COMPUTERIZED CRASH RECONSTRUCTION OF REAL WORLD CRASHES USING OPTIMIZATION METHODOLOGY

Vikas Hasija
GESAC, Inc
Erik G. Takhounts
Stephen A. Ridella
NHTSA
United States
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ABSTRACT

Computerized crash reconstruction of real world crashes involves dealing with a lot of unknown parameters and as such the reconstruction problem cannot be solved deterministically as was shown using a parametric methodology presented in our previous ESV paper titled "Computational Analysis of Real World Crashes: A Basis for Accident Reconstruction Methodology." This paper introduces a modified version of the parametric methodology, which involves using an optimization scheme to derive an optimal solution for the reconstruction problem in a given range of unknown parameters. Real world crashes were selected from the CIREN database and were solved using the proposed methodology. Human-Vehicle-Environment (HVE) software was used to generate the crash pulse where EDR data were missing. The problem was set up in MADYMO. During the set up, the unknown parameters were identified. ModeFRONTIER software was used for optimization. The identified unknown parameters were treated as design variables. The objective function and the constraints were defined such that they minimize the differences in injuries and occupant-vehicle contacts between the real world data and the model prediction. Since the objective function has a great effect on the final solution, a normalized form of the objective function, weighted based on the AIS level of the injuries sustained by the occupant, was formed in this study. A genetic algorithm with Sobol DOE (Design of Experiments) was used for optimization. Results of the simulations showed that the optimal solution correctly predicted both the occupant-vehicle contacts and the injuries sustained by the occupant. By viewing the occupant motion inside the vehicle during the crash, better occupant protection systems can be devised. Correlation studies were also carried out to find the critical parameters affecting the solution. In addition, a best case scenario study was carried out to find, using optimization, the design changes that could help mitigate all or some of the injuries sustained by the occupant.

INTRODUCTION

Computerized crash reconstruction is carried out to investigate crash sequences and to study occupant kinematics during crashes. Occupant kinematics can then be used to design better and more efficient safety systems for occupant protection. There are a lot of parameters that affect an occupant's kinematics and injury risk. Due to lack of information on these parameters, the reconstruction cannot be carried out accurately. Since assumptions have to be made for these unknown parameters, it is imperative not to predict the model outcome using just one set of parameter values as different sets of parameter values within the range can lead to quite different injury predictions for the same case, as was shown in Hasija et al [1]. In the past, injury evaluation based on reconstruction has been carried out using just one set of parameters. For example, Mardoux et al [2] and Franklyn et al [3] presented papers where computational models used to predict injuries were driven using data obtained from physical tests. The experiments can have errors associated with them that can lead to errors in the model's injury predictions. The effect of these uncertainties was not analyzed.

Optimization, which refers to the study of problems in which one seeks to minimize or maximize a function by systematically choosing the values of variables from within an allowed set, while satisfying the constraints, is becoming more popular for carrying out crash reconstruction analysis. For example, Untaroiu et al [4] presented a paper that was used to investigate the application of optimization techniques to the field of crash reconstructions of pedestrian accidents. In their papers, it was shown, using a mock or "ideal" crash reconstruction problem, that optimization algorithms combined with an appropriate objective function have the capability to identify accurately the pre-impact conditions of the pedestrian and vehicle. The pre-impact parameters of pedestrian and vehicle models were treated as unknown design variables. In

addition, optimization methodology was successfully applied to reconstruction of a real-world pedestrian crash. Also, Shen et al [5] presented a paper evaluating optimization-based method for reconstructing pedestrian-vehicle accident and testing its performance. By reconstructing two real-world pedestrian collisions, they concluded that optimization is very effective in finding a optimum solution, which not only reduces the number of cycles, but also saves manual operation.

The objective of this paper is to present a methodology that utilizes an optimization scheme to come up with the best possible solution to a reconstruction problem, which has a lot of unknown variables that cannot be fixed to a particular value but are defined by a range. The best possible solution helps provide the occupant kinematics during crash. The methodology is shown by reconstructing two real world crash cases selected from the CIREN database [6]. These cases include: a) a "moderate brain injury" case, and b) a "severe brain injury" case. The unknown parameters are treated as design variables. The occupant-vehicle contacts and the injuries sustained by the occupant as listed in CIREN are used to set up the objective function and constraints for optimization. The best solution obtained for the reconstruction problem from optimization is one that matches all the occupant-vehicle contacts and injuries sustained by the occupant. The paper also shows how crirtical parameters are identified and how a best case scenario (minimum injury or no-injury) for a given case can be obtained.

METHODOLOGY

The methodology introduced in this paper (Figure 1) for reconstructing real world crashes uses an optimization technique to find the best possible solution to the reconstruction problem.

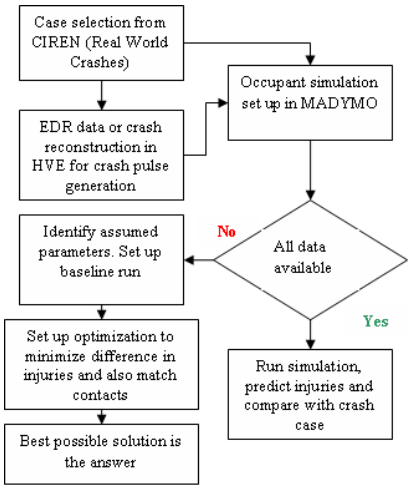


Figure 1. Optimization Methodology.

The methodology starts with the selection of a real world crash case from CIREN. Following the case selection, the Event Data Recorder (EDR) information available for the case is searched in CIREN to get the crash pulse. If the selected case has an EDR pulse available, the pulse is directly used for occupant simulation. If EDR pulse is not available, the crash details available from the case are used in HVE [7] to generate the crash pulse. Next, the occupant simulation is set up in MADYMO [8] based on the case information available from CIREN such as occupant information, restraints information etc. The EDR crash pulse or the HVE-generated crash pulse is used to drive the MADYMO model. During this set up, the data availability is checked to see if all parameters required for reconstruction are available. If all data is available, the occupant simulation set up in MADYMO can be run and the results can be compared with the real world data. But most of the time, all data required for reconstruction is not available. These unknown parameters are identified during the MADYMO set up stage and assumptions are made for these unknown parameters. Once the MADYMO model is set up, a baseline run is obtained by matching the occupant-vehicle contacts listed in CIREN for the particular crash case. Using these assumed parameters as design variables, the optimization is set up around the baseline run in modeFrontier software [9]. The range for these selected design variables is defined, the objective function is set up, and the constraints are defined. Two types of constraints are defined i.e. contacts based constraints and injury based constraints. The constraints and objective functions are defined such that they minimize the differences in injuries and occupant-vehicle contacts between the real world data and the model prediction. The solution obtained from optimization takes into account the variation in the assumed parameters, and thus gives the best possible answer to the reconstruction problem. The methodology is demonstrated by reconstructing two real world CIREN cases.

Case Selection

The real world crash cases were selected from CIREN. Only cases with single event, frontal impact with PDOF of 0 ± 10° and no rollover were considered. Cases were selected that provided enough information for reconstruction in HVE (vehicle type, collision partner involved, Collision Deformation Classification(CDC), Principal Direction of Force (PDOF), Crush and DeltaV) and also enough information for occupant simulation in MADYMO (age, height, weight of the occupant, occupant role (driver or passenger), restraints used, air bag information, seat performance information, etc). One

important criterion for case selection was good occupant-vehicle contacts that could be simulated. All cases with air bag failure, seat performance failure and seat belt failure were ignored. Cases where the occupant was asleep or in an out-of-position (OOP) states were ignored. In this study, only cases where the occupant had brain injury were considered. Only brain injury cases were selected so that these cases, in future, can be evaluated using NHTSA-developed finite element head model (FEHM). The cases were not filtered based on vehicle model year as the idea was to select cases and study the use of optimization as a reconstruction tool. Based on these criteria, two brain injury cases were selected.

