Assessment of Abdominal and Skeletal loadings and Kinematics during Frontal Impacts through a Novel Tool for HBM Variants Generation Based on the Occupant's BMI

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Abstract A new methodology for creating HBM variants is presented and utilised to investigate the loading on the abdomen area and skeleton of three different HBM variants subjected to frontal sled tests. In all scenarios the same IP model was used. The study was able to verify the differences observed on the loading patterns between occupants of different BMI. The loading of the seatbelt on the upper abdomen was reduced and exhibited more time to build-up on the obese and M95 occupant models compared to the average. On the other hand the bladder and rectum exhibited higher stresses. Furthermore, the lower extremities of the obese and M95 occupant seem to have higher probability of injury as higher stresses on a greater area of the femur and tibia are observed with shorter time to impact onto the instrument panel. Different loading profiles were observed also on the spine which align with accident statistics.

Keywords Abdomen, crash, frontal impacts, obesity,

I. INTRODUCTION

It is well stated through accident statistic, that obesity is a key factor in the injury risk of motorised vehicle occupants as different injury patterns and morbidity rates are observed for obese occupants compared to non-obese. Reference [1] studied the risk of fatality and injury for front-seat occupants. They found that occupants with a body mass index (BMI) of 30–35 kg/m2 are 97% more likely to die and have a 17% higher risk of the Abbreviated Injury Scale (AIS) 3+ injuries compared to occupants with a BMI of 18–25 kg/m^2. Reference [2] reported an odds ratio for death of 1.013 for each kilogram increase in body weight and an equivalent of 1.008 for sustaining an injury with Injury Severity Score of at least 9 for each kilogram increase in body mass. Reference [3] found that occupants of Class I (30 kg/m^2 ≤ BMI < 35 kg/m^2) and Class III (BMI ≥ 40 kg/m^2) obesity were at the highest risk of injury in motorised vehicle crashes to a ratio of 1.24 with respect to the normal BMI occupants. Regarding abdominal injuries, research has identified a conflicting relationship with obesity. Some authors report a protective effect of the adipose tissue [4-5] others find increased abdominal injury and mortality [6-7] with increased loading towards the middle-lower abdominal area [8-9] while a third group [10-11] reports a U shaped relationship, suggesting that the protective effect may be overshadowed as BMI increases. Efforts to localise the injury patterns and frequencies are reported. Reference [8] analysed 112 cases of abdominal injuries by seat belt to clarify the relationship between obesity and abdominal injuries. Obese occupants on the front seat were found to suffer from middle-lower abdominal injuries (57%), whereas, non-obese mostly suffer from upper abdominal injuries (liver and spleen) (77%). They further reported that severity of abdominal injuries largely depended on the pelvic displacement in both obese and nonobese occupants. Reference [12] also reported different injury patterns depending on the BMI of the occupant. Obese belted occupants of Class III exhibit high injury probability on the lower extremities, spine and thorax while obese occupants of Class II exhibit high injury risk of upper and lower extremities. Reference [13] analysed cases of 4,183 occupants in 3,249 vehicles from the UK Co-operative Crash Injury Study (CCIS) and found that drivers of BMI>30 have increased probability of abdominal injury. Those of BMI>35 have three times higher compared to normal. Overall, the liver and spleen were found to be the most frequently injured organs for drivers. For front seat passengers, the liver was the most frequent followed by the jejunum-ileum and the spleen. Jejunum-ileum injury was rare for drivers. Reference [14] conducted a field survey data analysis to examine the effects of BMI on the risk of severity injury to hollow organs of the abdomen, i.e., small intestine, large intestine, or mesentery, among belted occupants involved in frontal crashes. Although non-statistically significant, a positive association between BMI and injury risk was observed, especially among obese individuals.

