

Comparison of Whole Body Kinematics between Fracture and Non-Fracture Finite Element Human Body Models during Side Impact

Yasuki Motozawa, Masayoshi Okamoto, Fumie Mori*

Abstract The age-specific finite element human body models representing rib fracture were developed by the authors. They are limited, however, in quantitative evaluation of a strain that exceeds the predetermined threshold, i.e. the strain measured on the “fracture” model is censored. A “non-fracture” model can evaluate the value of the strain throughout the impact, but may be at a disadvantage when trying to represent post-fracture kinematics. The objective of the present study is to examine the compatibility between “fracture” and “non-fracture” finite element (FE) human body models in order to establish the method to quantify the probability of the rib fracture. The authors conducted a series of simulated side-impact tests using human body models with and without representation of the rib fracture. Simulations were done in compliance with the IIHS SICE protocol. The results indicate that the whole-body kinematics of the “non-fracture” human models was almost identical to those of “fracture” models. The locations in the “non-fracture” models, where the values of the strain reached the pre-determined threshold, were well correlated with those where the element elimination was activated in the “fracture” models. The results suggested the feasibility of the methods for probability-based evaluation of the rib fracture using human FE models.

Keywords human FE model, rib fracture, side impact, post-fracture, probability.

I. INTRODUCTION

Recently, age-related injury, especially rib fracture during vehicle impact, has become a social issue because of its serious outcome despite considerable low injury severity in AIS scale [1]. In order to establish relevant countermeasures to address this issue, such as restraint systems with an enhanced performance, a methodology is required to predict or evaluate the risk of rib fracture appropriately. Acceleration-based injury indices have been widely applied to the evaluation as the indices to represent “global” injury severity, but there have been many discussions regarding more appropriate indices that enable quantitative prediction of actual injury [2-5]. Among them, the maximum value of thoracic deflection is currently regarded as one of the most relevant indices [6-7]. However, because those indices were originally developed for the Anthropomorphic Testing Device (hereafter, ATD) measurement, being statistically but not directly correlated to the dominant physical quantity of the fracture, there is a strong limitation in the prediction of the specific rib fracture yielded due to local loading. Most of the currently proposed countermeasures were devised to distribute, or to bypass thoracic loading by establishing alternative load-path on the occupants’ less fragile body regions [8]. However, ATD-based measurements have limited sensitivity to the local loading condition due to limited biofidelity of ATD in skeletal structure. In other words, conventional “global” indices, i.e. ATD, could not sufficiently evaluate the effectiveness of the countermeasures, even if they could facilitate the optimisation of the boundary condition between the restraint system and the occupant’s body. Therefore, the development of human body models with higher biofidelity, which can represent realistic loading condition on the rib cage, has been a priority in the engineering field of vehicle impact safety.

In their previous study, the authors established age-specific finite element (hereafter, FE) human body models, which can represent rib fracture using the element elimination function of the FE code [9-13]. Although these models enable us to represent post-fracture kinematics as well as the fracture itself during impacts, the element elimination is activated by a pre-determined threshold of the maximum strain, so that the strain that exceeds the threshold cannot be quantitatively evaluated, i.e. the strain measured on the models is censored

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once fracture occurs. Accordingly, to quantify the severity of the applied impact to the whole body, in other words, the total risk of the fracture, the methods based on uncensored measurement of the maximum strain on the models should be considered, because it can be directly correlated to the probability of the fracture. It is deemed that a human body model without fracture representation allows for an evaluation of the magnitude of the strain throughout the impact. Recently, some researches were conducted to predict a thoracic injury risk using FE human body models [14-16]. Among them, Forman et al. proposed the method to quantify the total risk of the rib-fracture by using “non-fracture” FE human body models [16]. To the authors’ knowledge, this is the first attempt to establish a probability-based method to assess the rib-fracture using a human body model. However, as the method focuses mainly on the frontal impacts and stands on the assumption that the kinematics of the upper torso is considerably stable and occurrence of the fracture has a limited effect on the global loading conditions, it is deemed to have limitations in representing post-fracture conditions, such as the change of the distribution of the strain due to the “chain-reaction” initiated by the first fracture. Moreover, the kinematics during a side impact may be affected by the fracture because the occupant’s torso is limitedly constrained by the restraint system along the lateral direction. The objective of the present study is to compare the whole body kinematics of the “fracture” and “non-fracture” FE human body models in order to examine the feasibility of the probability-based methods to quantify the risk of the rib fracture using FE human body models, especially during side impacts. The two human FE models used in the present study represent the anthropometry and material property of 50th percentile American males (hereafter, AM 50 percentile) aged 35 years old and 75 years old (hereafter, YO), respectively. The thoracic responses of the two models were compared and analysed.

II. METHODS

Human FE Models

The human FE models used in the present study represent the anthropometry and material properties of AM 50 percentile aged 35 YO and 75 YO (hereafter, adult FE model and elderly FE model, respectively). The models incorporated the lower limbs, lumbar spine and thorax models previously developed by Ito et al. and Dokko et al. [9-13]. In order to determine the geometry of the 75 YO rib-cage model, a specific CT image for relevant age range that approximates AM 50 percentile was extracted from the medical database at the University of Michigan Program for Injury Research and Education (UMPIRE). Based on the image, the FE mesh of the 75 YO rib-cage model was created. In the model, the element size of the bones was between 2mm to 7mm. The model was compared to the statistical data from the previous study conducted by Kent et al. [17] and Gayzik et al. [18] and was found to indicate close match to the average geometry of the relevant age group. Whereas no appropriate CT image of 35 YO male was available in the database, thus the authors translated the 75 YO rib-cage model to 35 YO rib-cage model by using translation vectors derived from the formula proposed by Gayzik et al. [18]. The remaining body regions, i.e. the head, neck and upper extremities, were supplemented by the H-Model™, which is a commercially available generic human body FE model consisting of 304,390 elements [19]. Figure 4 indicates the perspective view of the adult FE model, along with the geometry of the rib cage in the adult and elderly FE models.