"**Moderate Brain Injury" Case:** Details of the "moderate brain injury" case are provided below.

The crash occurred during the hours of daylight. The weather at the time of the crash was clear and dry. The posted speed limit was 45 mph. The crash occurred on a two lane roadway (Figure 2). Case Vehicle (V1, in red), a 1999 four-door Chevrolet Cavalier, was traveling eastbound. Vehicle two (V2), a 1997 four-door Oldsmobile Achieva, was traveling westbound. V1 crossed into the westbound lane. V1's front struck the front of V2. The impact caused V1 to rotate counterclockwise, coming to rest in the eastbound lane. V2 was forced to the right and rearward coming to rest off the shoulder of the westbound lane. The case vehicle had a delta-V of 35 mph.

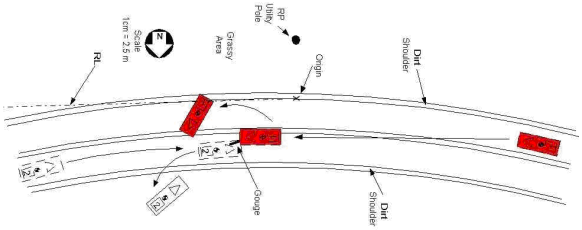


Figure 2. Crash Scene for "moderate brain injury" case

The case occupant was the driver of the vehicle (V1). The case occupant was a 61 year old male, 175cm in height and 86 kg in weight. He was wearing the available lap and shoulder belt and had frontal air bag deployment. The occupant sustained moderate brain injury (Table 1). The occupant also sustained AIS 1 injuries not listed in Table 1.

Table 1.
Occupant Injuries

AIS Code	Description
1610002	Cerebral concussion
5408243	Colon laceration, perforation
5420222	Mesentery laceration
8516143	Fibula Fx, bimalleolar

Occupant-vehicle contacts were taken from CIREN (Table 2).

Table 2.
Occupant-Vehicle Contacts

Contact	Component	Body Region
1	Air bag-driver side	Head
2	Steering wheel rim	Chest
3	Knee bolster	Knee-Left
4	Knee bolster	Knee-Right

"**Severe Brain Injury" Case:** Details of the "severe brain injury" case are provided below.

This crash occurred at night (with street lights) on the southbound lanes of a six lane divided freeway with dry conditions with a slight downgrade in the location (Figure 3). The posted speed limit was 60 mph. The case vehicle (in red), a 2006 Scion TC 2-door hatchback, was southbound in lane two of the three lanes and attempted to change lanes to the right and pass an unknown vehicle in lane two. Another unknown vehicle was traveling at a slower speed in lane one. V1 then swerved left and began braking as it crossed over lanes two and three before impacting the left side concrete barrier with the front of V1. V1 rotated slightly counterclockwise and came to final rest in lane three facing east. The case vehicle had a delta-V of 29 mph.

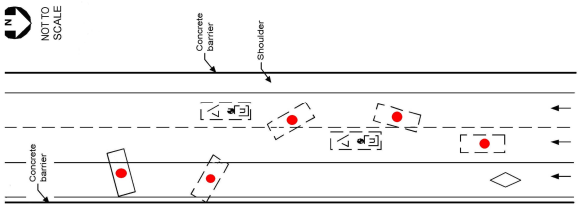


Figure 3. Crash Scene for "Severe Brain Injury" case

The case occupant was the driver of vehicle V1. The case occupant, a 27 year old male (178 cm in height and 69 kg in weight) was wearing the lap and shoulder belt with the seat belt pretensioners firing. The frontal steering wheel air bag as well as the knee bolster air bag deployed. The occupant rode down the deploying steering wheel air bag, causing the head to hit the steering wheel rim/hub. The occupant suffered severe brain injuries (Table 3). The occupant also suffered facial injuries and other AIS 1 injuries that are not listed in Table 3 as these were not considered in the analysis.

Table 3.
Occupant Injuries

AIS Code	Description
1406285	Cerebrum diffuse axonal injury (DAI)
1406843	Cerebrum subarachnoid hemorrhage
1504043	Vault skull fracture
8526043	Pelvis Fx
6506202	Lumbar Spine Fx, transverse process

The occupant-vehicle contacts were taken from CIREN (Table 4). Only contacts listed as "Certain" in CIREN were considered. "Probable" and "Possible" contacts were not considered in this study. This case was chosen particularly because the occupant sustained "diffuse axonal injury (DAI)", an injury which will be evaluated further using NHTSA developed FEHM and it was also the best-described case among all others.

Table 4.
Occupant-Vehicle Contacts

Contact	Component	Body Region
1	Steering wheel hub	Face
2	Steering wheel rim	Face
3	Knee air bag	Knee-Left
4	Knee air bag	Knee-Right
5	Knee bolster	Knee-Left
6	Foot controls	Right Foot

Crash Pulse Generation

The crash pulse generation for the two selected cases is described below:

"Moderate Brain Injury" case: For this case, the deceleration pulse of the case vehicle was available from the EDR. Since the selected case had a PDOF of 350°, there were both longitudinal and lateral components of the crash pulse. Since the EDR did not record any lateral component, it was estimated from HVE. The case was reconstructed in HVE. Not only the crash quantities were matched but also a consistent post impact motion, as given in CIREN crash analysis, was ensured. Generic vehicle models were selected for both vehicles. Both vehicle models were updated with respect to the exterior vehicle specifications: front overhang, rear overhang, overall length and width, wheelbase and weight. The exterior specifications for both vehicles were obtained from the CIREN case information. The total weight used was the sum of the "Curb weight," "Weight of the Occupants," and "Cargo weight." Vehicle stiffness plays an important role in correct crash pulse generation. Hence, the front, side, rear, top and bottom stiffnesses and the inertias of these generic vehicle models were updated based on the values available from actual vehicle models available in the HVE vehicle database. After the vehicle set up was completed in the vehicle mode, the crash event was set up in the event mode (Figure 4).

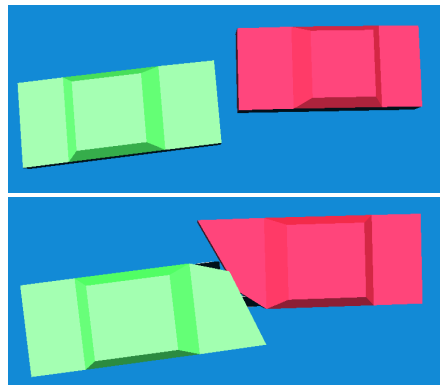


Figure 4. Crash Event.

The vehicles were positioned with respect to the global coordinate system. An estimated initial velocity was then assigned to each vehicle as their velocities were unknown. To generate a valid crash pulse for the selected CIREN case, various crash quantities (i.e. Principal Direction of Force (PDOF), Collision Deformation Classification (CDC), Crush and Delta-V) were matched between CIREN and the HVE simulation by carrying out parametric variations with respect to the impact location, vehicle velocities, inter-vehicle friction, etc. Since CIREN does not report all these quantities for the non-case vehicle, only Delta-V was matched for the non-case vehicle. A good match between CIREN and HVE was obtained for both the case and non-case vehicle (Table 5 and Table 6). The EDSMAC4 module [10], which is a 2D physics program in HVE, was used to generate the crash pulse.