In order to gain deeper insights into the injury patterns and mechanisms between the different occupant body types several physical and virtual experimental scenarios have been investigated. Reference [15] performed frontal impact sled tests with obese and non-obese post-mortem human surrogates (PMHSs). Larger forward motion of the head, pelvis and lower extremities was observed for the obese subjects compared to non-obese. In addition, the torso of obese surrogates did not pitch forward. Since building a whole body Human Body Model (HBM) takes several years of development and research, currently available models are generally limited to young and mid-sized male and female occupants and do not account for body shape variations among the entire population. As a result, several efforts are made to rapidly create HBM variants using morphing techniques. A method for morphing the Global Human Body Models Consortium (GHBMC) model into any given age, sex, stature, and BMI has been presented by [16-17]. Reference [18] performed physical side impact tests using PMHSs and virtual tests with two variants of the Total Human Model for Safety (THUMS) created using morphing techniques. The first variant was parametric based on statistical geometry models while the second was subject-specific based on Computer Tomography (CT) data. It was found that the parametric variant tended to provide similar accuracy to the subject specific variant, while results from both variants correlated better than the baseline model to the PMHS data in terms of CORA value. Reference [8] created an obese variant of THUMS by scaling up the body surface of the original THUMS to a BMI of 34 based on the thickness of subcutaneous fat shown in abdominal CT image. Reference [19] developed a method to rapidly generate variations of THUMS with a wide range of human attributes (size, age, obesity level, etc.), taking into account skeleton variations, based on statistical geometry targets and morphing techniques. Reference [20] used GHBMC as a baseline model and applied mesh morphing methods to morph it into target geometries in order to examine the increased injury risks for older, obese, and/or female occupants in frontal crashes.

To the authors' knowledge though, limited work if any on HBM variant creation methods has taken into account the volume of the abdominal organs as a function of the occupant's BMI, On the other hand morphing based methods despite the simplicity they offer towards the quick generation of new variants, stretch the outer layer of the skin/flesh elements in order to create the additional fat layers which can influence element quality in the areas whose geometries are very different to the baseline model [18] and the simulation process consequently. However the creation of variants with high quality new or morphed elements in these areas is a very time consuming process that requires a good background in meshing and morphing techniques. In this study we have addressed these two topics by presenting a method for generating a parametric model using the GHBMC as a test case. The generated HBM variant was subjected to frontal sled tests restricted by a seatbelt and Injury Panel (IP) and compared with the reference Male (M) 50 model and M95 variant in order to investigate the overall body kinematics and generated loading patterns.

II. METHODS

In this study, a new methodology is proposed in order to rapidly develop HMB variants using BMI value as an input. The method tackles the issue of complex meshing and morphing processes for the creation of high quality subcutaneous fat elements and allows the resizing of the internal organs in accordance to each variant's BMI. Specifically, unlike other methods [8] [19] [20] where underlying elements of the model's skin are stretched to follow skin's morphing, a new subcutaneous fat layer has been created underneath the skin of the reference model for each new variant, where new elements are generated, in order to obtain high quality mesh inthese areas. For morphing and meshing ANSA v.23 [BETA CAE Systems SA, Greece] was used while for the simulations LS-DYNA R12 [LSTC, Livermore, CA, USA] and for the post-processing analysis of the results META v.23 [BETA CAE Systems SA, Greece]. As a baseline model, the midsize male occupant GHBMC (M50-O V6.0) model of 75.7 kg, BMI 25 was used, however the working method described below can be expanded to any HMB model. For the occupant's cabin, parts of the interior of a Toyota Yaris model, developed by George Washington University National Crash Analysis Center and used in support of several NHTSA programs were used [21]. To increase calculation speed, both the IP and the metallic parts of the seat were modelled as rigid bodies under the assumption that these parts do not go under significant deformation. We deem that the assumption is valid as the impacting velocity is kept low since the focus of this study is placed on the occupant's kinematics and the deformation of the IP would add further complexity to the study. GHBMC has been previously validated against a number of frontal and lateral rigid impactor and sled tests [22-23] although its biofidelity is under constant assessment [24]

