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A Computational Study of Injury Severity and Pattern Sustained by Overweight Drivers in Frontal Motor Vehicle Crashes

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Abstract

The objective of this study was to examine the role of body mass and subcutaneous fat in injury severity and pattern sustained by overweight drivers. Finite element models were created to represent the geometry and properties of subcutaneous adipose tissue in the torso with data obtained from reconstructed magnetic resonance imaging datasets. The torso adipose tissue models were then integrated into the standard multibody dummy models together with increased inertial parameters and sizes of the limbs to represent overweight occupants. Frontal crash simulations were performed considering a variety of occupant restraint systems and regional body injuries were measured. The results revealed that differences in body mass and fat distribution have an impact on injury severity and pattern. Even though the torso adipose tissue of overweight subjects contributed to reduce abdominal injury, the momentum effect of a greater body mass of overweight subjects was more dominant over the cushion effect of the adipose tissue, increasing risk of other regional body injuries except abdomen. Through statistical analysis of the results, strong correlations ($p < 0.01$) were found between body mass index and regional body injuries

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except neck injury. The analysis also revealed that a greater momentum of overweight males leads to greater forward torso and pelvic excursions that account for higher risks ($p < 0.001$) of head, thorax, and lower extremity injury than observed in non-overweight males. The findings have important implications for improving the vehicle and occupant safety systems designed for the increasing global obese population.

Keywords

Overweight; Crash injury; Modeling; Cushion effect; Momentum effect

1. Introduction

The overweight and obese population has increased substantially over the last two decades and represents a major public health concern in most industrialized countries (Mokdad et al. 2001; World Health Organization 2003). Meanwhile, road traffic crashes are one of the leading causes of injuries and death, approximately 1.2 million deaths a year worldwide (Peden et al. 2004). Thus road safety is also a major public health issue throughout the world. Several epidemiologic studies have been conducted to explore the relationship between these two public health issues.

A current debate exists on the mechanistic role of subcutaneous fat in causing motor vehicle crash (MVC) injuries. Arbabi et al. (2003) found that a classification of overweight was associated with lower abdominal injury compared to leanness, perhaps due to a 'cushion effect' of the fat. The cushion effect was significant for women whose greater subcutaneous fat depth than men may contribute to a lower risk of incurring severe abdominal injuries (Wang et al. 2003).

Meanwhile, Chohan et al. (1991) found that obese persons, defined as a body mass index (BMI) greater than 31 (kg/m^2), had higher mortality after blunt injury than persons with $\text{BMI} < 31$. Boulanger et al. (1992) also found that obese victims of MVCs were likely to have rib fractures, pulmonary contusions, and pelvic and extremity fractures, but less likely to have head trauma and liver injuries. Mock et al. (2002) found that increased occupant body weight was associated with increased mortality in MVCs. A U-shaped association was observed between driver injury risk and BMI (Whitlock et al. 2003), in which participants in the highest BMI (> 28.7) and lowest BMI (< 23.5) were twice as likely to have injuries as ones in the reference BMI range of 25.9-28.6. Zhu et al. (2006) found such an association among men, but not among women. While obesity can be an independent predictor for risk factor of mortality after blunt injury (Neville et al. 2004), the role of obesity and fat distribution of vehicle drivers in MVC injuries is not currently well understood. Recently, the authors conducted an epidemiologic study using real-world crash data together with a computer simulation study to examine the role of obesity in regional body injuries following MVCs (Zhu et al. 2010). The authors observed that overweight/obese male drivers had a higher risk of injury to the head, face, thorax, and spine as compared to non-obese male drivers.

The objective of this study was to investigate the mechanistic role of increased body mass and torso adipose tissue of overweight drivers in MVCs. Computational dummy models of overweight subjects were developed and the details of the modeling process are described here. The models were validated against sets of sled tests and then various frontal crash simulations were performed.

2. Methods

2.1 Modeling of Overweight Dummies

The well-validated and widely used MADYMO (TNO, The Netherlands) Hybrid III ellipsoidal 50th percentile male (BMI 25, height 1.77 m) and 5th percentile female (BMI 22, height 1.52 m) were used as the standard male and female dummy, respectively. To take into account the subcutaneous adipose tissue (SAT) of overweight persons for the dummy models, SAT mask files (color-coded for each anatomical region) generated from grey-scale magnetic resonance imaging (MRI) datasets of overweight subjects (Shen et al. 2004; Shen and Chen 2008) were utilized in this study. Four sets of data from human patients, male BMI 30 and 35, and female BMI 30 and 35, whose heights are almost identical (within 2% difference) to the standard MADYMO dummies, were acquired. The exempt status of using the existing de-identified datasets was reviewed and approved by the Institutional Review Board of St. Luke's-Roosevelt Hospital. Each dataset consists of 43 SAT mask files with a 256×256 pixel resolution for each slice. A specialized in-house computer tool was developed to extract the SAT region and reconstruct it into a three-dimensional geometry using triangulated surfaces in a stereolithography format.

An attempt was to generate a finite element mesh directly from the SAT geometry and then integrate it into the standard dummy models. The inner surface of the SAT geometry, however, did not accord with the outer surface of MADYMO torso ellipsoids, due to not only different configuration between human and dummy but also the supine or prone position of the subjects during MRI scans. Such a model caused numerical errors in model simulations.

Therefore, an alternate approach was used for this study. Five axial planes of the SAT geometry from pelvis to chest were extracted. The thicknesses of the SAT were then measured at six places on the half of each axial plane. At the same time, the outer surface of MADYMO torso ellipsoids was extracted with a clearance to be less than 1 mm to avoid overlapping between the ellipsoids and SAT geometry. Five axial planes of the outer surface at the same locations as the SAT geometry were also extracted. For each plane, a contour of the skin was generated by spline curves (Shih et al. 2005) based on the measured SAT thicknesses and then by mirroring the curves. An artificial three-dimensional SAT geometry, referred to as 'fat vest' in this study, was then generated by connecting the contour of the skin on each axial plane using B-spline surface interpolation (Shih et al. 2005).