The range of the values for the material parameters – the Young’s modulus, yield stress, ultimate stress and strain of the femur, tibia, fibula, lumbar spine, clavicle, rib, sternum, costal cartilage and intercostal muscle – were determined based on the literature previously published. The material properties of the simplified internal organ models were determined in order to represent a spring-dashpot characteristics under dynamic loading based on the literature, resulting in relevant thoracic response. Table A1 indicates the representative material properties of these models. PAM-CRASH™ (ESI Group, 100-102 Avenue de Suffren 75015 Paris, France) was used as the FEM solver. The responses of the thoracic component of both human FE models during pendulum impacts were validated against the results of published experiments performed by Kroell et al. [20-21]. The responses during a table-top thoracic belt loading were also validated against the experiments performed by Lessley et al. [22]. The authors evaluated the differences between the simulation and the experiments using a ranking system to assess the biofidelity of the model according to the methodology proposed by Rhule et al. [23-24]. They compared the time histories of the displacement of the head, vertebral bodies and pelvis of the

adult human FE model to the averages of the sled experiments. Consequently, they successfully validated sufficient biofidelity of the models during frontal and side impacts. Table A2 indicates the biofidelity scores.

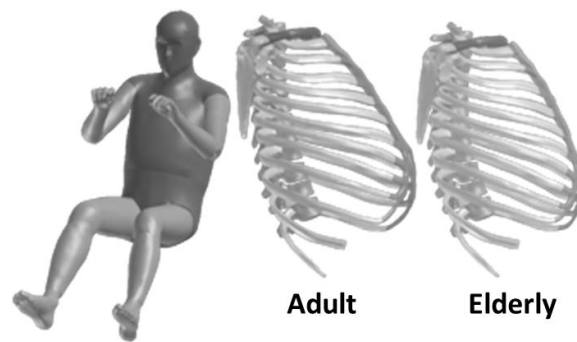


Fig. 1. Perspective view of adult FE model, along with geometry of rib-cage in adult and elderly models.

In the “fracture” models used in the present study, the element elimination option in PAM CRASH code is applied only to the cortical bones. For the bone elements, Material Type 143 (shell) and Type 36 (solid) based on elastic-plastic formulation in PAM CRASH code were applied, which is defined by stress-strain curve. The element elimination is activated when the principal strain of the respective element exceeds a pre-determined threshold i.e. the ultimate plastic strain. The meshes of the ribs were appropriately re-sized in order to represent the fracture trends of the PMHS testing through the validation [13].

Test Set-up FE Models

The authors developed a FE model of the test set-up for a vehicle to moving deformable barrier (hereafter MDB) side-impact experiment. The model consists of a full-scale FE model of the vehicle body (hereafter, vehicle model) and the MDB.

The vehicle model represents a test set-up of a small-size, sedan-type volume production vehicle (the Honda Civic), consisting of a whole-vehicle sheet-metal body structure (hereafter, white-body), a frontal seat equipped with a thoracic side airbag, doors with plastic door trims and a set of seatbelts. However, the spool and the tension retractor of the seatbelt were omitted to obtain stable boundary conditions. Drive train components with an engine, the front and rear suspension systems with tyres, an engine hood, a trunk lid, glazing, bumper-face and interior trims other than door trims were omitted, but the respective masses of the omitted components were applied to the vehicle model so that the total mass and inertia of the vehicle was maintained. The white-body structure, doors and door trims were modelled as a shell element. The seatbelts were modelled using bar and membrane elements in which force-elongation properties of the actual material of the webbing used in the small-size production vehicle were reflected. The padding of the front seat was modelled as a solid element and sculptured along the surface of the human FE models in the relevant seating position.

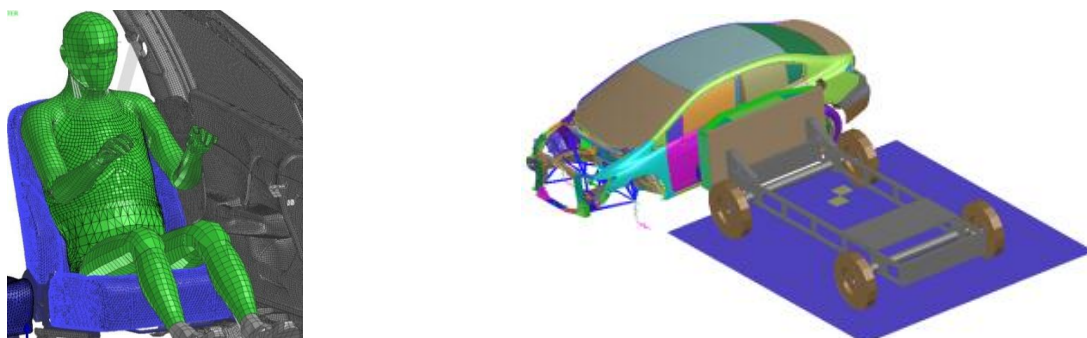


Fig. 2. Test set-up FE model accommodating the human FE model; *left figure* indicates perspective view of adult FE model accommodated in vehicle model, *right image* indicates perspective view of entire test set-up, consisting of vehicle model and MDB model based on SICE protocol.

The MDB model used in the present study was modelled to represent the MDB defined by the SICE (Side Impact Crashworthiness Evaluation Crash Test Protocol) protocol determined by IIHS (Insurance Institute for Highway Safety). The deformable structure (aluminium honeycomb) in the front end of the MDB was modelled as a solid element, but other structure was modelled as a rigid body for simplification.

In advance to the FE analysis using FE human body model, the authors conducted a FE simulation using the test set-up FE model accommodating a SID-II's ATD FE model in the same manner as the SICE protocol, and the time histories of the local velocity of the vehicle model, the velocity and the deflection of each rib of the SID-II's ATD FE model during impact were calculated and compared to those of the experiment using an actual test set-up identical to the model. Figures 3 and 4 indicate the comparisons of the time histories of the local velocity of the vehicle. Since the results of the FE simulation indicated close match to the experimental results, the author deemed that the entire test set-up model was validated against the actual experiment.