The variant generation process can be split in four phases. First for a given age, sex, stature and BMI, the external target geometry for the body shape is imported. There is not specific limitation to the body type that will be imported . For this study a 3D scan of a real person was used but statistical geometry models developed previously [25-27] can also be imported. Second, the imported geometry was overlaid on the reference M50 v.6.0 GHBMC model. Third, the external surface of the abdominal organs was morphed to reach the target volume of BMI 44 and 135 kg (Class III [3]), and a copy of the external surface (skin) of the reference model was morphed to fit the target geometry using morphing tools. For the volume of the liver, kidneys, pancreas, gallbladder and stomach gastric capacity per BMI, regression curves and measurements per BMI reported by [28-29] were used. The length of the stomach's lesser curvature that forms the upper right or medial border of the stomach which travels between the cardiac and pyloricorifices [30] was considered to be constant [31][32] while for the length of the stomach's greater curvature the regression curve presented by [31] was used. Due to limitations imposed by the skeleton, morphing of the internal organs is implemented up to BMI 33.3. For higher BMI values, morphing of the model's skeleton would be required in order to further expand the solid and hollow organs, something that is not currently supported by the model. As a result, the maximum volume that the internal organs can reach, corresponds to BMI 33.3. This limitation does not apply though for the external surface of the body where no space limitations were imposed, thus further expansion of subcutaneous fat is possible for the creation of variants of even higher BMI. For the morphing process, landmark-based Radial Basis Function (RBF) interpolation algorithm [33-34] was utilised, previously used by [19] [35] also for building parametric human models. Landmarks were manually selected onto locations of the baseline model, corresponding to locations on the predicted statistical geometry and provided to the RBF as inputs. All created landmarks and information used as input for the RBF function can be stored in a separate metadata file and re-used in other test cases with the same HBM model or adapted for different models.

Finally the space between the morphed surface and the outer external surface of the model was filled with elements that will constitute the new layer of subcutaneous fat. As the external surface of the HBM is very complex it was split into smaller, simpler ones. Closed volumes filled with hexa or tetras were created between these surfaces and the corresponding outer external surfaces. The volumes were subsequently merged into one that constituted the additional fat layer. The volumes on top of all areas on the HBM's skin that were originally meshed with quads, were meshed with hexas and above the intermediate skin areas that were meshed with trias tetras were created. This way, the new mesh is compatible with the pre-existing and continuity between the element types of the different areas is preserved.

The approach leaves intact the elements of the reference model while allowing the quality control of the new elements as defined prior to the model generation process through mesh quality criteria.

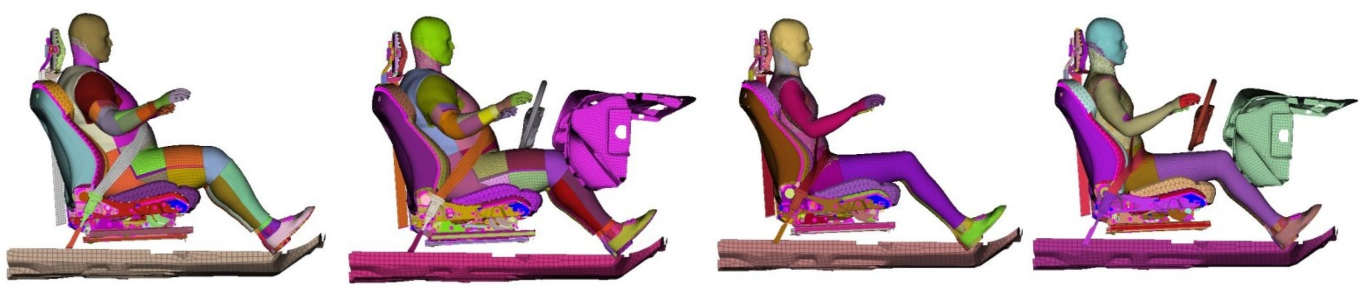

Fig. 1. Indicative positioning configurations of HMBs on seat (Left: Obese, Right: M50).

The model was primarily tested to see if the morphed organs can reach the required volume for a range of BMIs. Numerous variants were created for BMIs ranging from 25.6 to 33.3 and their volume was compared to the target.