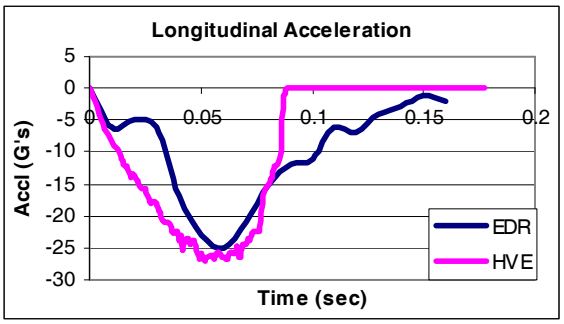
Table 5.
Case vehicle match

Chevrolet	CIREN	HVE
DeltaV, mph	35	35.5
Crush, in	32.6	31.49
CDC	12FYEW5	12FYEW6
PDOF(deg)	350	349.3

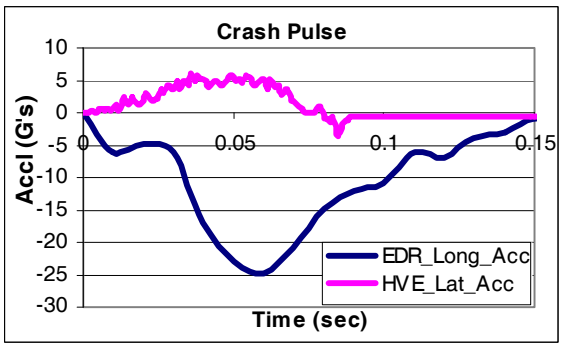
Table 6.
Non-case vehicle match

Oldsmobile	CIREN	HVE
DeltaV, mph	33.5	34.5

After the crash quantities were matched between HVE and CIREN, the longitudinal component of crash pulse obtained from HVE was compared with the EDR pulse. Since the longitudinal component obtained from HVE showed good match with that obtained from EDR (Figure 5a), it was decided to use the corresponding lateral component of crash pulse obtained from HVE for reconstruction. The final crash pulse used for this case included the longitudinal component from EDR and lateral component from HVE (Figure 5b).



(a)



(b)

Figure 5. (a) Longitudinal component of crash pulse. (b)Crash pulse used for "Moderate Brain Injury" case.

<u>"Severe Brain Injury" case:</u> For this case, no EDR data was available. HVE was used to generate the crash pulse. Similar to the "moderate brain injury" case, a generic vehicle model was used for the case vehicle. The vehicle model was updated with respect to the vehicle parameters. Since no stiffness and inertia information were available for the case vehicle, the stiffness and inertia information from 2006 Toyota Corolla model available in HVE database was used to update the case vehicle. The crash event was set up in HVE (Figure 6) and crash quantities were matched between HVE and CIREN to generate the crash pulse.

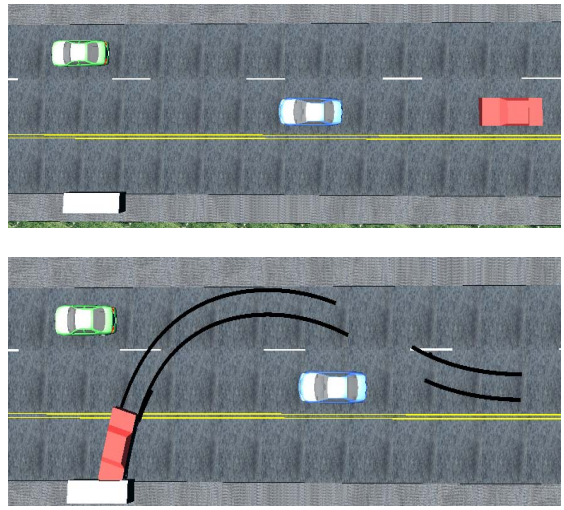


Figure 6. Crash Event

A good match between CIREN and HVE was obtained with respect to the crash quantities for the case vehicle (Table 7).

Table 7.
Case vehicle match

Scion TC	CIREN	HVE
DeltaV, mph	29.2	28.7
Crush, in	18.89	18.92
CDC	12FDEW3	12FDEW3
PDOF(deg)	350	351.4

The EDSMAC4 module in HVE was used to generate the crash pulse (Figure 7).

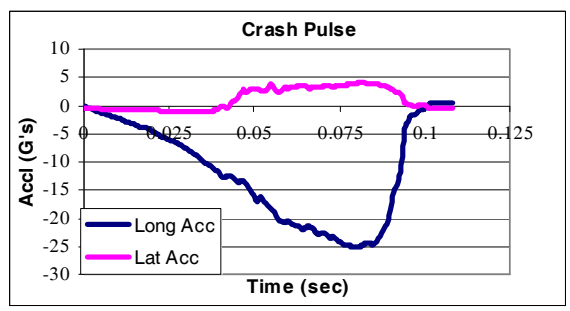


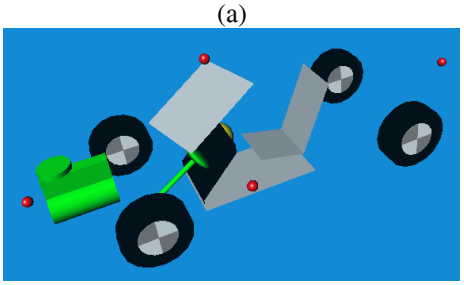
Figure 7. Crash Pulse for "Severe Brain Injury" case.

Occupant Simulation Set Up

<u>"Moderate Brain Injury" case:</u> The occupant simulation was set up in MADYMO, which is a widely used occupant safety analysis tool that can be used to simulate the response of an occupant in a dynamic environment. The occupant size for this "moderate brain injury" case was close to a 50th percentile size, and hence the Hybrid-III (H-III) 50th ellipsoid model was used as occupant model in MADYMO. The case vehicle interior surfaces were

created in MADYMO. The location of these surfaces was obtained from HVE, which had the actual vehicle model of a Chevrolet Cavalier available in its vehicle database. The contact surfaces were first created in HVE (Figure 8) and only the necessary contact surfaces were created based on the contacts listed in CIREN between the occupant and the vehicle interior. This information was then used to create the case vehicle in MADYMO (Figure 9).

(a)



(b)

Figure 8. Contact surfaces generated in HVE (a) Full View, and (b) No Body View.

The properties for the seat structure, seat back, seat cushion, knee bolster, steering column and the contact characteristics between the occupant model and the vehicle interior were taken from the frontal impact application file available in MADYMO [11], which has generic but realistic properties. Since the occupant (driver) had an air bag deployment during the crash, a generic air bag model was added to the steering wheel hub. The generic driver air bag model was selected from MADYMO applications. Since the occupant was wearing the lap/shoulder belt during the event, a finite element lap and shoulder belt was created and wrapped around the occupant (Figure 9). The properties for the belts were taken from MADYMO application file to be close to the realistic properties. This run was further set up according to the case information given in CIREN. The dummy was positioned in a normal posture with the right foot on the brake and the left foot on the floor.