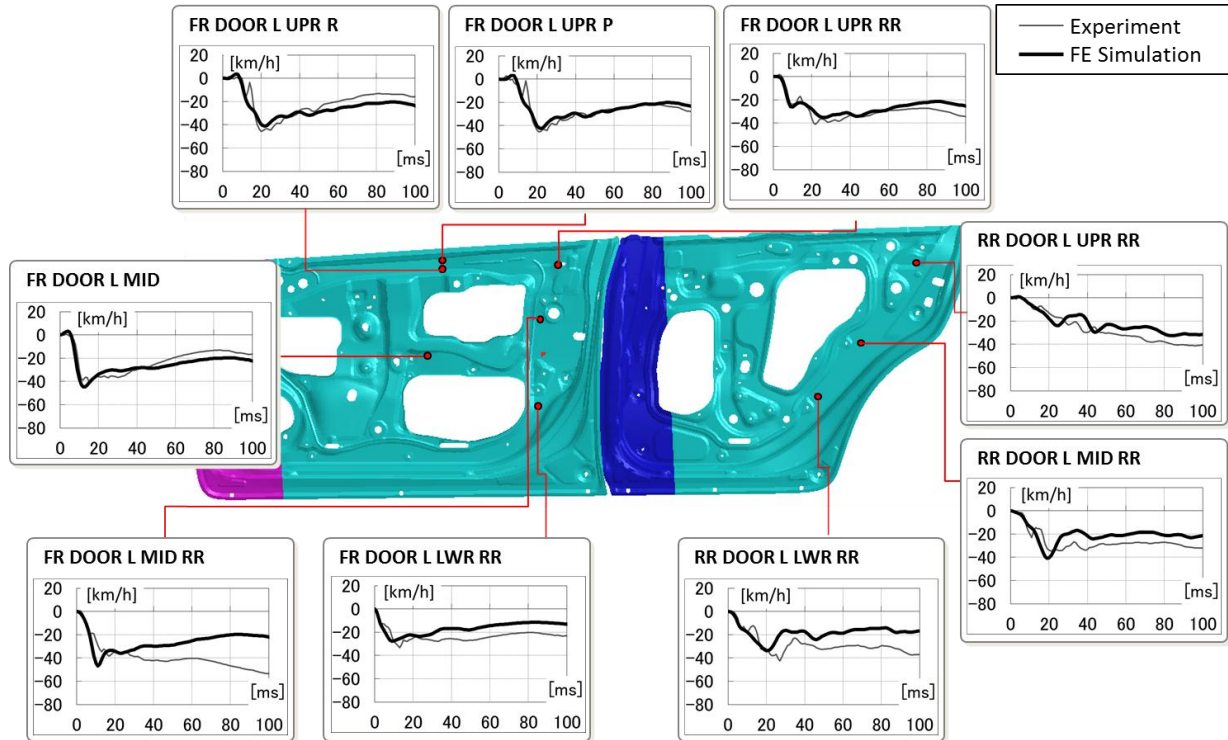


Fig. 3. Comparison of time histories of local velocity measured on doors between experiment and FE simulation.

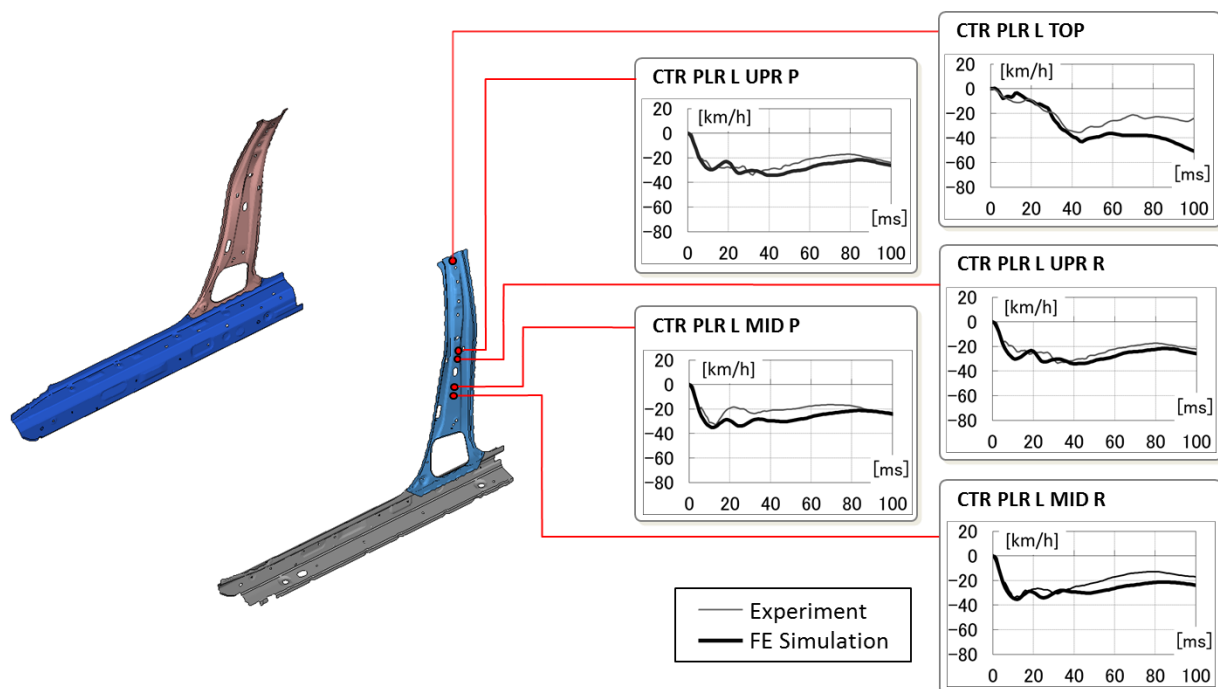


Fig. 4. Comparison of time histories of local velocity measured on B pillar between experiment and FE simulation.

Conditions for FE Simulation

The MDB model was collided to a stationary vehicle model accommodating the human FE model with an impact velocity of 50 km/h on the driver side at a 90 degree angle in the same manner as the SICE protocol. The human model was accommodated on the driver's seat (left-side, i.e. near-side against the MDB). The human model wore the set of seatbelts. The thoracic side airbag was activated at 7 ms from the initiation of the impact (hereafter, $t=7$ ms). The FE calculation was done until $t=80$ ms. Four conditions were chosen for the FE human models: adult and elderly FE models with activation of element elimination (i.e. "fracture" models) and adult and elderly FE models without activation of element elimination (i.e. "non-fracture" models), respectively. The respective kinematics and the time histories of the strain on the cortical bone of the ribs in the human FE models were measured.

III. RESULTS

Figures 5–9 indicate the results of the FE simulations. For convenience, the regions, organs, tissues and skins other than the thoracic skeleton are not displayed in these figures.

Figure 5 indicates the comparison of the kinematics of the rib cage of the "fracture" and "non-fracture" adult FE models at $t=30$ ms, 50 ms and 70 ms. Figure 6 indicates the close-up view of the upper ribs of those. Figures 7 and 8 indicate those of the elderly FE models in the same manner as Figures 5 and 6. Figure 9 indicates the strain contour chart on the cortical bone of the rib-cage in the "fracture" and "non-fracture" elderly FE models at $t=50$ ms. Figure 10 indicates the strain contour chart on the representative left ribs (R 7 and 8 respectively) in the same manner as Figure 9. Figures A1–A6 indicate the time histories of the acceleration and the absolute displacement of the specific points on the representative left ribs (R 7, 8 and 9 respectively) in the "fracture" and "non-fracture" elderly FE models. Figure A7 indicates the specific points on the representative left ribs where the acceleration and the absolute displacement were measured.