HyperMesh version 9.0 (Altair Engineering, Troy, MI) was employed to create a finite element mesh for the fat vest by around 55,000 tetrahedral elements for the SAT and around 18,000 shell elements for the skin. Since shell elements are geometrically modeled in their mid plane, the contour of the skin was offset by half skin thickness for geometric fidelity.

Coincide nodes were used at the boundary of the SAT and skin. The finite element fat vest model was then integrated into the standard MADYMO dummy model to represent an overweight dummy model. To avoid penetration of the fat vest to the dummy ellipsoids during model simulation, a contact interface was defined using a multibody and finite element kinematic contact model (TNO 2008).

Material constitutive models and properties used for the fat vest are listed in Table 1. The Mooney-Rivlin coefficients (C_1 and C_2) were determined based on available Young's modulus (E) of the human buttocks (Todd and Thacker 1994), and the relations; $E=6(C_1+C_2)$ (Gent 2001) and $C_2=0.25 \cdot C_1$ (Vannah and Childress 1996). The inertial parameters and sizes of the limbs were increased proportionally as BMI increased by referring a report for the anthropometry and mass distributions of body segments (Armstrong 1998). Consequentially, two overweight male (BMI 30 and 35, height 1.77 m) and two overweight female (BMI 30 and 35, height 1.52 m) dummy models were developed.

2.2 Model Validation

Several sets of sled test data using overweight post mortem human surrogates (PMHS) (Forman et al. 2009a; Forman et al. 2009b; Michaelson et al. 2008) and standard anthropomorphic test devices (ATD) (Forman et al. 2008) were reported, in which the subjects were seated in the rear seat. Since no test data of overweight PMHS in the front seat is available at present, the test data at the rear seat (Forman et al. 2008; Forman et al. 2009a; Forman et al. 2009b; Michaelson et al. 2008) were used for the validation of the standard and overweight dummy models. The dummy models were seated in the rear seat without airbag, steering wheel, and D-ring. The specific characteristics of the restraint system (e.g., a force-limiting, pretensioning seatbelt system) and crash pulse used in the tests were replicated in MADYMO model. MADYMO version 7.0 was used for model simulations.

2.3 Frontal Crash Simulations

A vehicle frontal impact model was constructed with seatbelt, airbag, and simplified vehicle interior components, based on the FMVSS (Federal Motor Vehicle Safety Standard) 208 frontal test configuration for drivers (NHTSA 2004). A baseline configuration (referred to as Case 1) of occupant restraint systems is as the following. A crash deceleration pulse from a rigid frontal barrier impact (Rouhana 2003) ($V=56$ km/h) was used. The seatbelt was modeled by finite elements and belt segments with webbing elongation setting to 10% at 10 kN force. The seatbelt was equipped with a load limiter with a maximum force of 4 kN. The load limiter was invented to reduce belt-related injuries (e.g., rib fractures, abdominal injuries) by spooling out the seatbelt webbing from the retractor when the belt loads reach a preset maximum force (Brumbelow 2007). The seatbelt was also equipped with buckle and retractor pretensioners, which rewound belt webbing a maximum of 75 mm at 1.4 kN (Forman et al. 2009a) to eliminate belt slack. The friction coefficient between the seatbelt and dummy was set to 0.25 (Rouhana 2003). A single stage airbag firing at 11 ms was used with a mass flow rate curve for deployment (Kiuchi 1998). The friction coefficient between the seat and body components (pelvis, hip and femur) of dummy was set to 0.3. The seatbelt loads and abdominal compression as well as body excursions were measured to investigate the effect of the SAT layer of overweight subjects on crash injuries. CPU time for

simulation up to 120 ms using the overweight dummy was approximately 3 hours, using a 2.0 GHz dual-core processor PC.

2.4 Case study: Effect of Variations of Restraint Systems, Dummy Positions, and Crash Pulses

Based on the Case 1 (baseline) configuration, more simulations were performed to consider a variety of occupant restraint systems, dummy positions, and crash pulses. Six simulations, with three male (BMI 25, 30, and 35) and three female (BMI 22, 30, and 35) dummies, were performed for each case. The cases are (1) the baseline, (2) acceleration magnitude of the crash pulse scaled up 1.2 times, (3) steering wheel tilted 10 degrees downward, (4) airbag depowered by reducing mass flow rate by 20 percent, (5) friction coefficient between the seat and dummy set to 0.5, (6) belt load limit force set to 3 kN, (7) belt load limit force set to 5 kN, (8) no belt load limiter/pre-tensioners, (9) belt elongation set to 10% at 7.5 kN, (10) seat moved backward 5 cm, (11) seat moved forward 5 cm, (12) airbag undeployed, (13) airbag undeployed and crash pulse scaled up 1.2 times, and (14) seatbelt unfastened.

A statistical test was then performed to evaluate the association between BMI and each regional body injury measure described in the next section. Friedman's rank based nonparametric procedure was used for testing a group effect in a blocked design. The blocks here are the 14 cases and the groups are determined by BMI. SAS (SAS Institute Inc., Cary, NC) was used for the test.