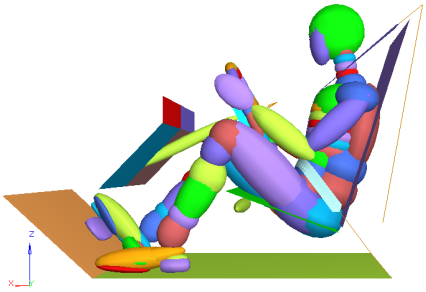


Figure 9. Impact Simulation model for "moderate brain injury" case.

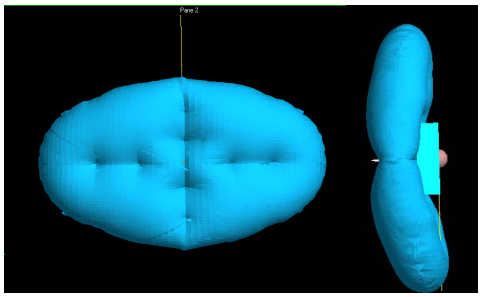
The EDR-HVE combination crash pulse (Figure 5b) was used to drive the simulation. The baseline run was obtained once the occupant-vehicle contacts in simulation were matched with those listed in CIREN.

<u>"Severe Brain Injury" case:</u>

A similar set up as explained for "moderate brain injury" case was followed for the "severe brain injury "case with a few additions. According to the case information given in CIREN, a seat belt pretensioner and a knee bolster air bag were installed in the vehicle. The pretensioner fired and the knee-bolster air bag deployed during the crash. A pretensioner and knee bolster air bag were added to the MADYMO model. Since no generic knee air bag model was available in MADYMO, one was created by scaling and shaping the generic steering wheel (SW) driver air bag model. The case vehicle's actual knee air bag model was 18 liters in volume and had tethers 50 mm in length. To create a knee air bag model close to the actual air bag, the following was done:

- The reference geometry of the SW driver air bag (which was circular) was scaled non-uniformly in X and Y direction to form an elliptical shape, which is a better approximation to the actual knee air bag (Figure 10a).
- Tethers were added (Figure 10b) each with a length of 50 mm. Since the location of the tethers was not known, eight tethers were added to control the shape of the deploying air bag.
- The air bag was scaled until inflated volume was close to 18 liters. The volume of the modeled air bag was 17.67 liters (Figure 11).

(a)



(b)

Figure 10. (a) Actual, and (b) modeled knee bolster air bag

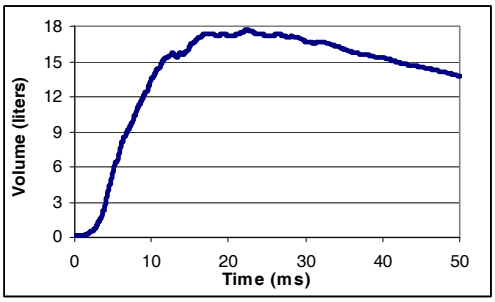


Figure 11. Volume vs. Time for knee air bag.

The crash pulse obtained from HVE (Figure 7) was used to drive the model. The baseline run (Figure 12) was obtained for this case by matching the occupant-vehicle contacts with those listed in CIREN.

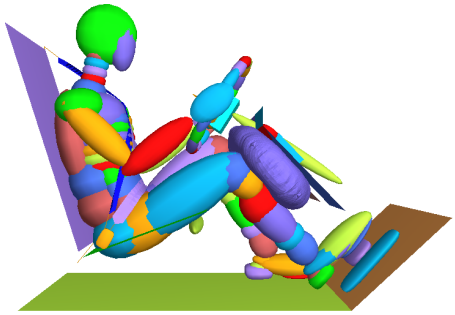


Figure 12. Impact Simulation model for "Severe Brain Injury" case.

The unknown variables were identified for both the cases during their respective baseline set up.

Injury Thresholds

The injury threshold values for the injuries sustained by the occupant were established from literature review. Injury thresholds were also found for the uninjured body regions. These injury thresholds were established for use in the optimization study. It is important to note that the point of the paper was to study the optimization methodology as a reconstruction tool. The injury threshold values were established from the best available data set, and are not intended as definitive transformation to the reported occupant injuries. Since any chosen injury threshold value would represent a certain probability of injury and there was an uncertainty in choosing a value, the injury threshold values given in reports and papers were selected. The injury thresholds for the two cases are described below:

"Moderate Brain Injury" case: The occupant sustained AIS 1 and AIS 2+ injuries. AIS 1 injuries were not incorporated into this analysis as these were minor injuries such as skin abrasion/laceration. The occupant sustained:

- Abdominal Injury (AIS 2 & AIS 3): Since the occupant had two abdominal injuries (AIS 2 and AIS 3), only the AIS3 injury was considered. It was mentioned in the CIREN injury analysis that the "Belt Restraint webbing/buckle" was the injury source for both the abdominal injuries. The Hybrid III 50th dummy model does not have any force output or any injury metric output for the abdomen. Therefore, the lap belt –abdomen contact force was related to abdominal injury. The force corresponding to 30% probability of injury was obtained as the injury threshold. Since there was no test data available for frontal impact, the threshold force was obtained from the experimental results obtained from lateral impacts [12].
- Lower Extremity Injury (AIS 3): The occupant suffered a bimalleolar fibula fracture on the right foot because of inversion-eversion. The inversion-eversion moment corresponding to 30% probability of injury was obtained as the threshold value [13]. The lower tibia moment was compared with this threshold value for predicting this injury.
- Head Injury (AIS 2): For cerebral concussion sustained by the occupant, the HIC value corresponding to mild traumatic brain injury (MTBI) given by Pellman et al [14] was used. This value is based on national football league (NFL) reconstruction data. This was the only published human volunteer data available that correlated HIC with concussion. Since concussion occurred because of soft contact, HIC_{36}, which is calculated over an extended time period of the acceleration pulse, was thought to be better predictor than HIC_{15} and hence HIC_{36} was used.

The threshold values for the uninjured body regions were also obtained from literature [15]. The established threshold values (Table 8) were used for optimization.

Table 8.
Established injury threshold values for "Moderate Brain Injury" case

Body Region	Injury Criteria	Threshold from Literature
Head	HIC_{36}	250
Neck	Nij	1.0
Thorax	Acceleration & Deflection	60g & 63mm
Abdomen	Abdominal Force	3200N
Femur	Resultant Force	10KN
Lx-Injury (Tibia Injury)	Tibia Index	1.3
Lx-Injury (Fibula-Injury)	Eversion-Inversion Moment	35Nm

"Severe Brain Injury" case: Similar to the "Moderate Brain Injury" case, the AIS 1 injuries were ignored for this case as well. The occupant in this case sustained:

- Cerebrum diffuse axonal injury (AIS 5): The injury threshold was obtained from Takhounts et al [16]. Angular acceleration was used as injury criteria. In [16] the relationship between DAI & maximum principal strain and that between angular acceleration & maximum principal strain was used to establish injury risk as a function of angular acceleration. The angular acceleration corresponding to 50% probability of injury was used as the threshold value.
- Cerebrum subarachnoid hemorrhage (AIS 3): The injury threshold was obtained from Takhounts et al [16]. Angular acceleration was used as injury criteria. The angular acceleration corresponding to 30% probability of injury was used as the threshold value. Data was not available to form AIS 3 and AIS 5 injury risk curves in [16]. Hence, assumptions were made for DAI and hemorrhage threshold from the injury risk curve available.
- Vault skull fracture comminuted (AIS 3): This was a depressed left, frontal skull fracture associated with overlying laceration. Since the Skull Fracture Criteria (SFC) is only applicable for linear skull fractures, HIC was used as the injury criteria for this depressed skull fracture. Since this was a contact-type injury, HIC_{15} was used to properly capture the short time period of the contact that causes the injury. The HIC_{15} value corresponding to 30% probability of AIS3+ injury was used as the threshold value [17].
- Pelvis fracture (AIS 3): According to CIREN injury analysis, this fracture was sustained by loading onto the lap belt. The Hybrid III 50th percentile dummy model does not have any force output or any injury metric output for the pelvis. Therefore, the lap belt–abdomen contact force was related to pelvis injury. The injury threshold was obtained from Salzar et al [18].
- Lumbar spine fracture transverse process (AIS2): Since no injury risk curves are available for lumbar spine, the lumbar spine load cell output from H-III 50th dummy model was simply monitored and was not used in the optimization analysis.
- Facial Injuries (AIS 3-Orbit fracture, AIS 2-Nose fracture): Since the H-III 50th dummy model does not have any load cell output from the face region, facial injuries were ignored in this analysis.