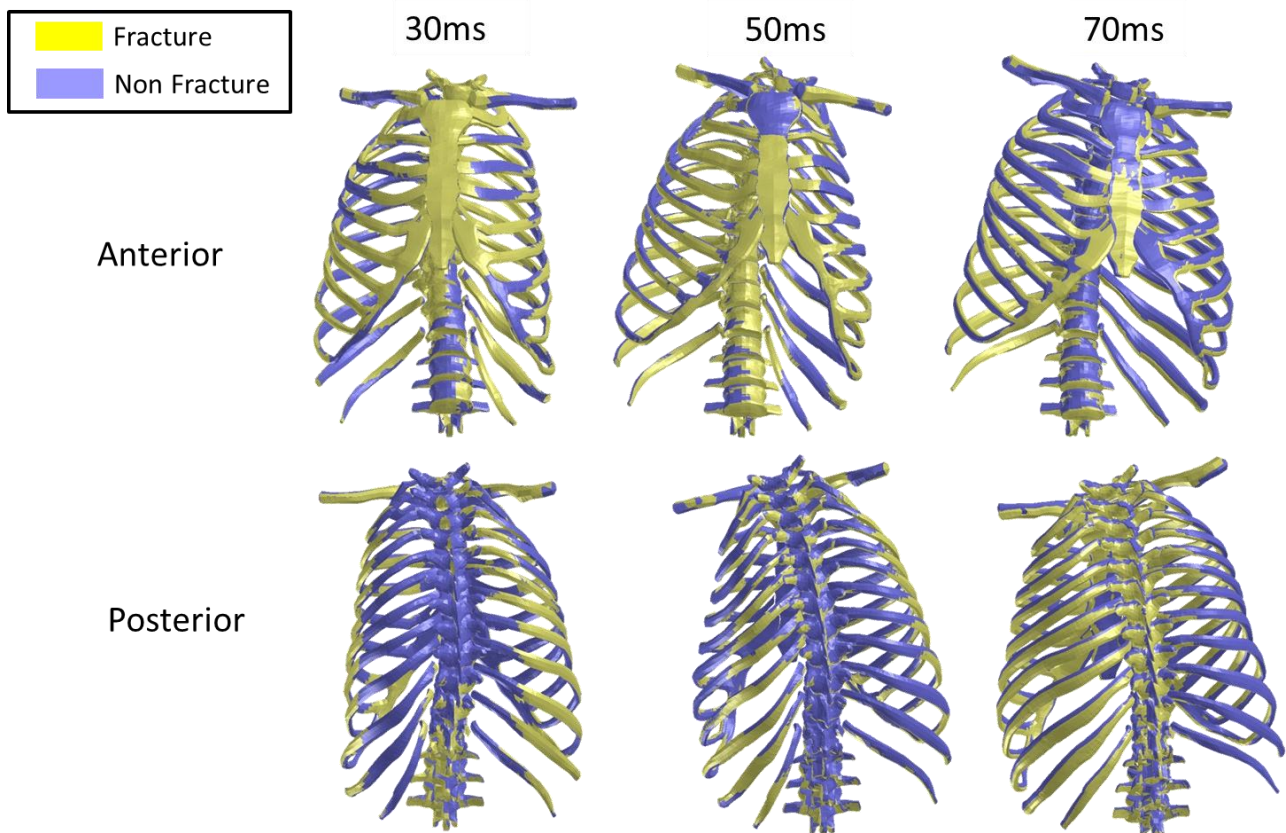


Fig. 5. Comparison of kinematics of "fracture" and "non-fracture" adult FE models at $t=30$ ms, 50 ms and 70ms; area in lighter colour indicates "fracture" model, but no significant difference between the two models.

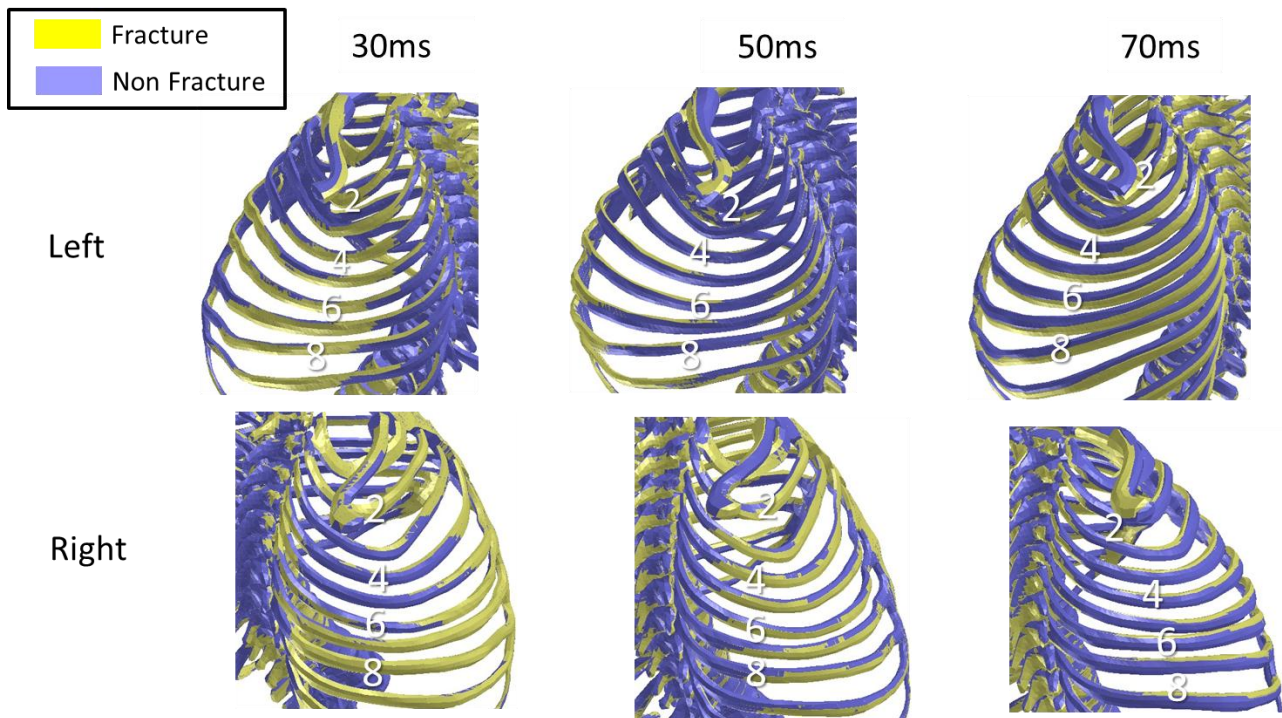


Fig. 6. Close-up of the upper ribs of “fracture” and “non-fracture” adult FE models.