2.5 Body Kinematics and Injury Measures

From simulation results, the accelerations, forces, moments, and deflections of the body components were measured, and they were used to assess the following regional body injuries, proposed by NHTSA (National Highway Traffic Safety Administration) (Eppinger et al. 2000). The head injury criterion (HIC) is a measure of the likelihood of head injury based on acceleration ($a(t)$) at the center of gravity of the dummy's head.

$$HIC_{t_2-t_1} = \max_{t_1 < t_2} \left[(t_2 - t_1) \left\{ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right\}^{2.5} \right] \quad (1)$$

where t_1 and t_2 are the selected initial and final times of the interval so as to maximize HIC. The time interval is set to 15 milliseconds to calculate HIC_{15} . The recommended limit of HIC_{15} is 700 for both male and female. The neck injury criteria, called N_{ij} , are critical limits for all four possible modes of neck loading: tension or compression (F_z) combined with either flexion or extension (M_y).

$$N_{ij} = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}} \quad (2)$$

The proposed critical intercept values for axial force (F_{int}) are 6806 N (tension) and 6160 N (compression) for male, and 4287 N (tension) and 3880 N (compression) for female. The proposed critical intercept values for moment (M_{int}) are 310 N (flexion) and 135 N (extension) for male, and 155 N (flexion) and 67 N (extension) for female. The recommended limit of N_{ij} is 1.0.

The resultant chest acceleration (3 ms clip) and maximum chest deflection are used for thoracic injury criteria. The recommended limit of chest acceleration (3 ms clip) is 60 g for both male and female, and the recommended limits of chest deflection are 63 mm for male and 52 mm for female. The lower extremity criterion (LEC) is a measure of femur force for tolerance of leg injury. The recommended limits of femur load are 10 kN for male and 6.8 kN for female. The recommended limits and intercept values are from the NHTSA standards (Eppinger et al. 2000).

3. Results

The fat vests with sectional views compared to the SAT geometry from MRI for male and female are illustrated in Figures 1 and 2, respectively. The SAT region of a female was much thicker than that of a male for the same BMI. The mass distributions of body components and centers of gravity of the dummy models are listed in Table 2 (male) and Table 3 (female).

3.1 Results of Model Validation

Tables 4 and 5 list the comparisons of the head and pelvis accelerations, body excursions, and belt loads of the standard and overweight subjects used in the model validation, respectively. The body kinematics and trajectories of standard and overweight dummies were consistent (within average difference of 10% for standard dummies and 15% for overweight dummies) with those of the PMHS (Forman et al. 2009a; Forman et al. 2009b; Michaelson et al. 2008) and ATD (Forman et al. 2008) used in the experiments.

3.2 Effect of the SAT

Figure 3 shows the comparisons of the kinematics of the standard and overweight dummies in the driver's seat during a frontal crash (Case 1). The body excursion and maximum seatbelt loads are also denoted in the figure. Since an explicit injury criterion for abdomen has not been specified by NHTSA nor implemented in MADYMO, viscous criterion (VC, compression times velocity of deformation) proposed by Viano and Lau (1988) was used to examine the effect of the SAT on abdominal injury. The measuring points for VC calculation are shown in Figure 4 and comparisons of VC curves are shown in Figure 5. The chest deflection (listed in the next section) was also affected by the torso SAT layer.

3.3 Results of the Case Study

The HIC_{15} was measured and listed in Table 6. The values of HIC_{15} were normalized by the recommended limit (Eppinger et al. 2000). In Case 2 (higher crash pulse), all of HIC_{15} were near or over the limit, and HIC_{15} of male BMI 35 dummy was much greater than the limit. The tilted steering wheel (Case 3) didn't have much effect on HIC_{15} . The depowered airbag (Case 4) reduced HIC_{15} , especially HIC_{15} of females. Increasing the friction efficient between the seat and dummy (Case 5) didn't have much effect on HIC_{15} . Decreasing maximum force of load limiter (Case 6) had a positive effect on females but a negative effect on males. Inversely, increasing maximum force of load limiter (Case 7) had a positive effect on males but a negative effect on females. Unequipped load limiter/pre-tensioners (Case 8) significantly increased HIC_{15} of both males and females. Reducing belt strength

(Case 9) increased HIC_{15} of females but didn't have much effect on HIC_{15} of males. The seat position (Cases 10 and 11) affected HIC_{15} of males, but did not much affect HIC_{15} of females. Undeployed airbag (Case 12) reduced HIC_{15} for both males and females except male BMI 35. Undeployed airbag with a higher crash pulse (Case 13), however, significantly increased HIC_{15} of all males and female BMI 35 due to the head contacting with the steering wheel. The Case 14 (no seatbelt) was the worst scenario for males whose heads were hit on the windshield.

The N_{ij} were measured and the maximum N_{ij} in time among the four loading modes are listed in Table 6. The N_{ij} from Case 3 to Case 7 and from Case 9 to Case 11 were not significantly different from ones from Case 1. The higher crash pulse (Case 2 and Case 13) and unequipped load limiter/pretensioners (Case 8) increased N_{ij} for both males and females. Undeployed airbag (Case 12) also increased N_{ij} of male BMI 35 and all of females. The Case 14 (no seatbelt) was the worst, especially to males.

The chest acceleration and deflection were measured and listed in Table 6. The patterns of variations of the chest acceleration were similar to those of HIC_{15} . The chest deflections of all cases were not significantly different from ones of Case 1, except Cases 2, 8, 11 and 14 of males and Cases 4 and 11 of females. The higher crash pulse (Case 2), unequipped load limiter/pretensioners (Case 8) and no seatbelt (Case 14) increased the chest deflection of males. The Case 11 (seat moved forward) reduced the chest deflection of both males and females. The depowered airbag (Case 4) also reduced the chest deflection of females. The LEC were measured and listed in Table 6. The levels of LEC were much lower than the limits for all cases except Case 14.

The mean values of the injury outcomes from Case 1 to Case 13 for each dummy are plotted in Figure 6. The injury outcomes from Case 14 (seatbelt unfastened) were excluded in the mean value calculation because they were extremely higher compared to those from other cases. The result of the Friedman's nonparametric test is listed in Table 7. The Case 14 was excluded in the test. All p-values are less than 0.01 except the N_{ij} .

4. Discussion

This study aimed at examining the injury mechanisms sustained by overweight drivers in frontal motor vehicle crashes using computational models of overweight subjects. Overweight males and females, who differed in weight, height and fat distribution, had corresponding differences in body kinematics and injury severity and pattern.

