$, where n is the sample size [29]. An exercise was conducted to compare the analysis with and without this step. The IRC for upper neck tension was used as an example, shown in Fig. 11. It was observed that the IARV at 50% injury risk approximately doubled after removing the highly influential points. This step is critical in IRC development to ensure accurate statistical results.

From this research, most parameters used in NCAP and IIHS have high values for GKG and AUROC (most are close to 1.0), except for Naf (anterior shear and flexion), Nae (anterior shear and extension), and upper neck posterior shear force (Fx). The Naf and Nae have GKG values of -0.37 and -0.31, and AUROC values of 0.32 and 0.35, respectively, indicating poor correlation for injury predictions. The upper neck posterior shear force (Fx) has a GKG of 0.04 and an AUROC of 0.52, also indicating poor correlation. It was observed that the upper neck anterior shear force (Fx), which is not specified in NCAP test requirements, has both GKG and AUROC values of 1.0, indicating perfect prediction between the model and the data samples used to create the IRC. From engineering common sense, the posterior shear force was used as an injury predictor due to the rear motion of the head in whiplash crash, although the finding in this study with statistical analysis did not show it was strong indicator for injury prediction. In addition, the lower neck posterior shear force showed as a strong indicator for injury prediction. These showed the complexity of the whiplash neck injury mechanism. More research is needed to understand the mystery. Regardless, based on the result in this study, it appears that the upper neck anterior shear force (Fx) should be considered as well in seat design.

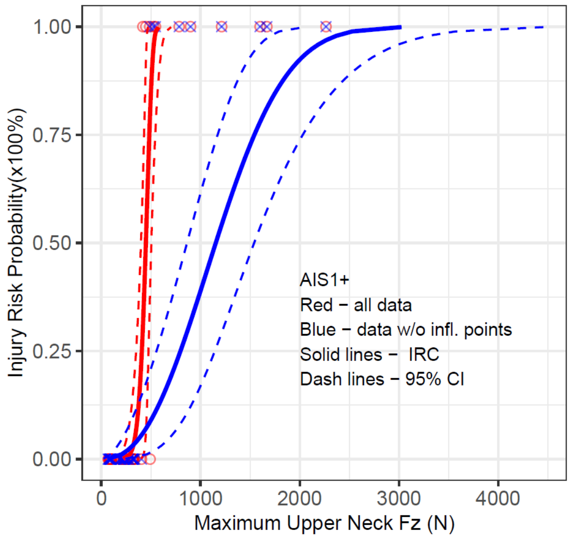

Fig. 11. IRCs for upper neck tension with and without the influential points

It was noticed that several indicators, for example, NIC, Npf and Npe have GKG and AUROC values at 1.0. Statistically, the GKG and AUROC value at 1.0 indicate perfect prediction of the existing data set. This could be attributed to the separation between injuries and non-injurious cases. The volunteer test data typically are at low speed (7 km/h), well below the injurious threshold for safety reason, while the PMHS test could be on the high side above AIS1+ to be identified in autopsy.

It was estimated that female whiplash injuries have approximately a 20% higher risk than male injuries [14] and it was recommended that EvaRID injury thresholds should be about 20% lower than those recommended for BioRID [12]. The Euro NCAP assessment protocol (version 9.3, December 5, 2023) specifies the thresholds for BioRID. These BioRID thresholds were scaled down by 20% to determine the target EvaRID thresholds. The injury risks were then calculated based on the scaled injury thresholds for EvaRID. The results are summarized in Table V. The values represent the scaled thresholds for EvaRID, and the injury risks (in percentage) in parentheses were calculated from the IRCs in this study.

It was observed that the upper neck posterior shear force (Fx +ve) has a 100% injury risk at lower and capping limits of 152 N and 232 N, respectively. As discussed earlier in this section, Fx (+ve) does not show a strong correlation with injuries in this analysis for EvaRID. Instead, Fx (-ve) appears to be a much better indicator for injury prediction, with a 47.0% injury risk at the scaled capping limit of 288 N. The upper neck extension and flexion (Myoc), lower neck flexion and extension (My), lower neck shear force (absolute values), and Nkm all show injury risks at or near 100%. These high injury risk values indicate that more work is needed to improve seat design and reduce female whiplash injuries.

TABLE V
COMPARISON OF SCALED EVARID INJURY THRESHOLDS AND THEIR CORRESPONDING INJURY RISKS CALCULATED FROM THIS STUDY

Limit	Medium Severity Pulse Higher Limit	Medium Severity Pulse Lower Limit	Medium Severity Pulse Capping Limit	High Severity Pulse Higher Limit	High Severity Pulse Lower Limit	High Severity Pulse Capping Limit
NIC	8.8 (0.1%)	19.2 (2.5%)	21.6 (4.1%)	10.4 (0.2%)	18.4 (2.1%)	20.4 (3.2%)
Nkm			0.552 (96.8%)			0.624 (99.6%)
Upper Neck Shear Fx (+ve) (N)	24 (0%)	152 (100%)	232 (100%)	24 (0%)	168 (100%)	291.2 (100%)
Upper Neck Shear Fx (-ve) (N)			288 (47.0%)			288 (47.0%)
Upper Neck Tension Fz (N)	288 (2.5%)	600 (13.6%)	720 (20.2%)	376 (4.7%)	616 (14.4%)	819.2 (26.4%)
Upper Neck Extension Myoc (Nm)			24 (100.0%)			24 (100.0%)
Upper Neck Flexion Myoc (Nm)			24 (100%)			24 (100%)
Lower Neck Shear Fx (ABS) (N)			288 (99.7%)			288 (99.7%)
Lower Neck Extension My (N)			24 (100%)			24 (100%)
Lower Neck Flexion My (N)			24 (100%)			24 (100%)
T1 acceleration (g)			12.44 (4.1%)			14.24 (6.8%)

There are a few limitations in this study. First, there is a lack of published test data from female volunteers and PMHS. As a result, male data were used in the analysis. However, females have a higher injury risk than males in whiplash injuries [12][14]. Therefore, the injury risk may be underestimated due to the use of male data in this analysis. Additionally, no scaling was carried out for male samples due to the inaccuracy of scaling between male and female data. The volunteers for the non-injury's cases are young adults and the female PMHS specimen are elderly. The data set is not sufficient to derive criteria for age related whiplash injury prediction.