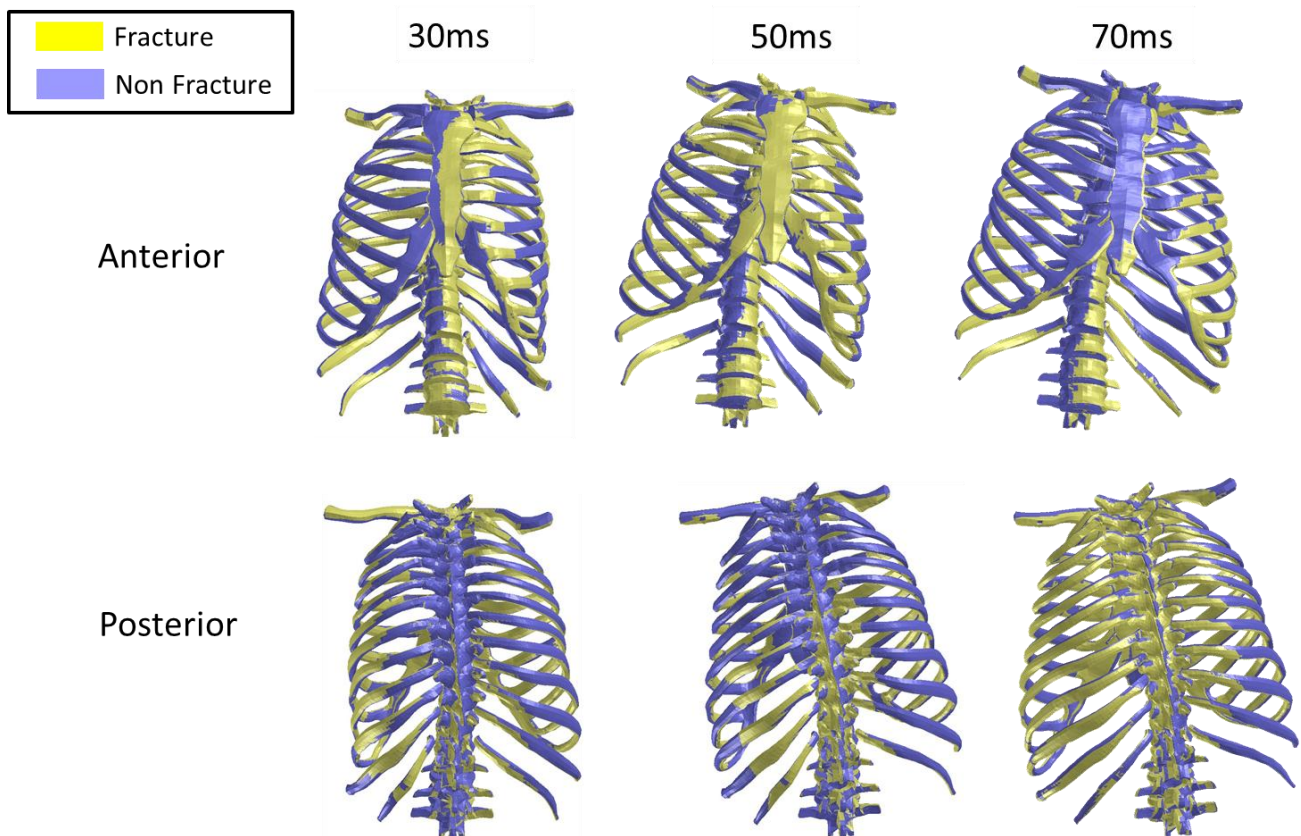


Fig. 7. Comparison of kinematics of “fracture” and “non-fracture” elderly FE models at $t=30$ ms, 50 ms and 70ms; “fracture” model in lighter colour indicates no significant difference from “non-fracture” model despite occurrence of fracture.