4.1 Effect of the SAT on Body Injuries

The overweight subjects exhibited greater forward body excursions than did non-overweight ones, as shown in Figure 3 and the sled tests using PMHS (Forman et al. 2009b; Kent et al. 2010). This is due to a greater 'momentum effect' of a greater body mass for both males and females. However, the subjects exhibited different body trajectories (see Figure 3). The torso pitch angles (defined as a ratio of sternum to pelvic excursion in this study) are significantly different between males (1.16 ± 0.01) and females (0.75 ± 0.02). This is mainly due to a greater distance of mass centers between head/upper body and pelvis as well as a greater

mass of upper body components of males. Consequentially, males show a greater upper body excursion compared to females. The less torso pitch of females is due to lower height and lower center of gravity as compared to males (Tables 2 and 3). Females show greatly increased pelvic excursion as BMI increased.

As BMI increased, the seatbelt loads were observed to increase due to a greater momentum effect of increased body mass. The abdominal injury expressed by the maximum abdominal VC, however, was decreased as BMI increased for males, as shown in Figure 5(a), which may be called as a ‘cushion effect’ of the adipose tissue. For females, the maximum VC of BMI 35 was slightly higher than that of BMI 30, in which the momentum effect of increased body mass may partially compensate for the cushion effect. It is worthy to note that the SAT affected the chest deflection differently to males and females, as shown in Table 6 and Figure 6. The chest deflection increased for males but decreased for females, as BMI increased to 35. This is due to the different torso pitch angles described above and a greater cushion effect of the thicker chest SAT layer of females.

4.2 Injury Statistics of the Case Study

Through the injury statistics of the case study, the HIC_{15} were observed to increase (21.5%, see Figure 6) for females ($p=0.00091$, see Table 7) and significantly increase (44.8%) for males ($p=0.00002$) as BMI increased to 35. The results are consistent with the finding that overweight passengers (BMI = 30) are more likely to suffer severe head trauma than non-overweight ones (BMI < 30) after frontal collisions (Tagliaferri et al. 2009). No explicit relationship between the driver obesity and neck injury was found for both males (6.9% increase, $p=0.02491$) and females (3.6% decrease, $p=0.02491$). The driver obesity substantially increased the chest accelerations (13.7% increase for males and 16.4% increase for females) and LEC (169.6% increase for males and 50.2% increase for females), which had strong associations with BMI ($p < 0.0001$). As BMI increased to 35, the chest deflections were observed to increase (13.5%) for males, but not to increase for females. Considering the chest acceleration and deflection as a combined thoracic injury criterion, the results are consistent with the finding that overweight occupants in MVCs had a higher risk of thoracic injury when compared to lean occupants, and the increment was more prominent in males (Cormier 2008).

4.3 Body Excursions versus Injury Outcomes

Additional statistical test using Spearman's rank correlation was performed to evaluate the association between the body excursions (head, sternum, and pelvis excursion) and the injury outcomes (HIC_{15} , N_{ij} , chest acceleration, chest deflection, and LEC). The data from the case studies were used for the analysis. The outcomes from Case 14 were excluded due to the same reason addressed above. The correlation coefficients and p-values are listed in Table 8. For males, strong correlation (correlation coefficient > 0.5, $p < 0.001$) are found between the sternum/pelvic excursions and HIC_{15} , chest acceleration, and LEC. The excessive body excursions induced by the momentum effect of overweight subjects would have a higher probability of excessive impact with the vehicle components (i.e., airbag, steering wheel, knee bolster), which would increase the head, thorax, and lower extremity

injury risks. Such strong correlations were also found among females, between the sternum/pelvic excursions and the LEC and between the pelvic excursion and chest acceleration.

4.4 Limitations

There are some limitations of this study. First, a computational model of 50th percentile (average size) female dummy is not available at present. Therefore, the results from the overweight female dummies, which were based on the Hybrid III 5th percentile female dummy whose height is 1.52 m, cannot represent injury patterns sustained by women of average height. Second, the Hybrid III ellipsoidal dummy models were used as baseline models. Future efforts could go towards a more robust validation and simulation with more 'human-like' models such as full finite element models (particularly for chest/abdomen deflection analysis with an anatomically accurate torso model), when fully-validated models for both male and female are available. Third, overweight dummy models up to BMI level of 35 were developed in this study due to limited availability of MRI data sets. Modeling of morbidly obese subjects (BMI = 40) would enable us to closely examine the cushion effect of the fat layers as well as a more momentum effect. Fourth, no additional adipose tissue and mass were added to the neck and head of the overweight dummies due to lack of such information. The adipose tissues in the neck and head, however, would not have significant effect on the injury severity. Fifth, the intercept values and recommended limits for the standard dummies were used for the overweight dummies in the injury analysis. Lastly, this study considered only frontal crashes. Nevertheless, this study is the first attempt (to our knowledge) to computationally assess the risk of injury sustained by overweight occupants.

5. Conclusion

In summary, the torso subcutaneous adipose tissue of overweight subjects contributed to reduce abdominal injury. However, the momentum effect of a greater body mass of overweight subjects was more dominant over the cushion effect of the adipose tissue, increasing risk of other regional body injuries except abdomen. Strong correlations ($p < 0.01$) were found between BMI and regional body injuries except neck injury. A greater momentum of overweight males resulted in greater forward torso and pelvic excursions that accounted for higher risks ($p < 0.001$) of head, thorax, and lower extremity injury than observed in non-overweight males. Such strong correlations were also found among females, between the sternum/pelvic excursions and the lower extremity injury and between the pelvic excursion and chest acceleration.