As the data on exact abdominal loadings are very scarce in the literature with studies mostly focusing on overall injury patterns, the aim of this study is to confirm these patterns as a means of model validation. M50 and M95 are state of the art HBMs widely used in a series of studies [16-20]. In order to assess the results acquired by the generated HBM variant, the kinematics and loads are analysed based on accident statistics and previous FE studies where obese occupants are studied during crashes. A comparative study between reference (M50) model and the two other variants (M95 and Generated Variant of MBI 44) is thus subsequently conducted. Different crash scenarios were subsequently investigated where the reference model (GHBMC M50 v6.0) and a generated variant of high BMI (GHBMC BMI 44) were subjected to frontal impacts of 6 m/s in the presence of a car's IP and without it. As the M95 variant can be associated with heavier occupants since there are no obese variants commercially available, frontal impacts with the M95 (102 kg, BMI 28) in the presence of IP were also conducted in order to investigate the extent to which this heavier variant is a good surrogate for obese occupant models. For the HBM's position on the seat, the HBM Articulation tool was used while for the creation of the belt, the Belt tool of ANSA was used. The positioning of the belt was chosen to be such that the belt angle remained the same for all cases as it is considered to be one of the main parameters of influence in the occupant's kinematics and the manifestation of submarining [36]. The seatbelt was modelled using a simple elastic material MAT_B01_SEATBELT [22] of LS-DYNA database. For the seatbelt a pretensioner was used with a maximum force of 6kN as presented in [37]. Finally the kinematics of the models and the loading profiles on the skeleton and internal organs were compared amongst the different scenarios in order to define the produced kinematics and loading patterns. For the assessment of the upper abdominal injuries, Strain Energy Density (SED) was measured for the liver and spleen, the most commonly injured solid organs [8] [13] and Von Misses Stress for the hollow intestines (colon, small intestine, bladder).

III. RESULTS

The current variant generation approach demonstrates the ability to develop HBM variants based on the occupant's BMI within less than an hour without critical impacts on mesh in terms of element length and jacobian value. Regarding Jacobian as a measure of elements' quality [19] [38], all solid elements created for the external fat tissue had a value above 0.5 and only 0.2% of shell elements had a value of less than 0.7. No morphed elements of the abdominal organs were observed outside these threshold values. External body surfaces were morphed according to the specified BMI, until the morphed surface coincides with the target. In the area of the belly where the maximum distance from the reference model is located (128 mm), 6 new rows of elements were created with length of 21.3 mm towards the target surface. The length of the reference model elements in this area is 14.9 mm. If solely morphing techniques were applied on GHBMC to reach the target surface, the total displacement of 128 mm would have to be distributed to the three pre-existing rows of elements which would equal to an increase of 284% of their element length towards this direction as opposed to 42% which is with the current approach. The increase could be even less if additional rows were created. Similarly, the required target volume for the abdominal organs was achieved with deviations less than 8% difference between target volume and finally achieved for all organs, and BMIs apart from pancreas that reached 12%, but only for BMI values above 31 (Fig. 2). Compared to time-consuming manual methods, the proposed method significantly reduced the generation time of the variant.