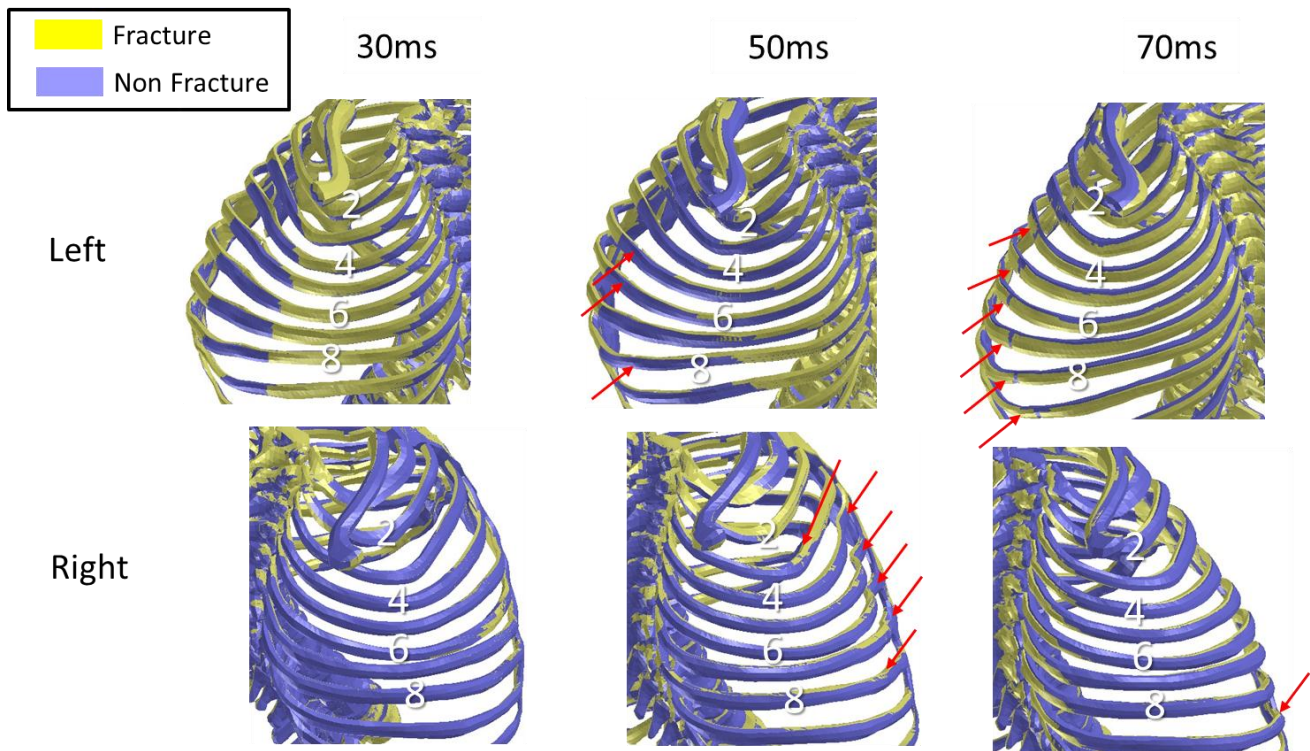


Fig. 8. Close-up of upper ribs of "fracture" and "non-fracture" elderly FE models; "fractures (red arrows indicate)" at R 3, 4, 5, 6, 7, 8 and 9 on right ribs, and R 4, 5, 6, 7, 8 and 9 on left ribs were observed.

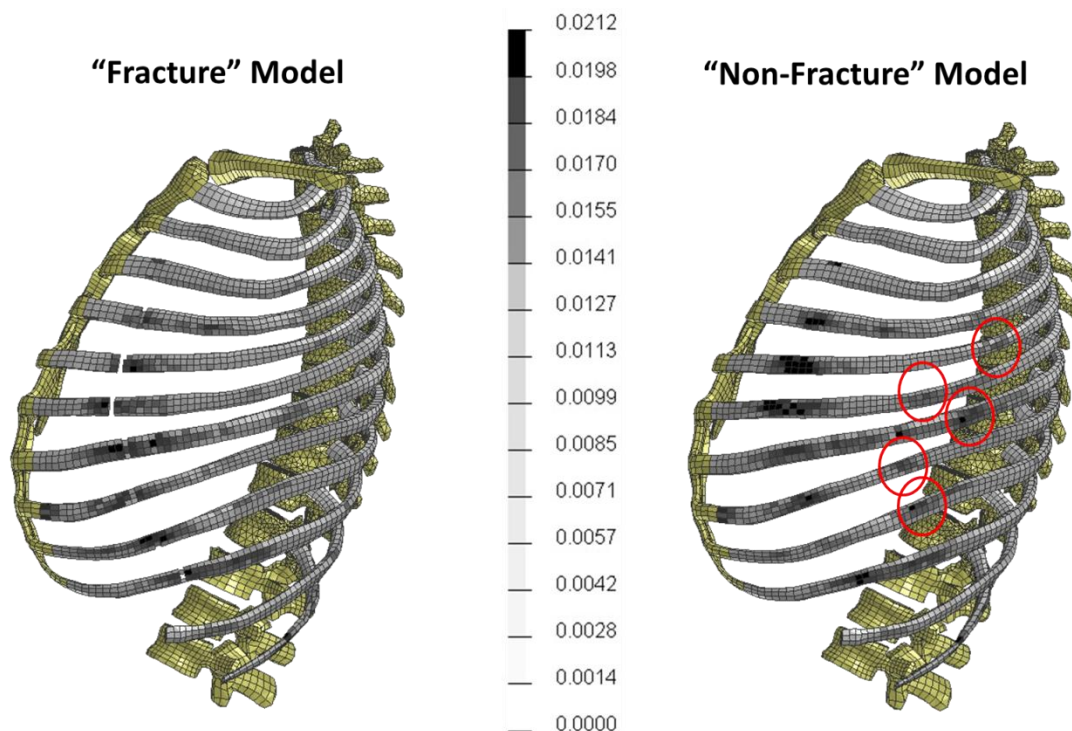


Fig. 9. Strain contour chart of left rib-cage of "fracture" and "non-fracture" elderly FE models at t=50 ms; "non-fracture" model indicates "hotspot" on posterior side of R 5, 6, 7, 8 and 9 (red circles), whereas "fracture" model indicates no "fracture", smaller strain at identical location.

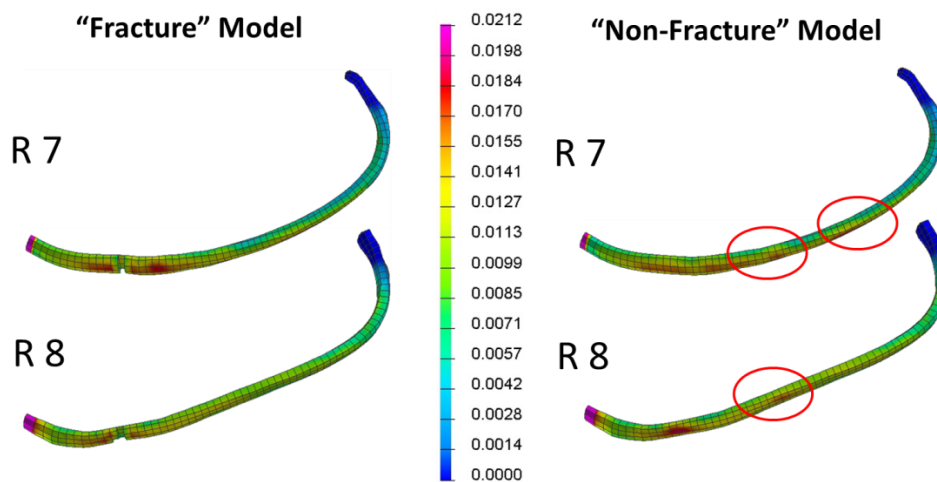


Fig. 10. Strain contour chart of left R 7 and R 8 ribs in "fracture" and "non-fracture" elderly FE models at $t=50$ ms; "non-fracture" model indicates "hotspot" on posterior side of the both ribs (red circles), whereas "fracture" model indicates no "fracture", smaller strain at identical location.

Taking a look at Figures 5 and 6, it was observed that there was little difference in the global kinematics of upper torso (ex. absolute amount of lateral excursion) between "fracture" and "non-fracture" adult models. It is deemed that this is mainly because the rib cage of adult models are less fragile, therefore the strain on the ribs did not reach the pre-determined threshold, i.e. the threshold of the fracture. Noticeably, taking a look at Figures 7 and 8, it was observed that there was little difference in the global kinematics between the elderly models as well as adult models. Close observation of Figure 8 indicates the multiple rib fractures (R 3, 4, 5, 6, 7, 8 and 9 on the right ribs, and R 4, 5, 6, 7, 8 and 9 on the left ribs) in the "fracture" model. The difference in the curvature of the ribs where fracture occurred in the "fracture" model was observed as well. However, no significant difference was observed in the curvature other than the region where fracture occurred.