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References

- Agache P, Monneur C, Leveque JL, De Rigal J. Mechanical properties and Young's modulus of human skin in vivo. *Arch Derm Res.* 1980; 269:221–232. [PubMed: 7235730]
- Alekseev SI, Radzievsky AA, Logani MK, Ziskin MC. Millimeter wave dosimetry of human skin. *Bioelectrom.* 2008; 29(1):65–70.

- Arbabi S, Wahl WL, Hemmila MR, Kohoyda-Inglis C, Taheri PA, Wang SC. The cushion effect. *J Trauma*. 2003; 54(6):1090–1093. [PubMed: 12813327]
- Armstrong, HG. Anthropometry and mass distribution for human analogues, Vol. I. Military male aviators. Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base; Ohio: 1998. p. AAMRL-TR-88-010.
- Boulanger BR, Milzman D, Mitchell K, Rodriguez A. Body habitus as a predictor of injury pattern after blunt trauma. *J Trauma*. 1992; 33:228–232. [PubMed: 1507286]
- Brumbelow, ML.; Baker, BC.; Nolan, JM. Effects of seat belt load limiters on driver fatalities in frontal crashes of passenger cars.. Proceedings of 20th International Technical Conference on the Enhanced Safety of Vehicles; Arlington, VA. 2007; p. 07-0067.
- Choban PS, Weireter LJ, Maynes C. Obesity and increased mortality in blunt trauma. *J Trauma*. 1991; 31:1253–1257. [PubMed: 1920556]
- Cormier JM. The influence of body mass index on thoracic injuries in frontal impacts. *Accid Anal Prev*. 2008; 40(2):610–615. [PubMed: 18329413]
- Eppinger, R.; Sun, E.; Kuppa, S.; Saul, R. Technical Report. U.S. Department of transportation; Washington, DC.: 2000. Supplement: Development of improved injury criteria for the assessment of advanced automotive restraint systems - II, National Highway Traffic Safety Administration..
- Eppinger, RH.; Marcus, J.; Morgan, R. Development of dummy and injury index for NHTSA's thoracic side impact protection research program. SAE; 1984. p. 840885
- Fidanza F, Keys A, Anderson JT. Density of body fat in man and other mammals. *J Appl Physiol*. 1953; 6:252–256. [PubMed: 13108819]
- Forman J, Lopez-Valdes F, Lessley D, Kindig M, Kent R, Ridella S, Bostrom O. Rear seat occupant safety: An investigation of a progressive force-limiting, pretensioning 3-point belt system using adult PMHS in frontal sled tests. *Stapp Car Crash J*. 2009a; 53:49–74. [PubMed: 20058550]
- Forman J, Lopez-Valdes FJ, Lessley D, Kindig M, Kent R, Bostrom O. The effect of obesity on the restraint of automobile occupants. *Annu Proc Assoc Adv Automot Med*. 2009b; 53:25–40.
- Forman J, Michaelson J, Kent R, Kuppa S, Bostrom O. Occupant restraint in the rear seat: ATD responses to standard and pre-tensioning, force-limiting belt restraints. *Annu Proc Assoc Adv Automot Med*. 2008; 52:141–154.
- Gent, AN. Engineering with rubber: How to design rubber components. 2nd edition. Carl Hanser Verlag; Munich: 2001.
- Kent RW, Forman JL, Bostrom O. Is there really a “cushion effect”? A biomechanical investigation of crash injury mechanisms in the obese. *Obesity*. 2010; 18(4):749–753. [PubMed: 19798067]
- Kiuchi, T. A simulation study on inflation induced injury and NCAP with depowered airbag.. Proceedings of the 16th International Technical Conference on the Enhanced Safety of Vehicles Windsor; Ontario. 1998; National Highway Traffic Safety Administration; p. 2518-2525.
- Michaelson J, Forman J, Kent R, Kuppa S. Rear seat occupant safety: Kinematics and injury of PMHS restrained by a standard 3-point belt in frontal crashes. *Stapp Car Crash J*. 2008; 52:295–325. [PubMed: 19085167]
- Mock CN, Grossman DC, Kaufman RP, Mack CD, Rivara FP. The relationship between body weight and risk of death and serious injury in motor vehicle crashes. *Accid Anal Prev*. 2002; 34(2):221–228. [PubMed: 11829292]
- Mokdad AH, Ford ES, Bowman BA, Dietz WH, Vinicor F, Bales VS, Marks JS. Prevalence of obesity, diabetes, and obesity-related health risk factors, 2001. *J Am Med Assoc*. 2003; 289:76–79.
- Neville AL, Brown CV, Weng J, Demetriades D, Velmahos GC. Obesity is an independent risk factor of mortality in severely injured blunt trauma patients. *Arch Surg*. 2004; 139(9):983–987. [PubMed: 15381617]
- NHTSA (National Highway Traffic Safety Administration). Federal motor vehicle safety standard, Sec. 571.208 Standard No. 208 - Occupant crash protection. Department of Transportation; 2004.
- Peden, M., et al. World report on road traffic injury prevention: Summary. World Health Organization; Geneva: 2004.