The established injury thresholds (Table 9) were used for optimization.

Table 9.
Established injury threshold values for "Severe Brain Injury" case

Body Region	Injury Criteria	Threshold from Literature
Head-DAI	Angular Acceleration	8000 rad/s^2
Head-Hemorrhage	Angular Acceleration	6250 rad/s^2
Head-Skull Fracture	HIC_{15}	1177
Neck	Nij	1.0
Thorax	Acceleration & Deflection	60g & 63mm
Pelvis	Force	5470N
Femur	Resultant Force	10KN
Lx-Injury	Tibia Index	1.3

Optimization Set up

After setting up the baseline run and obtaining the injury thresholds, optimization was set up for the two cases using modeFrontier software. The assumed parameters identified earlier were used as design variable and their ranges were defined (Table 10).

Table 10.
Design Variables

	Parameters	Range [Ref]
SEAT	Seat Friction	0.2-0.4
KNEE BOLSTER	Knee Bolster Properties	± 20% [1]
	Knee Bolster Friction	0.1-0.4
BELT SYSTEM	Belt Segment Properties	± 20% [1]
	FE Lap/Shoulder Belt Properties	± 20% [1]
	Belt Friction	0.1-0.3
	Retractor Properties (film spool effect)	± 20% [1]
	Retractor Locking Time	1ms -30ms
DRIVER AIR BAG	Air bag Firing Time	5ms-55ms
	Air bag Friction	0.1- 0.3
	Steering Column Angle (Air bag Deployment Angle)	± 5° around baseline
	Air bag Mass Flow Rate (MFR)	± 20% [1]
	Steering Col. Position (translation –X-dir i.e. Air bag Position)	± 15% around baseline
CRASH PULSE	Crash Pulse-Y component	Scaling 0.5-1.5
OTHER PARAMETERS	Thorax-SW Loading	Scaling 0.8-1.2
	Friction Shoes –Toe board/Floor	0.5-0.7
	Friction Brake-Right Shoe	0.5-0.7
ADDITONAL PARAMETERS FOR "SEVERE BRAIN INJURY" CASE	Knee Air bag (KAB) MFR	± 20% [1]
	KAB Firing Time	5ms-40ms
	Pretensioner Firing Time	10ms-40ms
	Load Limiting Force	3.5KN-5.5KN
	Crash Pulse-X component	Scaling 0.85-1.4

Objective function and constraints were set up. Since the objective function has a great effect on the final solution, a normalized form of the objective function (Equation 1 & Equation 2) weighted based on the AIS score of the injuries sustained by the occupant, was formed in this study. This form of objective function can be generalized to other cases. For "moderate brain injury" case we had:

$$\left(1-\left(\frac{2\times HIC\times AbF_{cr}\times M_{cr}+3\times HIC_{cr}\times AbF\times M_{cr}+3\times HIC_{cr}\times AbF_{cr}\times M}{8\times HIC_{cr}\times AbF_{cr}\times M_{cr}}\right)\right)^2 \quad (1)$$

where,

(2, 3, 3)= AIS score of the respective injuries used as weights, HIC=HIC value from simulation, HIC_{cr}=Critical HIC value (250), AbF=Abdomen force value from simulation, AbF_{cr}=Critical abdomen force value (3200N) M=Inversion-Eversion moment from simulation, M_{cr}=Critical inversion-eversion moment value (35 N-m).

For "severe brain injury" case we had:

$$\left(1-\left(\frac{5\times DAI\times H_{cr}\times P_{cr}\times SFC_{cr}+3\times DAI_{cr}\times H\times P_{cr}\times SFC_{cr}+3\times DAI_{cr}\times H_{cr}\times P_{cr}\times SFC+3\times DAI_{cr}\times H_{cr}\times P\times SFC_{cr}}{14\times DAI_{cr}\times H_{cr}\times P_{cr}\times SFC_{cr}}\right)\right)^2 \quad (2)$$

where,

(5, 3, 3, 3)= AIS score of the respective injuries used as weights, DAI=Head angular acceleration from simulation, H = Head angular acceleration from simulation, SFC= HIC from simulation, P=Pelvis force from simulation, DAI_{cr}=Critical head angular acceleration value for diffuse axonal injury (8000 rad/s^2), H_{cr} = Critical head angular acceleration value for hemorrhage (6250 rad/s^2), SFC_{cr}= Critical HIC value for skull fracture (1177), P_{cr}=Critical pelvis force value for pelvis injury (5470N)

Two types of constraints were defined i.e. contact constraints and injury constraints (Table 11 & Table 12).

Table 11.
Constraints for "moderate brain injury" case

Contact Constraints	Injury Constraints
Head-Air bag $_{\text{contact-force}}$>0	HIC_{36} > 250
Thorax-SW $_{\text{contact-force}}$>0	Abdomen_Force>3200 N
KneeL-Bolster $_{\text{Contact-force}}$>0	RF_IE_Moment>35Nm
KneeR-Bolster $_{\text{Contact-force}}$>0	RF_DF_moment <52Nm
Head-Hub $_{\text{contact-force}}$=0	Nij<1
Head-SW $_{\text{contact-force}}$=0	Thorax_acc <60g's
Thorax-Hub $_{\text{contact-force}}$=0	Thorax_deflection<63mm
	Femur_force<10KN
	TI <1.3

where *RF_IE-Moment* is the right foot inversion-eversion moment and *RF_DF_moment* is the right foot dorsi-flexion moment. Right foot dorsi-flexion moment was used to make sure that the right foot fracture was because of inversion-eversion and not because of dorsi-flexion moment.

Table 12.
Constraints for "Severe Brain Injury" case

Contact Constraints	Injury Constraints
Head-Air bag $_{contact\text{-}force}$>0	HIC_{15} > 1177
Head-Hub $_{contact\text{-}force}$>0	Ang_Acc>8000 rad/s^2
Head-SW $_{contact\text{-}force}$>0	Pelvis_force>5470 N
KneeL-Bolster $_{Contact\text{-}force}$>0	Nij<1
KneeR-Bolster $_{Contact\text{-}force}$=0	Thorax_acc <60g's
Thorax-Hub $_{contact\text{-}force}$=0	Thorax_deflection<63mm
Thorax-SW $_{contact\text{-}force}$=0	Femur_force<10KN
KneeR-Knee Air bag>0	TI <1.3
KneeL-Knee Air bag>0	

The objective function and the constraints were defined such that they minimize the differences in injuries and occupant-vehicle contacts between the real world data and the model prediction. A multi objective genetic algorithm (MOGA) with Sobol DOE was used for optimization. Convergence for the "moderate brain injury" case was obtained after 571 simulations and convergence for the "severe brain injury" case was obtained after 452 simulations. A correlation study and a best case scenario study were also carried out.