All male PMHS studies resulted in either non-injury [18] or injury outcomes [21,22,30] except for the Mertz study (2 male PMHS), in which one PMHS (69yo, 170.2cm, and 58.9kg male) did not sustain neck injury, while the other PMHS (66yo, 162.6cm, and 60.8kg male) sustained minor ligamentous damage. Since the height and weight of the injurious PMHS from the Mertz study were very close to EvaRID (161.8cm and 62.3kg for EvaRID vs. 162.6cm and 60.8kg for Mertz injurious PMHS), the high severity load case (24km/h) was considered as an injurious EvaRID load case. However, caution should be noted for this approach, as the other male PMHS did not sustain any injuries in the Mertz study. It should also be noted that the PMHS were tested multiple times in the Mertz study so that the injury outcome might be due to damage accumulation. More research should be conducted to generate female biomechanical data and injuries in low-speed rear impacts and not to use male data for female IRC development.

EvaRID data were not scaled to male volunteers and PMHS. The current scaling techniques for female-to-male or male-to-female scaling only consider subject mass and anthropometric length ratios in the scaling equations. However, the female neck has different anatomy, geometry, physiology, range of motion, strength, material and structural properties from the male neck. Therefore, the simple mass and length ratios used in the scaling equations cannot account for differences in neck responses between females and males. For this reason, the female EvaRID data were not scaled to any male subject data. None of the male PMHS studies resulted in mixed no injury and injury outcomes except for the Mertz study (one PMHS sustained a minor ligamentous injury while the other PMHS did not sustain injury). EvaRID was simulated in the same conditions as male volunteer and PMHS studies and not scaled to individual male subjects under the assumption that non-injurious male volunteer tests were also non-injurious female tests for EvaRID simulations. Previous volunteer studies [3,7] investigated female volunteer responses in low-speed rear impacts and compared them to male volunteers at the same severity. Both male and female volunteers did not sustain injuries up to ΔV of 8km/h. All male volunteer load cases used for EvaRID FE simulations were less than 8km/h, so the EvaRID data were treated as no injury data point. For the injurious male load cases, it was assumed that females would sustain injuries at the same severity and in the same test condition in which the male PMHS sustained injuries. This assumption was based on the field injury data showing that female has a higher risk of neck injuries than males in rear-ended crashes [8, 10]. However, caution should be made when the IRCs are used due to the assumptions made in this approach. More female-specific biomechanical responses in low-speed rear impacts should be generated to avoid this assumption and improve EvaRID IRCs.

CONCLUSIONS

Injury risk curves were generated for EvaRID, a 50th percentile female dummy scaled from the BioRID design concept, to predict average female whiplash injury risks using the method defined in ISO TS 18506. The injury predictors used in Euro NCAP and their scaled thresholds were applied to estimate the injury risk at these thresholds. It was found that most of the predictors were statistically significant for injury prediction, except for Nae, Naf, and upper neck posterior shear force (Fx), which all had very low GKG and AUROC values, indicating weak correlation with injuries. It is recommended that the upper neck posterior shear force (Fx) may need to be replaced with the anterior shear force (Fx) as a requirement in seat designs. The injury risk calculated from the IRCs in this study for the scaled injury thresholds for EvaRID was very high for a few parameters, including upper neck flexion and extension (Myoc), lower neck flexion and extension (My), and upper neck shear force (Fx), all of which had injury risks at or near 100%. This suggests that some rating thresholds could be revisited and optimized to maximize the benefit for occupant protection in vehicle development.

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APPENDIX

The details of the specimen, test condition, seat type, sex, and delta speed for each load case are documented in Table A1. All male specimens tested under the same test condition are counted as one in this analysis.

TABLE A1
SUMMARY OF THE SPECIMEN AND TEST CONDITIONS USED IN THIS RESEARCH

Reference	Specimen No.	Subject	Seat Type	Sex	Delta Speed (km/h)	MAIS
Mertz et al. 1967 [15]	2 males	PMHS	rigid	M	16	0
	LMP	Volunteer	rigid	M	16	0
	2 males	PMHS	rigid	M	24	1
Yoganandan et al. 2000[19]	101	PMHS	rigid	F	24.8	2
	102	PMHS	rigid	F	14.9	0
	103	PMHS	rigid	F	15.9	2
	104	PMHS	rigid	M	23.7	2
	105	PMHS	rigid	F	24.5	2
Bertholon et al. 2000 [18]	9 males	PMHS	rigid	M	10.8	0
Davidsson et al. 1998 [17]	7 males	Volunteer	V	M	6.8	0
Carlsson et al. 2021 [3]	A	Volunteer	V	F	7.0	0
	B	Volunteer	V	F	6.6	0
	C	Volunteer	V	F	6.7	0
	D	Volunteer	V	F	6.7	0
	E	Volunteer	V	F	6.9	0
	F	Volunteer	V	F	6.9	0
	G	Volunteer	V	F	6.7	0
	H	Volunteer	V	F	6.9	0
	A	Volunteer	V	F	6.9	0
	B	Volunteer	V	F	6.8	0
	C	Volunteer	V	F	6.9	0
	D	Volunteer	V	F	6.9	0
	E	Volunteer	V	F	6.7	0
	F	Volunteer	V	F	6.9	0
	G	Volunteer	V	F	6.9	0
	H	Volunteer	V	F	6.9	0
Ono et al. 2006 [9]	I	Volunteer	Rigid	F	6.0	0
	II	Volunteer	Rigid	F	6.0	0
	4 males	Volunteer	Rigid	M	6.0	0