- Rouhana SW, Bedewi PG, Kankanala SV, Prasad P, Zwolinski JJ, Medufsky A, Rupp J, Jeffreys T, Schneider L. Biomechanics of 4-point seat belt systems in frontal impacts. *Stapp Car Crash J*. 2003; 47:367–399. [PubMed: 17096257]
- Seidenari S, Giusti G, Bertoni L, Magnoni C, Pellacani G. Thickness and echogenicity of the skin in children as assessed by 20-MHz ultrasound. *Dermatology*. 2000; 201(3):218–222. [PubMed: 11096192]
- Shen W, Punyanitya M, Wang Z, Gallagher D, St-Onge MP, Albu J, Heymsfield SB, Heshka S. Total body skeletal muscle and adipose tissue volumes: Estimation from a single abdominal cross-sectional image. *J Appl Physiol*. 2004; 97:2333–2338. [PubMed: 15310748]
- Shen W, Chen J. Application of imaging and other noninvasive techniques in determining adipose tissue mass. *Meth Mol Biol*. 2008; 456:39–54.
- Shih AM, Yu TY, Gopalsamy S, Ito Y, Soni BK. Geometry and mesh generation for high fidelity computational simulations using non-uniform rational B-splines. *Appl Num Math*. 2005; 55(3): 368–381.
- Tagliaferri F, Compagnone C, Yoganandan N, Gennarelli T. Traumatic brain injury after frontal crashes: Relationship with body mass index. *J Trauma*. 2009; 66(3):727–729. [PubMed: 19276745]
- TNO.. MADYMO Theory Manual, release 7.0. Delft; The Netherlands: 2008.
- Todd BA, Thacker JG. Three-dimensional computer model of the human buttocks in vivo. *J Rehabil Res Dev*. 1994; 31(2):111–119. [PubMed: 7965867]
- Vannah WM, Childress DS. Indentor tests and finite element modeling of bulk muscular tissue in vivo. *J Rehabil Res Dev*. 1996; 33(3):239–252. [PubMed: 8823672]
- Viano DC, Lau IV. A viscous tolerance criterion for soft tissue injury assessment. *J Biomech*. 1988; 21(5):387–399. [PubMed: 3417691]
- Wang SC, Bednarski B, Patel S, Yan A, Kohoyda-Inglis C, Kennedy T, Link E, Rowe S, Sochor M, Arbabi S. Increased depth of subcutaneous fat is protective against abdominal injuries in motor vehicle collisions. *Annu Proc Assoc Adv Automot Med*. 2003; 47:545–559. [PubMed: 12941250]
- Whitlock G, Norton R, Clark T, Jackson R, MacMahon S. Is body mass index a risk factor for motor vehicle driver injury? A cohort study with prospective and retrospective outcomes. *Int J Epid*. 2003; 32:147–149.
- World Health Organization. Joint WHO/FAO expert consultation on diet, nutrition and the prevention of chronic diseases. 2003 WHO Technical Report Series 916.
- Zheng Y, Mak AFT, Lue B. Objective assessment of limb tissue elasticity: Development of a manual indentation procedure. *J Rehabil Res Dev*. 1999; 36(2):71–85. [PubMed: 10661523]
- Zhu S, Layde PM, Guse CE, Laud PW, Pintar F, Nirula R, Hargarten S. Obesity and risk for death due to motor vehicle crashes. *Am J Pub Health*. 2006; 96(4):734–739. [PubMed: 16537660]
- Zhu S, Kim JE, Ma X, Shih AM, Laud PW, Pintar F, Shen W, Heymsfield SB, Allison DB. BMI and risk of serious upper body injury following motor vehicle crashes: Concordance of real-world and computer-simulated observations. *PLoS Med*. 2010; 7(3):e1000250. [PubMed: 20361024]

Figure 1.
Comparison of axial planes between SAT geometry from MRI and fat vest for male BMI 35 dummy

Figure 2.
Comparison of axial planes between SAT geometry from MRI and fat vest for female BMI 35 dummy

Figure 3.
Comparison of the kinematics of standard and obese dummies for male (left column) and female (right column) during a frontal crash (Case 1): The snap-shots are captured at a moment in which the head excursion is the maximum. The body excursion data are maximums in time

Figure 4. Abdominal compression and measuring points (circles). The arms are hidden for clear view: (a) at initial time; (b) at time when abdominal compression is a maximum.

Figure 5.
Comparison of viscous criterion for abdomen: (a) male; (b) female

Figure 6.
Mean values of each injury outcome from Case 1 to Case 13: (a) male; (b) female

Table 1

Material properties used for the SAT (subcutaneous adipose tissue) and skin

Constitutive model		Properties	References
SAT	Hyperelastic (Mooney-Rivlin)	C_1 : 8.64 kPa, C_2 : 2.16 kPa (male)	Todd and Thacker (1994)
		C_1 : 6.33 kPa, C_2 : 1.58 kPa (female)	Todd and Thacker (1994)
		Density: 900.7 kg/m ³ (male)	Fidanza et al. (1953)
		Density: 899.5 kg/m ³ (female)	Fidanza et al. (1953)
Skin	Isotropic elastic	Thickness: 2 mm	Seidenari et al. (2000)
		Young's modulus: 635 kPa	Agache et al. (1980)
		Poisson's ratio: 0.45	Zheng et al. (1999)
		Density: 1000 kg/m ³	Alekseev et al. (2008)

Table 2

Mass (kg) distribution and center of gravity of male dummies (height 1.77 m)

Component	BMI 25		BMI 30		BMI 35	
	Mass	%	Mass	%	Mass	%
Head & Neck	5.45	8.0%	5.45	5.9%	5.45	5.0%
Thorax & Spine	19.66	25.1%	19.66	21.2%	22.89	20.8%
Pelvis & Abdomen	17.25	22.0%	17.25	18.6%	19.06	17.3%
Upper extremity	8.74	11.1%	9.99	10.8%	12.01	10.9%
Lower extremity	27.33	34.8%	30.97	33.4%	37.13	33.8%
Fat vest	-	-	9.38	10.1%	13.48	12.3%
Total	78.43	100.0%	92.70	100.0%	110.02	100.0%
Center of gravity ^a (cm)	92.5		93.7		94.6	

^a measured from bottom

Table 3

Mass (kg) distribution and center of gravity of female dummies (height 1.52 m)

Component	BMI 22		BMI 30		BMI 35	
	Mass	%	Mass	%	Mass	%
Head & Neck	4.41	9.0%	4.41	6.4%	4.41	5.5%
Thorax & Spine	13.33	27.1%	13.33	19.2%	13.52	16.7%
Pelvis & Abdomen	10.31	20.9%	10.31	14.9%	10.45	12.9%
Upper extremity	4.80	9.7%	5.86	8.4%	8.04	10.0%
Lower extremity	16.42	33.3%	21.22	30.6%	27.20	33.7%
Fat vest	-	-	14.26	20.6%	17.17	21.3%
Total	49.27	100.0%	69.39	100.0%	80.79	100.0%
Center of gravity ^a (cm)	75.7		77.6		75.7	