Taking a look at Figures 9–10, the "non-fracture" model at $t=50$ ms indicates "hotspot" (location where strain reached pre-determined threshold) on the posterior side of the ribs, whereas the "fracture" model indicates no "fracture" and smaller strain at the identical location to the "non-fracture" model.

Taking a look at Figures A1–A6, it was found that the differences were very small (within 10 mm) in the absolute displacement despite the acceleration pulses on the ribs in the "fracture" model indicates steep spikes at $t=30$ ms (the time when the initial fracture was observed).

IV. DISCUSSION

In the first place, the authors discussed the comparison of the global kinematics of the elderly "fracture" and "non-fracture" human FE models. Between the two models, there was little difference in the global kinematics of the rib-cage, the general curvature of the ribs and the time histories of the local displacement and acceleration, despite the multiple rib fractures in the "fracture" model. The authors believe this is because the ribs are supported at both ends by the sternocostal joints and the costovertebral joints, and also constrained by the costal cartilage and intercostal muscle. Therefore, global stiffness of the rib cage deemed to be maintained after the rib fracture. This may support the contention that the "non-fracture" human model can represent the whole body kinematics of the occupant throughout the side impact, and also the feasibility of the methods for probability-based evaluation of the rib fracture by using "non-fracture" human FE models during side impacts. The authors determined the material property of the based on the previous research. However, due to lack of published information, the age-specific variation of the material properties of the intercostal muscles was not incorporated into the models used in the present study. Therefore, the effect of the variation of the stiffness of the intercostal muscles to the thoracic response including "fracture" should be further investigated.

Next, the authors discussed the distribution of the strain on the ribs. Taking a look at Figures 9–10, the "non-fracture" model at $t=50$ ms indicates a number of "hotspots" on the ribs, whereas the "fracture" model indicates no "fracture" and smaller strain at the identical locations to the "hotspots" on the "non-fracture" model. The authors believe this is because the stress on the rib was insulated by the primary "fracture" and the secondary fracture was inhibited. The observation suggests that the strain contour chart of the "non-fracture"

model may not represent the trend of the distribution of the “fracture” (i.e. may indicate false-positive in predicting the number of the fracture). The method proposed by Forman et al. was to quantify the risk of the rib fracture based on the number of the “hotspot” identified from the strain contour chart of the “non-fracture” FE human body model [16]. However, the observation may suggest the limitation in the prediction of the number of the fracture in their method. Whereas Mendoza-Vazquez et al. predicted the risk of the rib fracture using the measurements from the “non-fracture” FE human model during frontal crash and compared to the real-world data [15]. In their analysis, the prediction from the measurement of the shear stress and the first principal strain on the human model overestimated the risk of the rib fracture in higher impact velocities. The authors presume this is because the model did not reflect the post-fracture loading conditions in high-energy impacts. Therefore, the combination of the “fracture” and “non-fracture” models will be a relevant method to obtain perspective understanding of the kinematics of the thorax throughout the impact, although the authors hasn’t established the method to combine those two models in the present study. However, Figure 9 indicates little difference in the total (abbreviated) severity of the fracture (i.e. the number of the fractured ribs) between the “fracture” and “non-fracture” models regardless the location of the fracture. Therefore, this observation may justify the methods for probability-based evaluation using “non-fracture” human FE models. In the present study, the authors solely focused on the side-impacts with smaller constraints from the restraint systems. However, their justification for the probability-based evaluation method could obviously be feasible to the impacts with larger constraints from the restraint systems to occupants’ kinematics such as frontal or rear impacts.

Limitation

The elderly model used in the present study has not yet been validated against the response from experimental results due to the lack of literature regarding whole body experiment using PMHS of age 75 YO or older.

Due to lack of previous research, the range i.e. the age-specific or the individual variation of the material property of the intercostal muscles was not incorporated into the human body models used in the present study.

V. CONCLUSIONS

The objective of the present study is to examine the compatibility between “fracture” and “non-fracture” FE human body models in order to establish the method to quantify the probability of the rib fracture using FE human body models during side impacts. The authors conducted a series of simulated side-impact tests using human body models with and without representation of the rib fracture. The results indicate that the whole-body kinematics of the “non-fracture” human models were almost identical to those of “fracture” models. The locations of the ribs in the “non-fracture” models where the values of the strain reached the pre-determined threshold were well correlated with those where the element elimination was activated in the “fracture” models. The results suggested the feasibility of the methods for probability-based evaluation of the rib fracture using human FE models.

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VII. APPENDIX

Table A1. Representative material properties of FE human models.

Bones

		Yield Stress (MPa)		Yield Strain		Ultimate Stress (MPa)		Ultimate Strain	
		35 YO	75 YO	35 YO	75 YO	35 YO	75 YO	35 YO.	75 YO
Rib	Cortical	64.1	54.0	0.0071	0.0070	79.9	68.3	0.0246	0.0212
	Trabecular	3.31	1.25	0.0696	0.0061	68.3	1.25	1.00	1.00
Clavicle	Cortical	96.7	80.4	0.0102	0.0100	118.0	97.7	0.0430	0.0350
	Trabecular	3.31	1.25	0.0696	0.0061	68.3	1.25	1.00	1.00
Sternum	Cortical	64.1	54.0	0.0071	0.0070	79.9	68.3	0.0246	0.0212
	Trabecular	3.31	1.25	0.696	0.0766	3.31	1.25	0.0897	0.100
	Costal Cartilage	4.12	1.27	0.255	0.092	4.12	1.27	1.00	1.00
	Intercostal muscle	Bulk Modulus (identical to all models): 2.1 (MPa)							

Internal Organs

	Bulk Modulus (MPa)	Short Time Shear Modulus (MPa)	Long Time Shear Modulus (MPa)
35 YO	0.0264	2,400	900
75 YO	0.022	2,000	750

Table A2. Biofidelity scores of lateral (left) and vertical (right) displacement for body parts defined by Ruhle et al. (<1:Excellent, <2:Good, <3:Marginal, >3:Poor)

Head	0.40	0.86
T 1	0.37	0.66
T 6	0.23	0.94
T 11	1.00	1.70
L 3	1.18	1.79
Pelvis	0.94	0.71

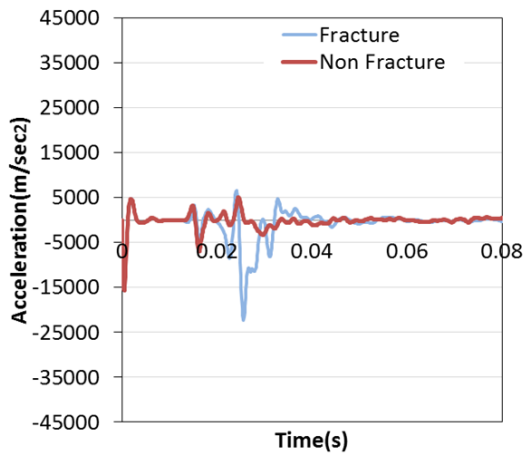


Fig. A1. Time histories of acceleration of specific point on left R 7 in "fracture" and "non-fracture" elderly FE models.