Reference	Specimen No.	Subject	Seat Type	Sex	Delta Speed (km/h)	MAIS
Sato et al. 2014 [16]	F-I	Volunteer	Rigid	F	8.1	0
	F-II	Volunteer	Rigid	F	8.1	0
	F-III	Volunteer	Rigid	F	8.1	0
	F-IV	Volunteer	Rigid	F	8.1	0
	F-V	Volunteer	Rigid	F	8.1	0
	F-VI	Volunteer	Rigid	F	8.1	0
	F-VII	Volunteer	Rigid	F	8.1	0
	F-VIII	Volunteer	Rigid	F	8.1	0
	12 males	Volunteer	Rigid	M	8.1	0
Roberts et al. 2002 [20]	10 males	Volunteer	Experimental Seat	M	7.0	0
Kang et al. 2012 [30]	7 males	PMHS	rigid	M	17	1
	7 males	PMHS	rigid	M	24	1
Kang et al. 2014 [21]	2 males	PMHS	Seat B	M	17	2
	1 male	PMHS	Seat B	M	24.7	2
Kang et al. 2021 [22]	3 males	PMHS	2018 Honda Accord	M	24	3
	3 males	PMHS	2018 Honda Accord	M	16	3
	3 males	PMHS	2018 Honda Accord	M	18.2	3

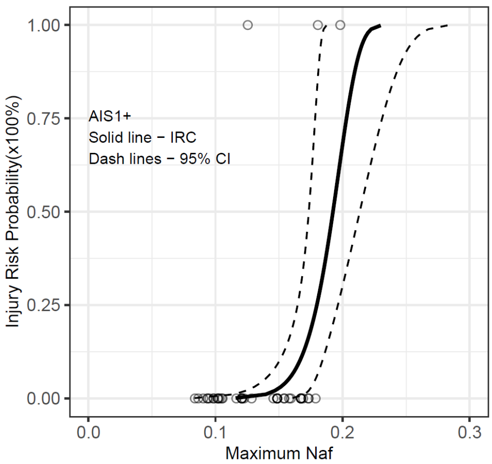


Figure A1. IRC for neck anterior shear and flexion (Naf)

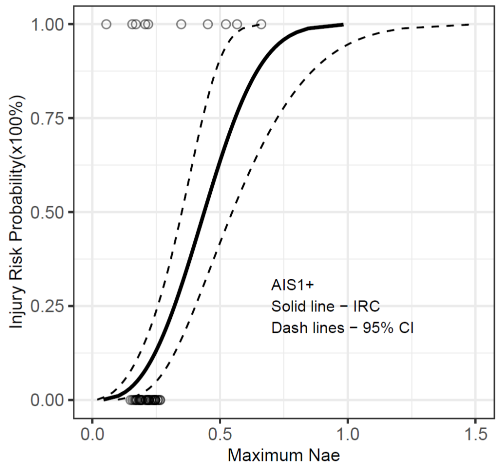
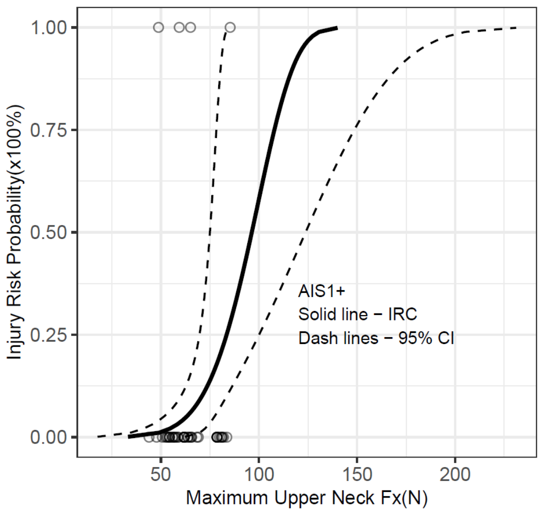


Figure A2. IRC for neck anterior shear and extension (Nae).



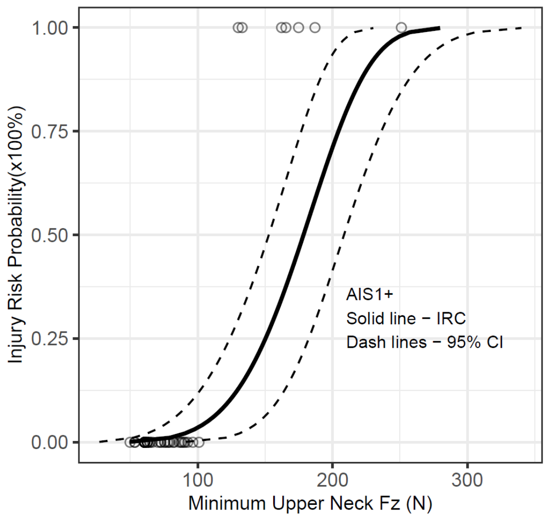
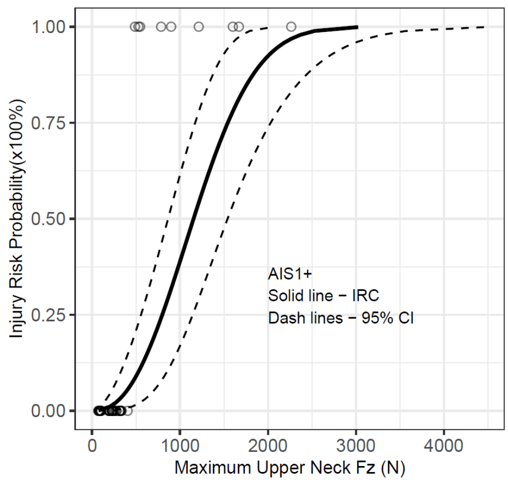


Figure A3. IRC for upper neck posterior shear Fx

Figure A4. IRC for upper neck compression force Fz.