^a measured from bottom

Table 4

Comparison of maximum accelerations, body excursions and belt loads of the standard subjects between the experiment and simulation

	Male		Female	
	Exp ^a	Simulation ^b	Exp ^c	Simulation ^d
HIC ₁₅	227±163	186.7	250±51	237.3
Head accel. resultant 3 ms clip (g)	49±14	41.3	51±4.2	50.1
Pelvis accel. resultant 3 ms clip (g)	40±3.5	36.7	45±2.4	38.1
Head excursion ^e (cm)	57±2.3	54.1	n/a ^f	44.1
Shoulder excursion ^e (cm)	44±4.1	37.4	n/a	30.5
Pelvis excursion ^e (cm)	18±4.3	19.0	n/a	23.3
Knee excursion ^e (cm)	21±8.1	19.2	n/a	23.0
Upper shoulder belt tension (kN)	4.4±0.13	5.0	3.9±0.19	3.7
Lower shoulder belt tension (kN)	n/a	4.6	3.2±0.04	3.0
Lap belt tension (kN)	4.7±0.3	5.7	5.1±0.12	4.8

^a Average ± standard deviation from the PMHS test # 1386, 1387, 1389 (males) (Forman et al. 2009a), Data scaled for 50th percentile male, height 1.75 m, weight 77 kg, BMI 25

^b Standard male dummy, height 1.77 m, weight 78.4 kg, BMI 25

^c Average ± standard deviation from the Hybrid III 5th percentile female ATD tests (Forman et al. 2008), height 1.52 m, weight 50 kg, BMI 22

^d Standard female dummy, height 1.52 m, weight 49 kg, BMI 22

^e The excursions were measured along forward direction

^f Not available

Table 5

Comparison of maximum accelerations, body excursions and belt loads of the obese subjects between the experiment and simulation

	Exp ^a	Male		Exp ^d	Female	
		Simulation ^b			Simulation ^e	
		unscaled	scaled ^c		unscaled	scaled ^f
HIC ₁₅	n/a ^g	251.7	230.4	295.0	343.7	286.9
Head accel. resultant 3 ms clip (g)	n/a	53.0	50.9	56.6	57.8	52.4
Pelvis accel. resultant 3 ms clip (g)	n/a	29.5	28.3	23.7	33.2	30.1
Head excursion ^h (cm)	79.5	56.6	60.4	46.9	39.1	42.4
Shoulder excursion ^h (cm)	68.6	41.5	44.3	29.9	20.9	22.7
Pelvis excursion ^h (cm)	37.2	36.1	38.5	35.8	28.2	30.6
Knee excursion ^h (cm)	46.3	37.5	40.1	27.9	21.3	23.1
Upper shoulder belt tension (kN)	6.43	5.24	5.24 ⁱ	6.22	5.27	6.43
Lower shoulder belt tension (kN)	6.29	5.98	5.98 ⁱ	4.82	4.52	5.51
Lap belt tension (kN)	8.29	8.97	8.97 ⁱ	5.97	5.83	7.11

^aThe PMHS test # 1333 (male) (Forman et al. 2009b), subject height 1.89 m, weight 124 kg, BMI 35

^bObese male dummy, height 1.77 m, weight 110 kg, BMI 35

^cScaling data using the method (Forman et al. 2009a; Eppinger et al. 1984), considering the PMHS subject as a target anthropometry. Scaling factor: $\lambda_m=(124\text{ kg}/110\text{ kg})^{1/3}$, $\lambda_L=(1.89\text{ m}/1.77\text{ m})$

^dThe PMHS test # 1263 (female) (Michaelson et al. 2008), subject height 1.65 m, weight 109 kg, BMI 40. No force-limiting, pretensioning seatbelt system was used.

^eObese female dummy, height 1.52 m, weight 81 kg, BMI 35. No force-limiting, pretensioning seatbelt system was used.

^fScaling data using the method (Forman et al. 2009a; Eppinger et al. 1984), considering the PMHS subject as a target anthropometry. Scaling factor: $\lambda_m=(109\text{ kg}/81\text{ kg})^{1/3}$, $\lambda_L=(1.65\text{ m}/1.52\text{ m})$

^gNot available

^hThe excursions were measured along forward direction

ⁱThese forces were not scaled since load limiter was used.