RESULTS

Optimization Results

"Moderate Brain Injury" case: Of the 571 simulations (Figure 13), 388 were feasible i.e. matched CIREN listed occupant-vehicle contacts and injuries. From these feasible solutions, the one with the minimum value of objective function was selected as the solution.

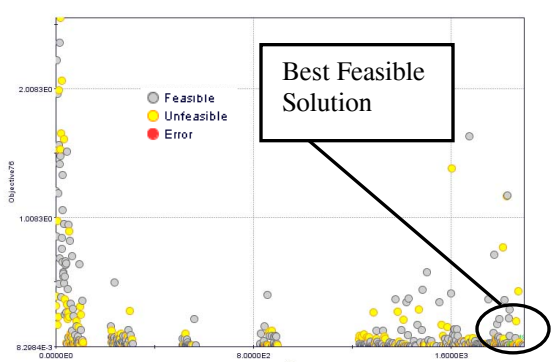


Figure 13. Design ID vs. Objective Function.

The optimized solution obtained was checked for injuries and contacts. The optimized solution matched all the CIREN listed contacts (Table 13, Figure 14).

Table 13.
Contact Match

CIREN listed Contacts	Matched by Simulation
1. Head –Air bag	√
2. KneeL-Bolster	√
3. KneeR-Bolster	√
4. Thorax-SW	√

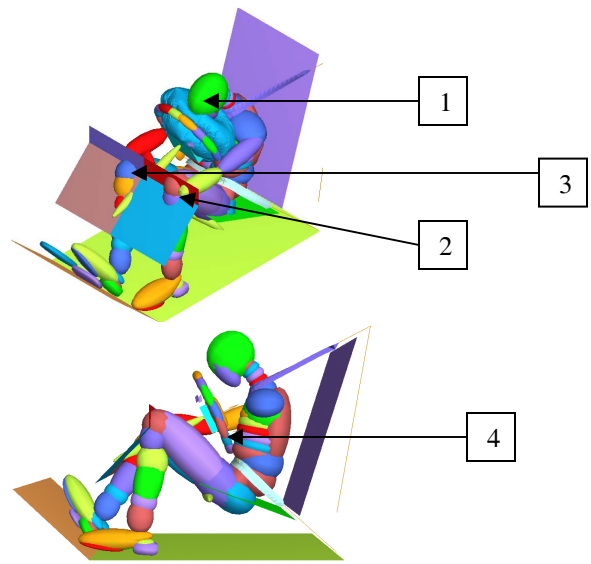


Figure14. Occupant-Vehicle contacts.

The occupant injuries predicted by the optimized solution were also matched with those listed in CIREN. The optimized solution predicted all the injuries listed in CIREN correctly (Table14).

Table 14.
Injury Match

Body Region	CIREN listed Outcome	Injury Criteria	Threshold from Literature	Opt-Solution Prediction	Predicted Correctly
Head (AIS 2)	Injured (Concussion)	HIC_{36}	250	251.3	√
Neck	Not Injured	Nij	1.0	<1.0	√
Thorax	Not Injured	Acceleration & Deflection	60g & 63mm	<60 g & <63mm	√
Abdomen (AIS 3)	Injured	Lapbelt-abdomen contact force	3200N	3464.6 N	√
Femur	Not Injured	Resultant Force	10KN	<10 KN	√
Lx-Injury (Tibia Injury)	Not Injured	TI	1.3	<1.3	√
Lx Injury (Fibula Injury) (AIS 3)	Injured (Fx) Right Foot	Eversion-Inversion Moment	35Nm	42.7 Nm	√
Lx Injury	Right Foot	Dorsi-Flexion Moment	52 Nm	<52Nm	√
Lx Injury (Fibula Injury)	Uninjured Left Foot	Eversion-Inversion Moment	35Nm	<35Nm	√

"Severe Brain Injury" case: This case converged after 452 simulations of which 119 were feasible (Figure 15). The solution with minimum objective function value was selected as the solution to the problem.

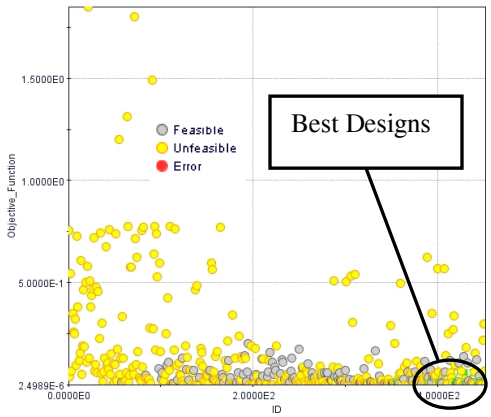


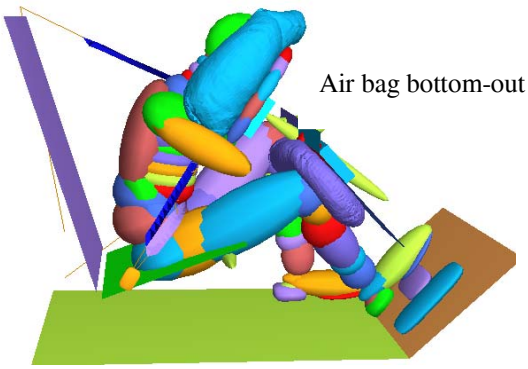
Figure 15. Design ID vs. Objective Function.

The optimal solution was checked for both injuries and occupant-vehicle contacts. The occupant-vehicle contacts (Table 15, Figure 16) and injuries (Table 16) predicted by the optimal solution matched with those listed in CIREN.

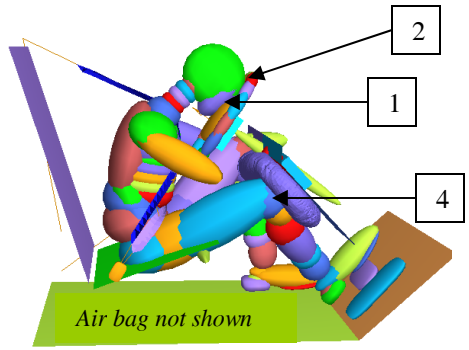
Table15.
Contact Match

CIREN Listed Contacts	Matched by Simulation
1. Face- SW Hub	√
2. Face-SW Rim	√
3. KneeL-Knee Air bag	√
4. KneeR-Knee Air bag	√
5. KneeL-Knee Bolster	√
6. Right Foot-Foot Controls	√

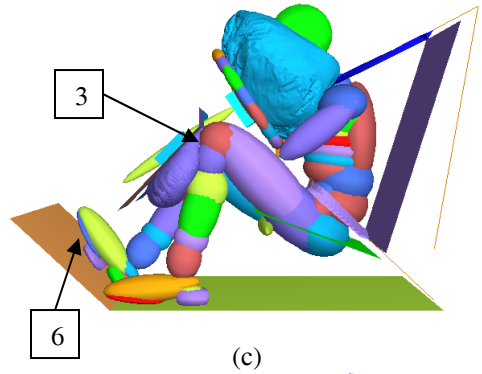
According to the case information, the occupant rode down the SW driver-air bag causing the head to hit the steering wheel rim and hub. The SW driver-air bag bottom-out was captured by the optimal solution (Figure 16a).