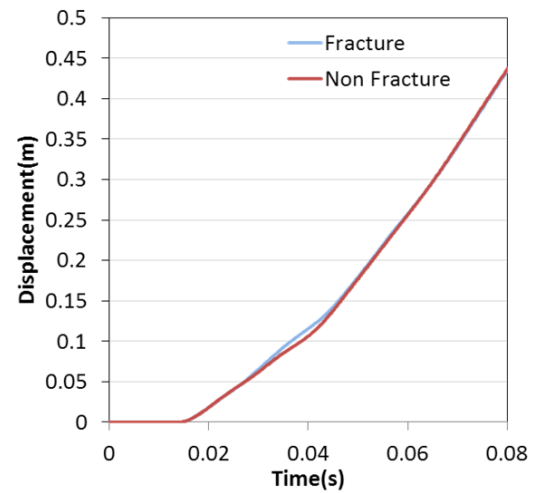


Fig. A2. Time histories of absolute displacement of specific point on left R 7 in "fracture" and "non-fracture" elderly FE models.

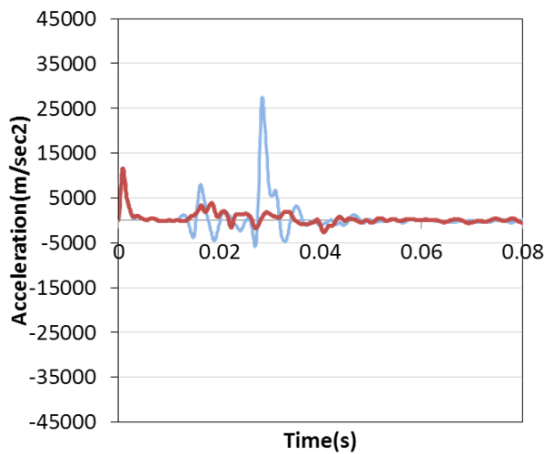


Fig. A3. Time histories of acceleration of specific point on left R 8 in "fracture" and "non-fracture" elderly FE models.

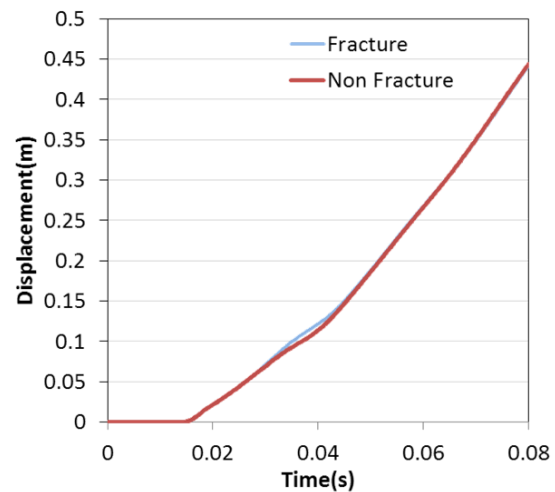


Fig. A4. Time histories of absolute displacement of specific point on left R 8 in "fracture" and "non-fracture" elderly FE models.

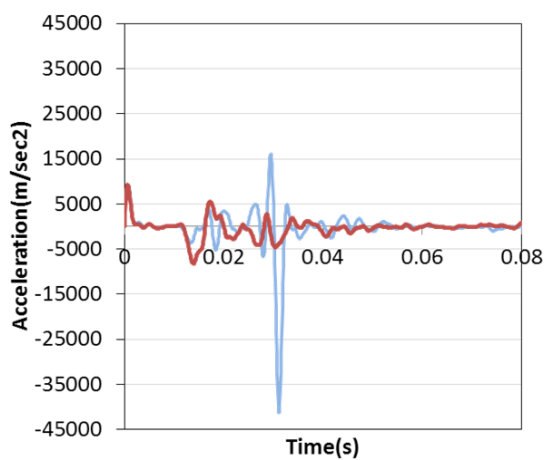


Fig. A5. Time histories of acceleration of specific point on left R 9 in "fracture" and "non-fracture" elderly FE models.

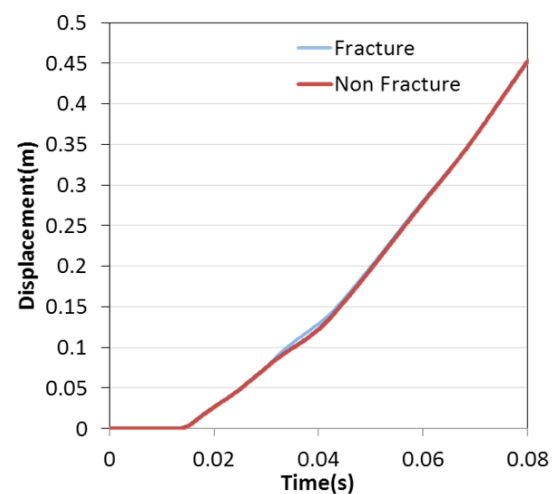


Fig. A6. Time histories of absolute displacement of specific point on left R 9 in "fracture" and "non-fracture" elderly FE models.

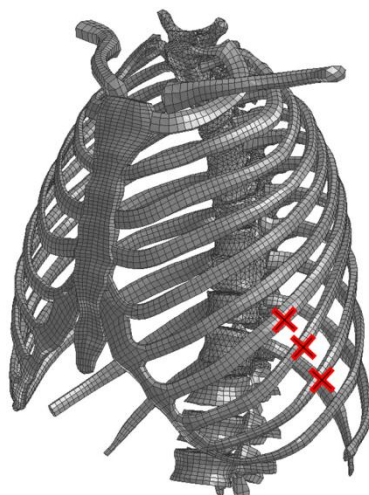


Fig. A7. Specific points on representative left ribs where acceleration and absolute displacement were measured.