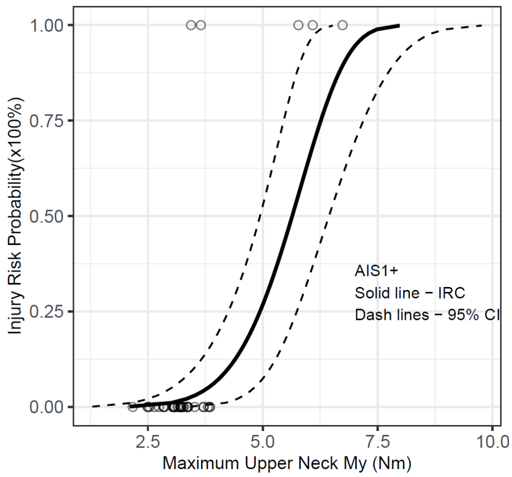
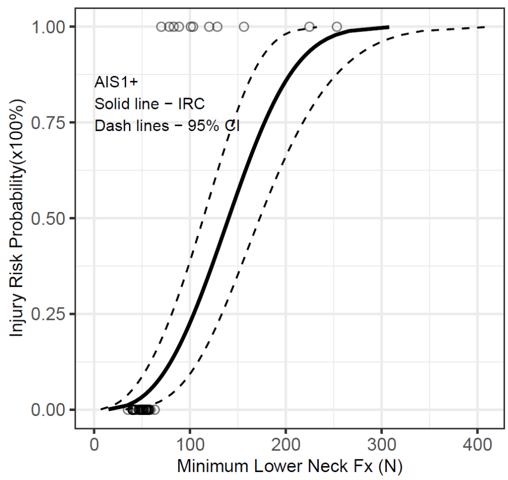


Figure A5. IRC for upper neck tension force Fz.

Figure A6. IRC for upper neck flexion moment My.



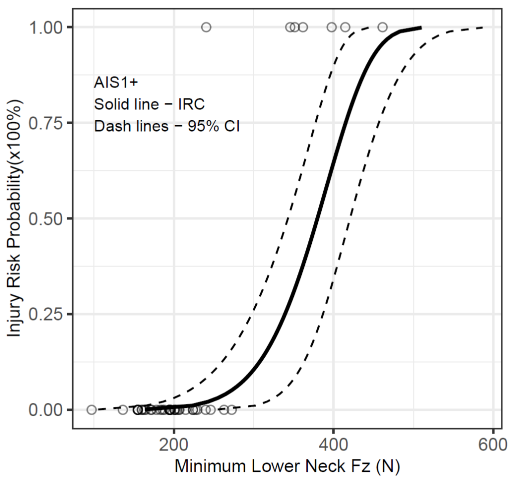
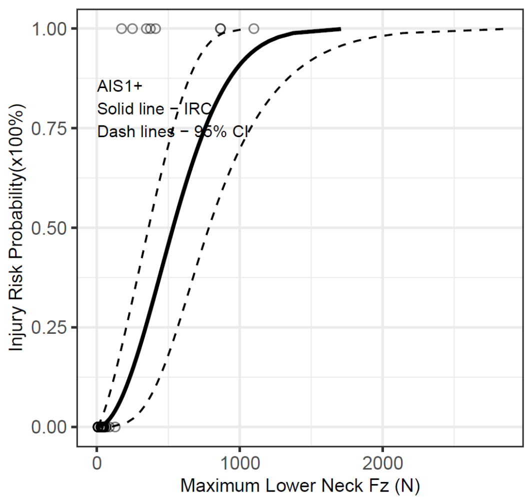


Figure A7. IRC for lower neck anterior shear Fx.

Figure A8. IRC for lower neck compression Fz.



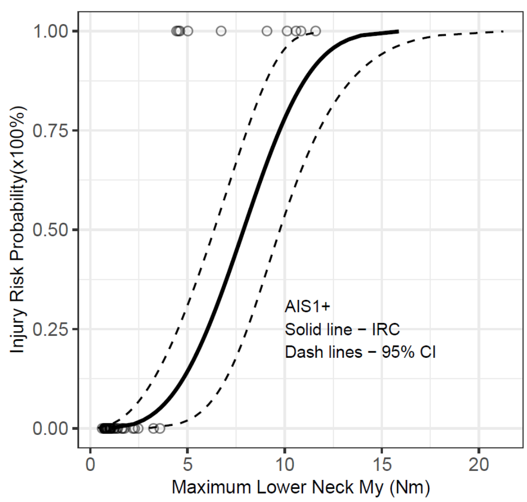


Figure A9. IRC for lower neck tension Fz.

Figure A10. IRC for lower flexion moment My.

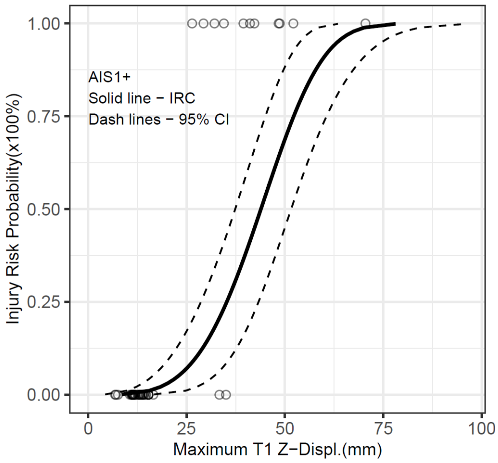


Figure A11. IRC for T1 downward acceleration Az.

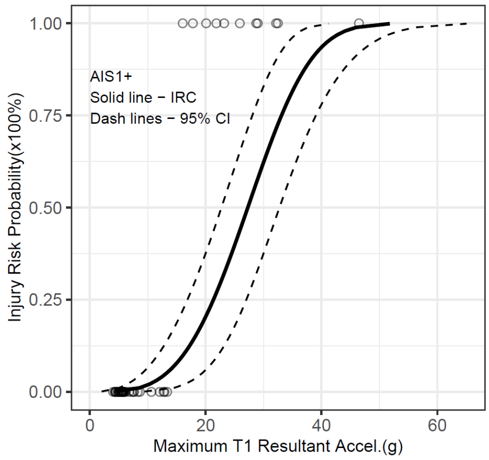


Figure A 12. IRC for T1 resultant acceleration.