Table 6

Injury measures

Injury/Dummy	BMI	case 1	case 2	case 3	case 4	case 5	case 6	case 7	case 8	case 9	case 10	case 11	case 12	case 13	case 14
HIC₁₅^a Male	25	0.54	1.01	0.53	0.49	0.53	0.65	0.51	0.81	0.58	0.49	0.59	0.28	1.09	3.07
	30	0.57	1.01	0.58	0.50	0.59	0.64	0.58	0.92	0.58	0.53	0.66	0.29	0.94	2.97
	35	0.73	1.28	0.68	0.65	0.72	0.82	0.69	1.05	0.79	0.63	0.86	0.75	2.09	2.48
HIC₁₅^a Female	22	0.67	1.04	0.67	0.44	0.69	0.56	0.70	0.81	0.71	0.60	0.68	0.22	0.46	0.63
	30	0.69	1.01	0.67	0.50	0.66	0.61	0.82	0.90	0.81	0.63	0.66	0.30	0.83	0.67
	35	0.70	1.12	0.70	0.49	0.71	0.69	0.83	0.97	0.85	0.76	0.65	0.35	1.22	0.69
N_{ij}^b Male	25	0.32	0.38	0.32	0.31	0.33	0.37	0.31	0.44	0.34	0.32	0.34	0.34	0.53	1.61
	30	0.33	0.37	0.34	0.28	0.32	0.36	0.32	0.42	0.37	0.33	0.34	0.33	0.49	1.58
	35	0.32	0.45	0.32	0.31	0.38	0.38	0.32	0.44	0.34	0.37	0.39	0.42	0.54	0.99
N_{ij}^b Female	22	0.42	0.47	0.44	0.36	0.44	0.39	0.42	0.48	0.42	0.42	0.44	0.48	0.50	0.60
	30	0.41	0.46	0.42	0.33	0.39	0.38	0.42	0.48	0.42	0.40	0.37	0.54	0.48	1.09
	35	0.41	0.50	0.39	0.33	0.40	0.37	0.43	0.47	0.40	0.42	0.36	0.57	0.41	1.15
CA^c Male	25	0.53	0.71	0.53	0.52	0.56	0.59	0.57	0.66	0.58	0.59	0.58	0.52	0.67	1.17
	30	0.60	0.74	0.59	0.56	0.55	0.64	0.59	0.72	0.64	0.59	0.60	0.52	0.70	1.15
	35	0.61	0.83	0.61	0.59	0.64	0.68	0.60	0.77	0.66	0.69	0.59	0.57	0.80	0.98
CA^c Female	22	0.66	0.75	0.67	0.59	0.65	0.57	0.66	0.65	0.66	0.60	0.70	0.56	0.61	1.07
	30	0.67	0.85	0.74	0.65	0.68	0.66	0.76	0.79	0.75	0.62	0.73	0.59	0.71	0.81
	35	0.78	0.89	0.76	0.68	0.66	0.70	0.81	0.87	0.80	0.68	0.72	0.62	0.74	0.76
CD^d Male	25	0.56	0.61	0.56	0.55	0.54	0.55	0.59	0.65	0.58	0.57	0.52	0.55	0.59	0.67
	30	0.52	0.58	0.53	0.50	0.48	0.55	0.52	0.59	0.52	0.53	0.46	0.45	0.50	0.60
	35	0.63	0.75	0.62	0.60	0.67	0.63	0.65	0.72	0.63	0.67	0.54	0.59	0.70	0.63
CD^d Female	22	0.64	0.68	0.65	0.60	0.63	0.60	0.65	0.66	0.64	0.62	0.64	0.58	0.63	0.63
	30	0.71	0.78	0.70	0.62	0.68	0.77	0.74	0.73	0.68	0.59	0.59	0.67	0.69	0.60
	35	0.64	0.63	0.67	0.55	0.61	0.63	0.65	0.61	0.62	0.64	0.54	0.58	0.75	0.68

Injury/Dummy	BMI	case 1	case 2	case 3	case 4	case 5	case 6	case 7	case 8	case 9	case 10	case 11	case 12	case 13	case 14
LEC ^e Male	25	0.13	0.13	0.13	0.13	0.11	0.13	0.13	0.13	0.11	0.11	0.15	0.13	0.13	0.65
	30	0.12	0.22	0.13	0.12	0.14	0.18	0.13	0.17	0.18	0.14	0.24	0.13	0.18	0.97
	35	0.32	0.42	0.33	0.32	0.22	0.39	0.30	0.33	0.42	0.19	0.39	0.33	0.45	1.01
LEC ^e Female	22	0.19	0.18	0.19	0.19	0.23	0.19	0.19	0.20	0.18	0.25	0.16	0.19	0.19	0.70
	30	0.23	0.35	0.23	0.25	0.23	0.29	0.24	0.27	0.28	0.21	0.25	0.26	0.35	0.85
	35	0.30	0.34	0.31	0.31	0.28	0.29	0.30	0.27	0.29	0.28	0.26	0.32	0.25	0.89

^a HIC₁₅, Head injury criterion, normalized by 700

^b Nij, Neck injury criterion

^c C_A, Chest acceleration (g), normalized by 60 g for male and female

^d C_D, Chest deflection (mm), normalized by 63 mm for male and 52 mm for female

^e LEC, Lower extremity criteria, femur load normalized by 10 kN for male and 6.8 kN for female

Table 7

p-values from Friedman's nonparametric test for testing the null hypothesis that BMI is not related to each injury outcome

	Male	Female
HIC ₁₅	0.00002	0.00091
N _{ij}	0.02491 ^a	0.02491 ^a
Chest acceleration	0.00007	0.00001
Chest deflection	0.00000	0.00917
LEC	0.00004	0.00019

^a
p > 0.01

Table 8

Spearman's correlation coefficients (N=39) and p-values (testing zero correlation) between the body excursions and injury outcomes

	Body excursion	Male			Female		
		Head	Sternum	Pelvis	Head	Sternum	Pelvis
HIC ₁₅	coefficient	0.2617	0.5113	0.6322	-0.0832	0.2338	0.4164
	p-value	0.1076	0.0009	<0.0001	0.6146	0.1520	0.0084
N_{ij}	coefficient	0.3510	0.4410	0.4402	0.2585	-0.0075	-0.1115
	p-value	0.0284	0.0050	0.0050	0.1120	0.9639	0.4992
Chest acceleration	Coefficient	0.2718	0.5492	0.7137	-0.2221	0.3533	0.6027
	p-value	0.0942	0.0003	<0.0001	0.1742	0.0274	<0.0001
Chest deflection	coefficient	0.3125	0.4403	0.6238	-0.1215	0.1064	0.1562
	p-value	0.0528	0.0050	<0.0001	0.4614	0.5192	0.3424
LEC	coefficient	0.3283	0.6071	0.6032	0.3532	0.7900	0.7732
	p-value	0.0413	<0.0001	<0.0001	0.0274	<0.0001	<0.0001

Bold font: strong correlation (correlation coefficient > 0.5, p < 0.001)