(a)



(b)



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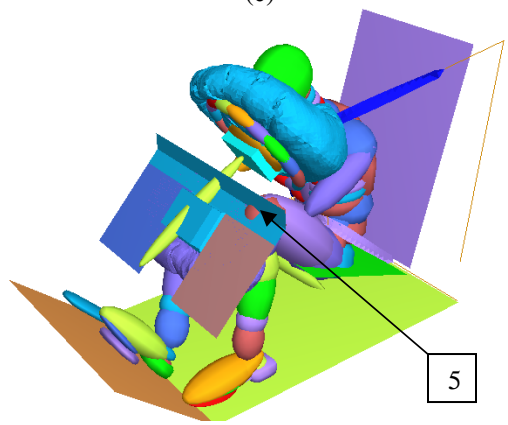


(d)

Figure 16. (a) Air bag bottom-out, (b, c, d) Occupant-Vehicle contacts.

Table16.
Injury Match

Body Region	CIREN listed Outcome	Injury Criteria	Threshold from Literature	Opt Solution Prediction	Predicted Correctly
Head	Diffuse Axonal Injury	Angular Acceleration	8000 rad/s^2	8011.6 rad/s^2	√
	Hemorrhage	Angular Acceleration	6250 rad/s^2	8011.6 rad/s^2	√
	Skull Fracture	HIC_{15}	1177	1183.3	√
Neck	Not Injured	N_{ij}	1.0	< 1.0	√
Thorax	Not Injured	Acceleration & Deflection	60g & 63mm	< 60g & < 63mm	√
Pelvis	Injured	Lapbelt-Abdomen Force	5470 N	6283.6	√
Lumbar Spine	Injured	Only Monitored	------	------	------
Femur	Not Injured	Force	10KN	< 10KN	√
Lx Injury	Not Injured	Tibia Index	1.3	< 1.3	√

Correlation Study and Best Case Scenario Study Results

A Correlations study and a best case scenario study were carried out to find critical parameters and to find a minimal or no injury scenario for the given case. These studies are only presented for the "moderate brain injury" case due to space limitations.

Correlation Study: This study was carried out to find critical parameters. The critical parameters were identified by analyzing the correlation coefficients obtained from the optimization study. Apart from the design space explored by the optimizer (571 design points), the design space was explored further by adding 150 Sobol DOE points and 100 random DOE points (Figure 17) in the design space to make sure the design space was properly explored to generate good correlations. This "physically possible" design space with 821 design points was then used to generate the correlation coefficients.

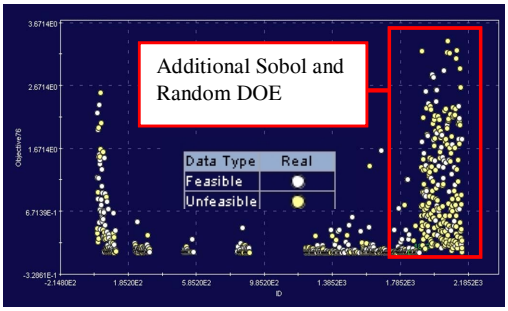
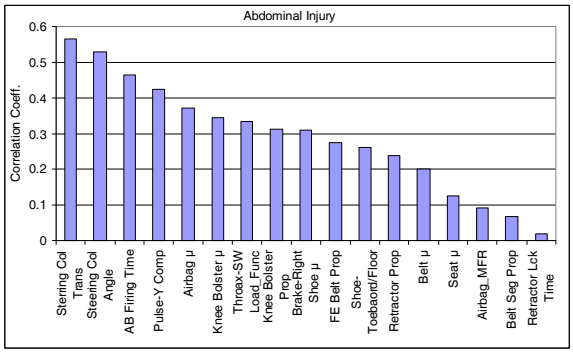
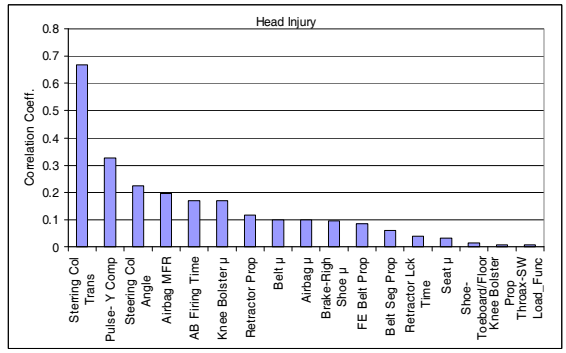


Figure 17. Correlation Study

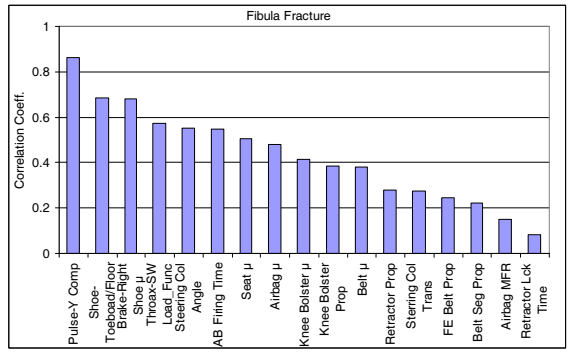
The software modeFrontier provides correlation coefficients between all the input variables and the output variables. The correlation coefficients range from -1 (strong negative correlation) to 1 (strong positive correlation). Using the correlation coefficients, the critical parameters were identified for each of the injuries sustained by the occupant (Figure18). No particular threshold was used to separate out the critical parameters. Only a ranking from most-critical to least-critical was obtained. Absolute values were used for the correlations plots (Figure 18).



(a)



(b)



(c)

Figure 18. Critical parameters for (a) abdominal injury (b) head injury, and (c) fibula fracture

If more detailed information can be obtained on some of the critical parameters, a more accurate reconstruction analysis may be carried out.

Best Case Scenario Study: For identifying the best case scenario for the "moderate brain injury" case, the pulse was fixed to the optimized pulse obtained previously. For this given pulse, seat position and seat track position, an optimization attempt was made to see if a scenario with no-injury or minimum injuries could be obtained. A few changes as listed below were made to the problem set up.

- After fixing the pulse to the optimized one, the Sobol DOE was re-generated.
- The injury constraints on HIC, abdomen force and fibula fracture were modified with respect to the threshold. Lower non-injurious thresholds were selected. For HIC a threshold value of 100, for abdomen force a value of 1000N and for Fibula fracture a value of 10N-m was selected. The constraints and objective function were modified accordingly.

The problem converged after 602 simulations (Figure 19).

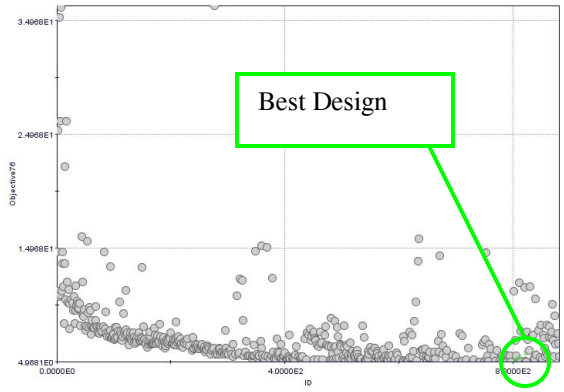


Figure 19. Best Case Scenario

After the convergence was obtained, the injuries were compared between the best case scenario solution and the optimized solution obtained before (Table 17).