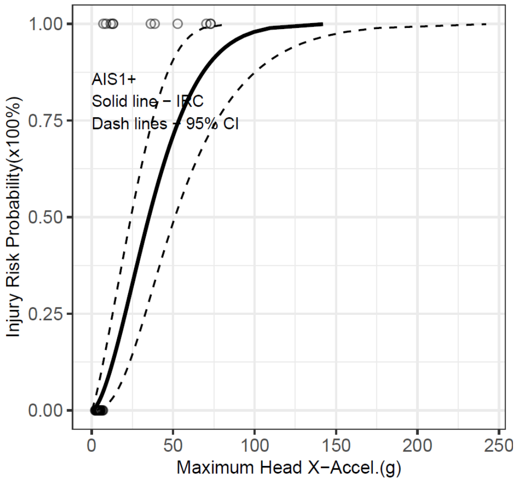


Figure A13. IRC of head forward acceleration.

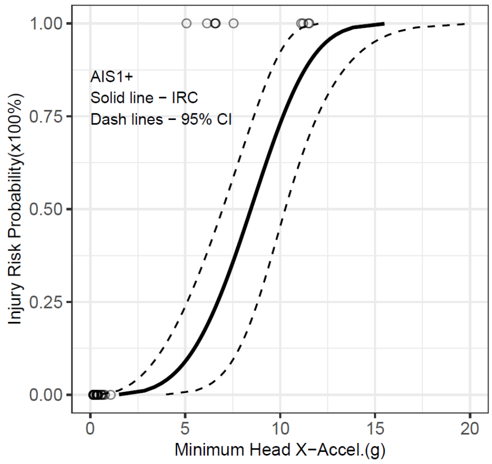


Figure A14. IRC of head rearward acceleration.

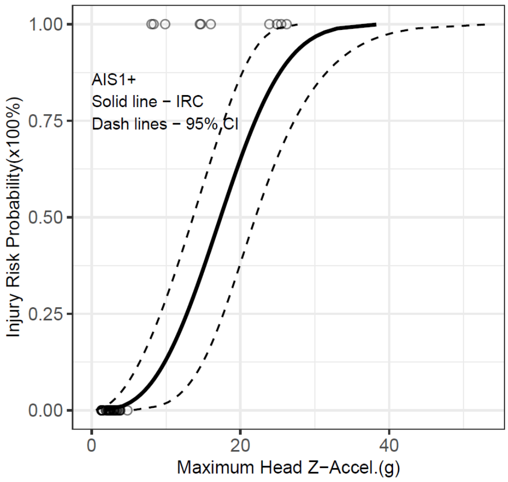


Figure A15. IRC of head downward acceleration.

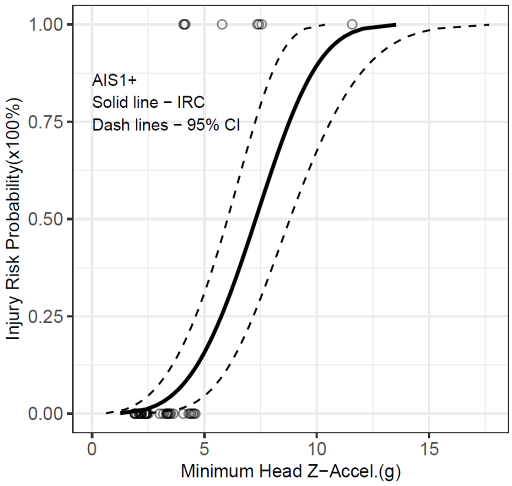


Figure A16. IRC of head upward acceleration.

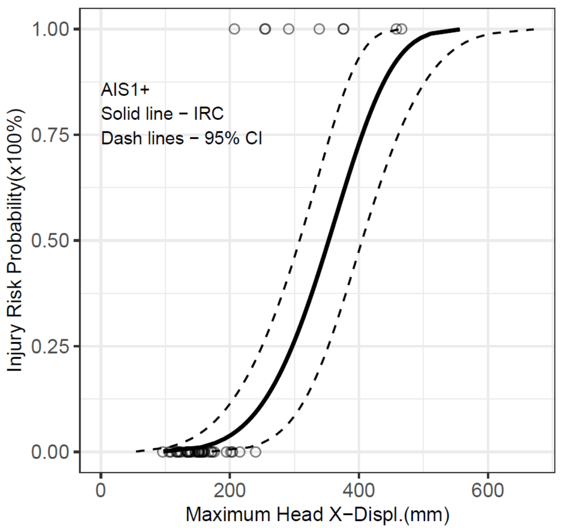


Figure A17. IRC of forward head x-displacement.

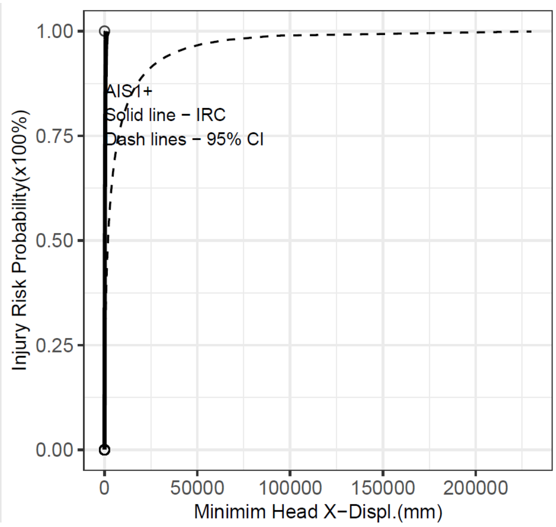


Figure A18. IRC of rearward head x-displacement.

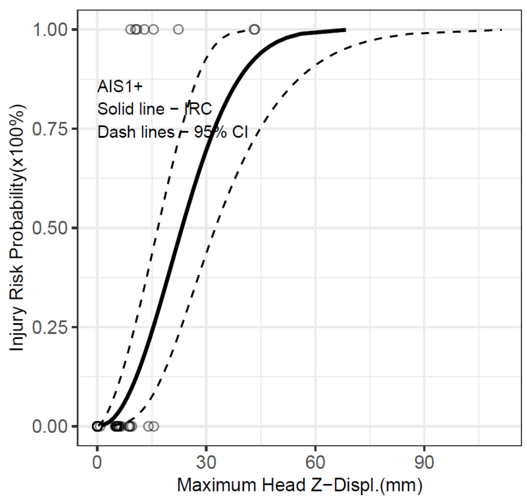


Figure A19. IRC of head downward z-displacement.

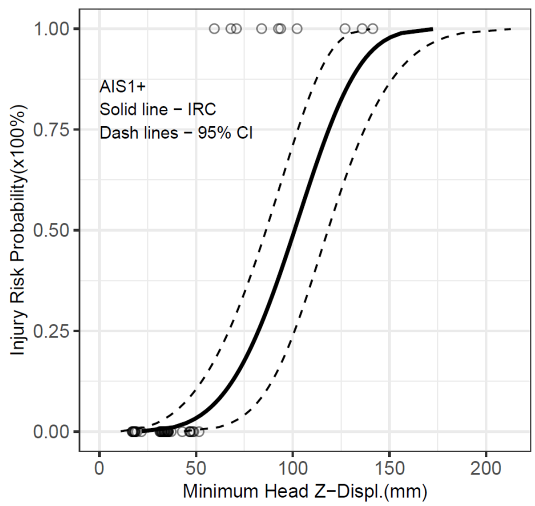


Figure A20. IRC of head upward z-displacement.

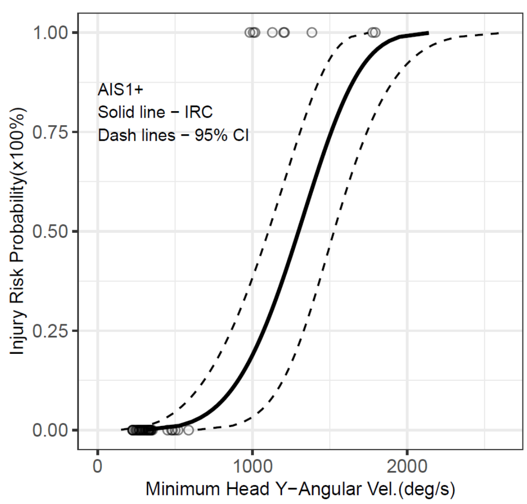


Figure A21. IRC of head forward y-angular velocity.

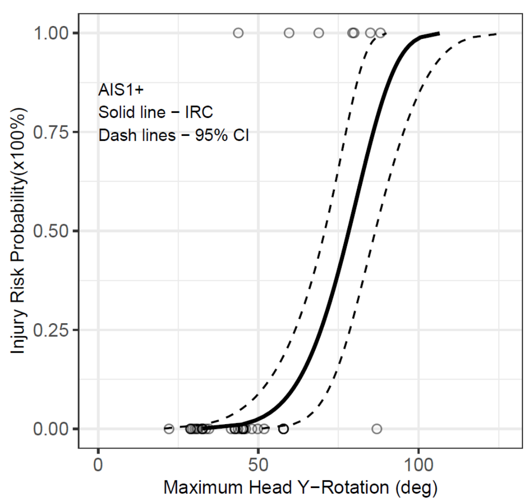


Figure A22. IRC of head rearward y-rotation.

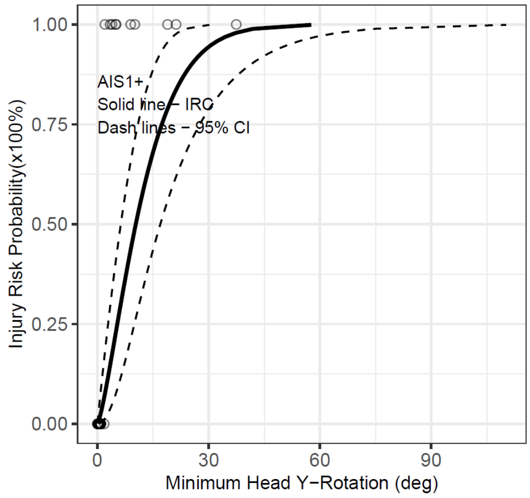


Figure A23. IRC of head forward rotation.

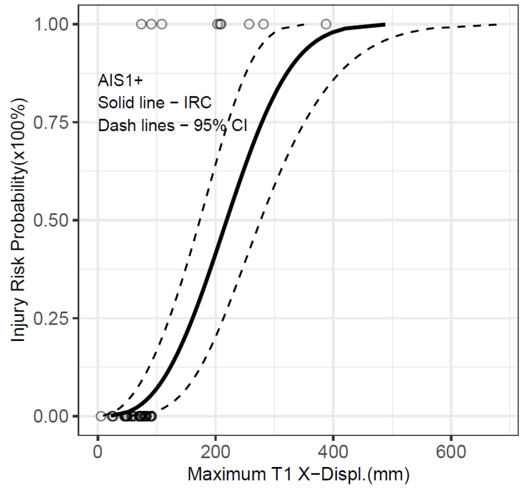


Figure A24. IRC of T1 forward displacement.