Table 17.
Best case scenario: Injury comparison

Body Region	CIREN listed Outcome	Injury Criteria	Threshold Value	Injury Value	Best Case Scenario Injury Prediction	Optimal Solution Injury Prediction
Head (AIS2)	Injured (Concussion)	HIC36	250	175.4	**Not Injured**	**Injured**
Neck	Not Injured	Nij	1.0	<1.0	**Not Injured**	**Not Injured**
Thorax	Not Injured	Acceleration & Deflection	60g & 63mm	<60 g & <63mm	**Not Injured**	**Not Injured**
Abdomen (AIS3)	Injured	Lapbelt-abdomen contact force	3200N	3256 N	**Injured**	**Injured**
Femur	Not Injured	Resultant Force	10.6KN	<10.6KN	**Not Injured**	**Not Injured**
Lx-Injury (Tibia Injury)	Not Injured	TI	1.3	<1.3	**Not Injured**	**Not Injured**
Lx Injury (Fibula Injury) (AIS3)	Injured (Fx) Right Foot	Eversion-Inversion Moment	35Nm	41.8 Nm	**Injured**	**Injured**
Lx Injury	Right Foot	Dorsi-Flexion Moment	52 Nm	<52Nm	**Not Injured**	**Not Injured**
Lx Injury (Fibula Injury)	Uninjured Left Foot	Eversion-Inversion Moment	35Nm	<35Nm	**Not Injured**	**Not Injured**

It was found that for the given crash pulse, seat position and seat track position, the optimizer was able to find a scenario without head injury. The abdominal injury and fibula fracture were still present. The abdominal force and fibula inversion-eversion moment did come down to lower values as compared to the optimized solution, but did not go below the injury threshold for the given range of design variables.

DISCUSSION

This paper presents an optimization methodology for reconstructing real world crashes. Optimization was selected as it can give the best possible solution to a problem where instead of a specific value; a range is available for the input parameters. Even though crash reconstruction engineers collect a lot of details from the crash site, computerized crash reconstruction utilizes much more information than reported and, as such, specific values for a lot of parameters are not available and one has to work with a range of the parameters. Since even within a given range of input parameters the injury predictions can be quite different for the same case [1], it is imperative to take into account all the variability and come up with the best possible solution to the reconstruction problem. Optimization is one way to accomplish this and hence was used.

Crash pulse plays a very vital role when predicting occupant injuries. The crash pulse for "moderate brain injury" case was available from the EDR data with part of it, i.e. the lateral component, generated using HVE. Also HVE was used to generate the crash pulse for "severe brain injury" case. HVE has its own limitations, insofar as the stiffness of the vehicle, which plays an important role in generating the right crash pulse, can only be defined as linear and homogenous for any given side of the vehicle. Additionally, hard spots cannot be defined. As a result, the crash pulse obtained from HVE is not precise, but approximate. Hence the Y-component of crash pulse for the "moderate brain injury" case and the entire crash pulse (X &Y components) for the "severe brain injury" case were used as design variables during optimization. An EDR pulse, if available, should be used to reduce the design variables in the optimization process.

Genetic algorithms (GA) are categorized as global search heuristics and can help find global minima as compared to gradient based methods that have a tendency of converge to local minima. Also genetic algorithms can provide solutions for highly complex search spaces. For these reasons, a GA was used for the optimization. Sobol DOE was used to generate starting population for the GA as it uniformly distributes starting points in the design space.

For the "severe brain injury" case, the retractor locking time was not used as a design variable. This case had a pretensioner and varying the retractor locking time was creating belt problems and thus it was fixed to 1ms. Also for the "severe brain injury" case, the knee air bag, created from the driver air bag, had the same mass flow rate characteristics as the driver air bag. During optimization, both the abscissa and the ordinate of the mass flow rate curve (for driver air bag and knee air bag) were used as design variables. They were varied by $\pm 20\%$ around the baseline. The "Thorax-SW loading" design variable as used in the "moderate brain injury case" was not

used as a design variable in the "severe brain injury" case as CIREN did not list any contact between the thorax and the steering wheel for this case.

The optimal solution obtained for both the "moderate brain injury" case and "severe brain injury" case correctly predicted the occupant-vehicle contacts and the injuries sustained by the occupant. Since these are the two most important things that can be matched with the real world data, the kinematics predicted by the optimal solution are believed to be reliable. The accuracy of the solution strongly depends on the quality of the crash data reported. It is reasonable to assume that further improvement in field investigation and amount of data collected can help improve reconstruction analysis.

The optimized solution for the "severe brain injury" and "moderate brain injury" cases did match the injury thresholds obtained from literature. Since injury risk curves were not available for some of the injuries sustained by the occupant, threshold were obtained from the best available data set, for example for the abdominal injury sustained by the occupant in the "moderate brain injury" case, injury threshold for frontal impact was obtained from lateral impact experiments. So it is possible for some of the optimized force levels not to match the reported injury severities. A more accurate analysis may be carried out if injury risk curves are available for all body regions under different impact conditions.

The correlation study conducted helped identify critical parameters. Figure 18a shows that for abdominal injury, air bag parameters (air bag position, air bag deployment angle, air bag firing time) and Y-component of the crash pulse were most critical. For head injury, air bag parameters (air bag position, air bag deployment angle, air bag firing time, air bag mass flow rate) and Y-component of the crash pulse were most critical (Figure 18b) and for fibula injury, Y component of crash pulse and shoe related friction parameters were most critical (Figure 18c). These critical parameters can help understand how different body regions are affected by the vehicle environment and what design changes can be made to mitigate these injuries. If more detailed information can be obtained on some of the critical parameters, a more accurate reconstruction may be carried out. The correlation coefficients in this paper were generated using all the design points i.e. around 821 simulations. A few hundred more simulations exploring the design space and further division of the step size defined for the design variables might help further improve the correlation coefficients.

A best case scenario study was done to find, using optimization, the design changes that could help mitigate all or some of the injuries sustained by the occupant for the given crash pulse, seating position and seat track position. The no-head injury outcome predicted by the optimizer was achieved by making changes to the belt system and air bag system. Compared to the optimal solution, the optimizer, for the best case scenario, predicted scaling down the belt properties i.e. reducing the stiffness of the seatbelts, increasing the air bag mass-flow rate and late firing of the air bag as design changes that can help mitigate head injury. These three design variables converged to values within the feasible range of variability. The correlation study results and best case scenario results predicted are only for this particular "moderate brain injury" case and should not be generalized. The results may be different for other cases.

In this study, neither full finite element nor human facet models that better define human geometry and material properties were used because of the prohibitive run times. For better reconstruction, human models should be preferred if the run time can be reduced. This study was done only for frontal impacts but an analysis such as the one presented in this paper can easily be extended to other types of impacts.

Future work may involve, among other things, reconstructing more real world crashes with different kinds of impacts, using human FE or facet models for better occupant simulations.

CONCLUSION

The optimization methodology presented in this paper and demonstrated by successfully reconstructing two real world crash cases introduces a method that can be applied for finding solutions to crash reconstruction problems, which due to unavailability of data, cannot be solved deterministically. The kinematics predicted by the optimal solution can give insight on how the occupant moved inside the vehicle during the crash, which can help provide better understanding of the crash. The paper also showed how critical parameters can be identified using the correlation coefficients calculated during the optimization process.

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