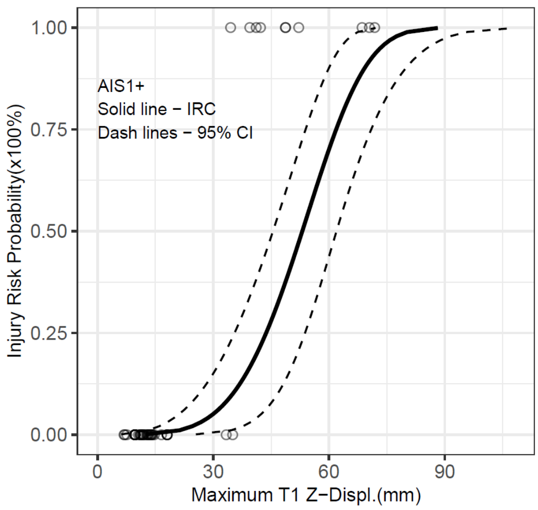


Figure A25. IRC of T1 downward displacement.

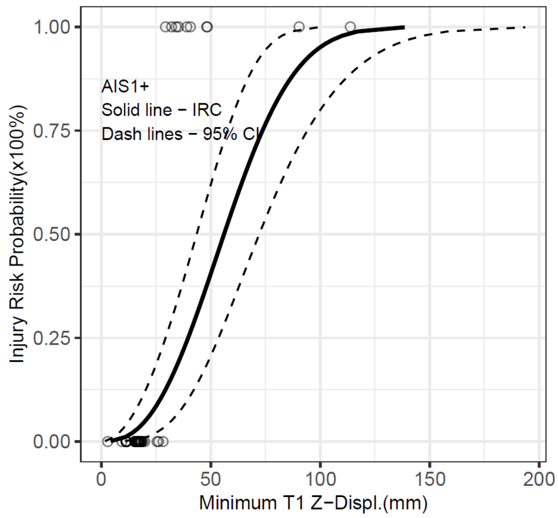


Figure A26. IRC of T1 upward displacement

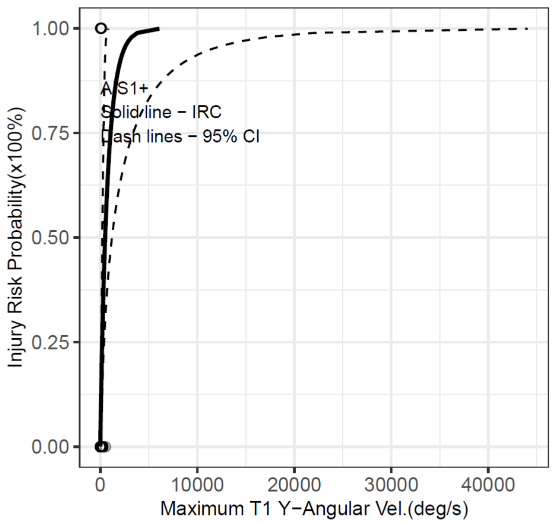


Figure A27. IRC of T1 rearward y-angular velocity.

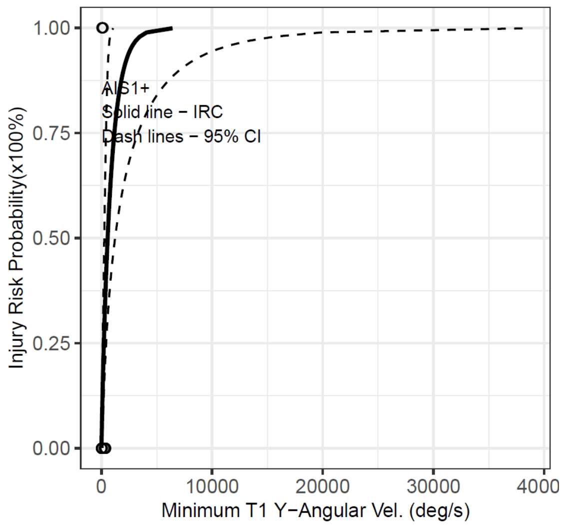


Figure A28. IRC of T1 forward y-angular velocity.

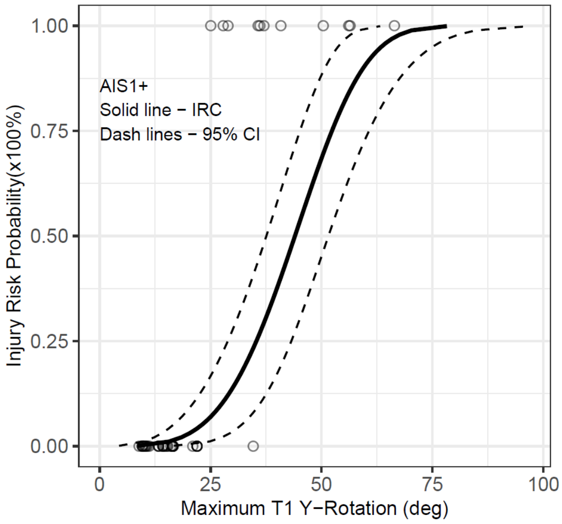


Figure A29. IRC of T1 rearward y-rotation.

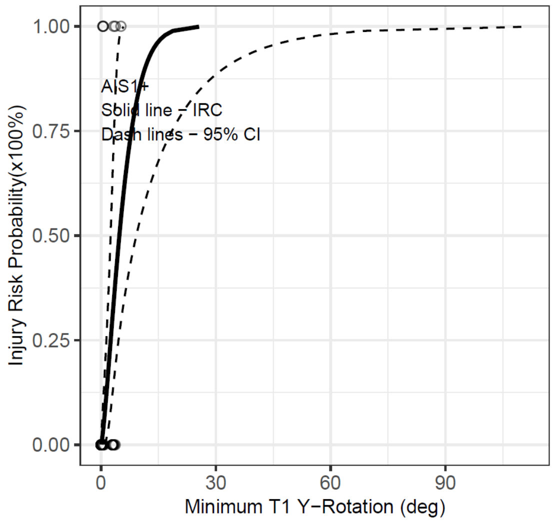


Figure A30. IRC of T1 forward y-rotation.

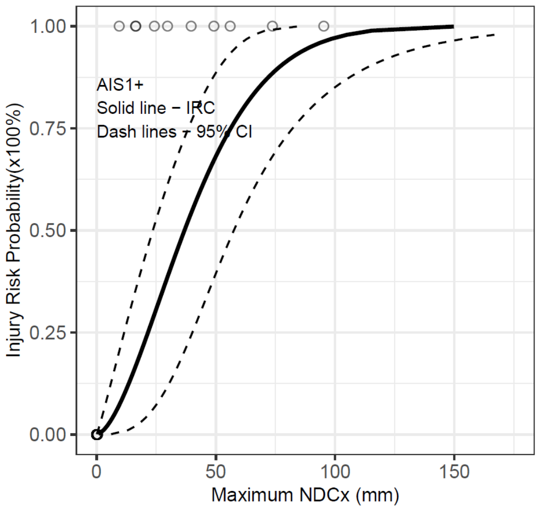


Figure A31. IRC of head-to-T1 forward displacement (NDCx).

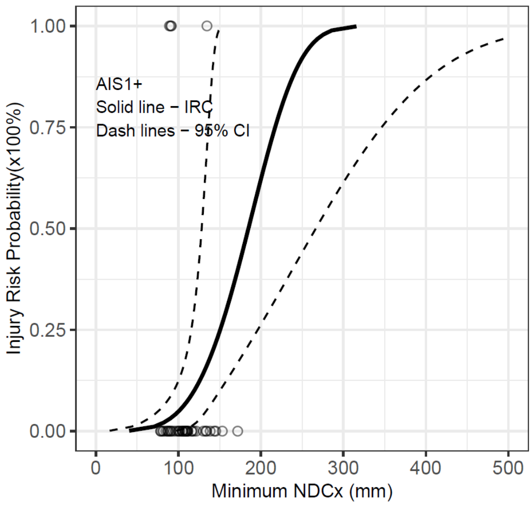


Figure A32. IRC of head-to-T1 rearward displacement (NDCx).

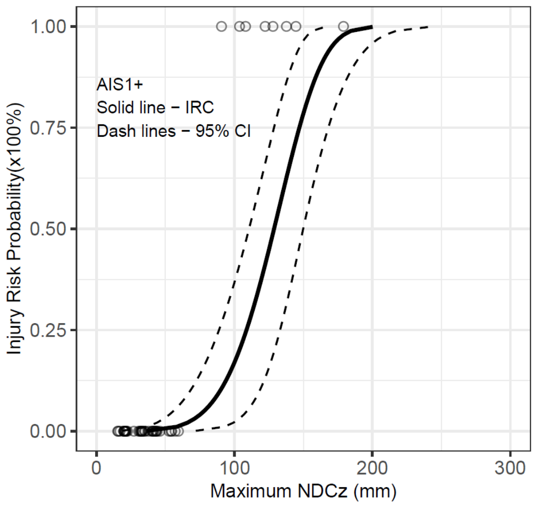


Figure A33. IRC of head-to-T1 downward z-displacement (NDCz).

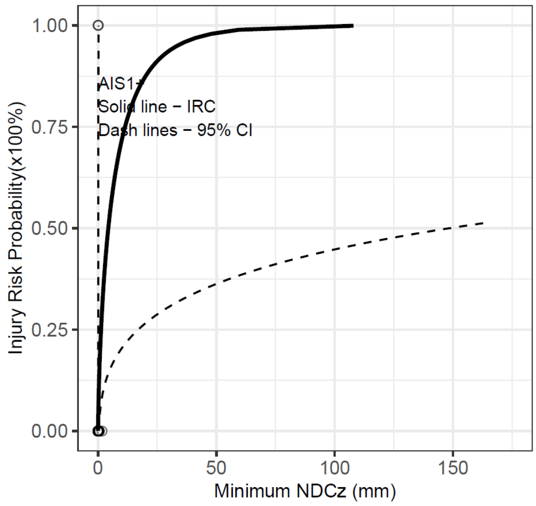


Figure A34. IRC of head-to-T1 upward z-displacement (NDCz).

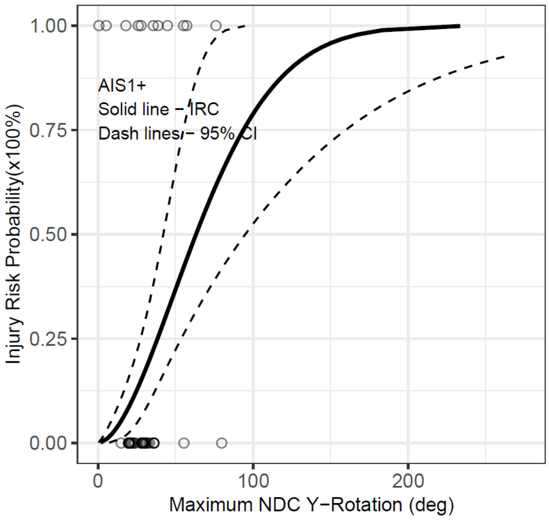


Figure A35. IRC of head-to-T1 rearward y-resultant rotation (NDCr)

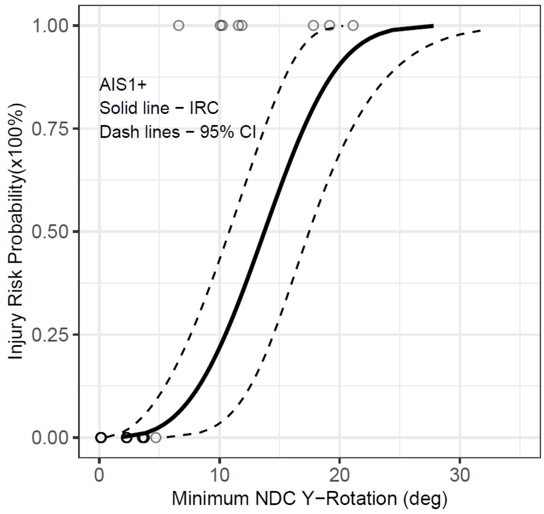


Figure A 36. IRC of head-to-T1 forward resultant y-rotation (NDCr)