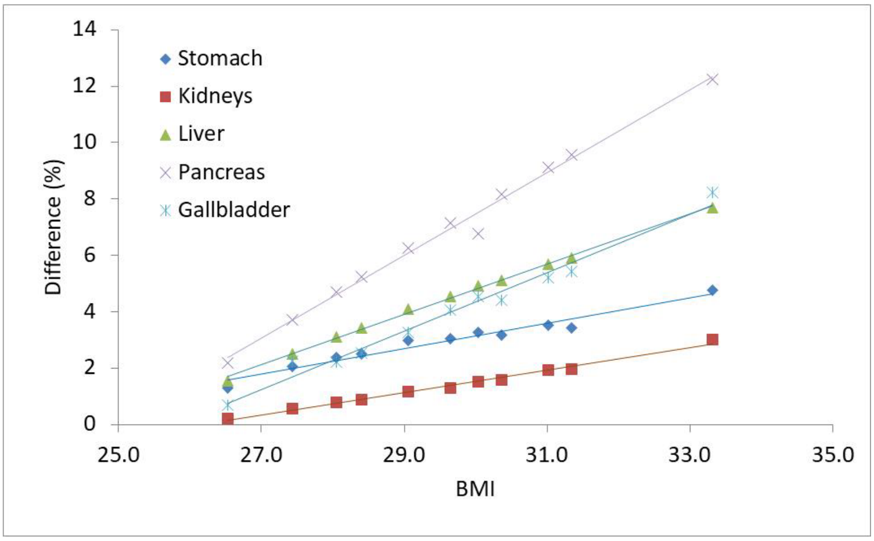

Fig. 2. Difference between organs' target volume and final volume after morphing.

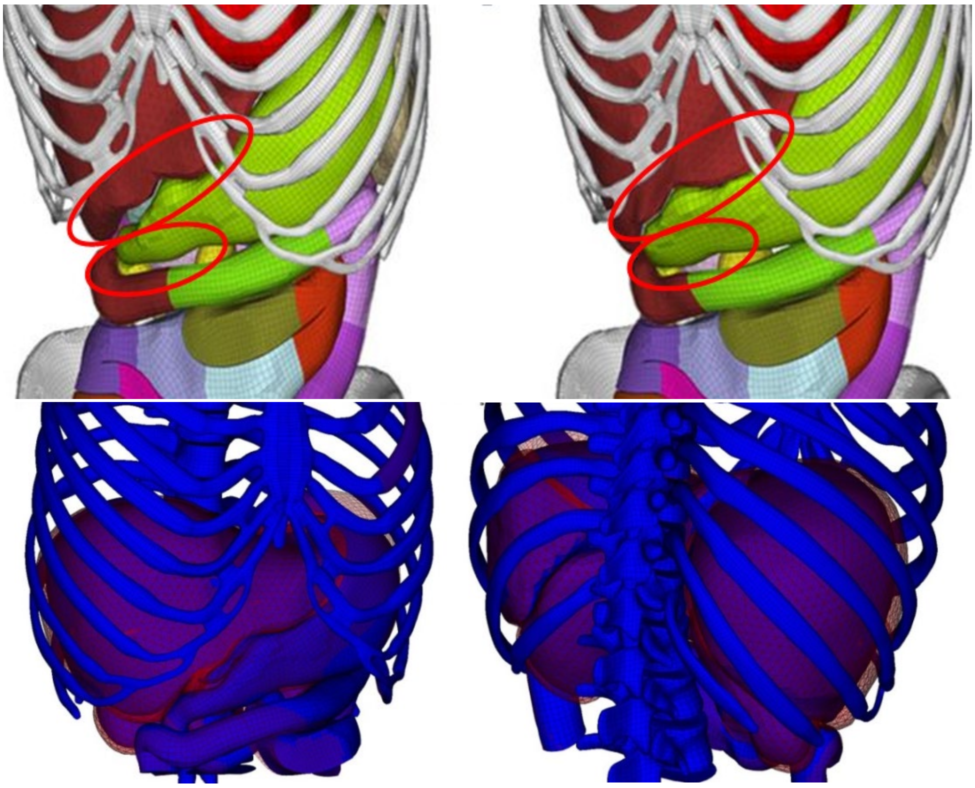

Fig. 3. Upper row: Resized to BMI organs inside the rib cage showing the increase in liver's and stomach's volume. Left: Reference, Right: Obese, Lower row: Overlay of organs in the ribcage, Blue for Reference and Red for Obese, Left: Frontal view, Right: Back view.

The generation of a new variant did not take more than 25 minutes

Overall Kinematics and Belt Loading

The overall kinematics reveal a difference in the motion profile in the presence of and without IP. In the scenario without IP where there is enough space for the occupant to slide on the seat's surface, obese occupants exhibited greater forward motion of the head and the pelvis compared to reference M50 while the torso seemed less prompt to rotate forward compared to non-obese subjects [15].

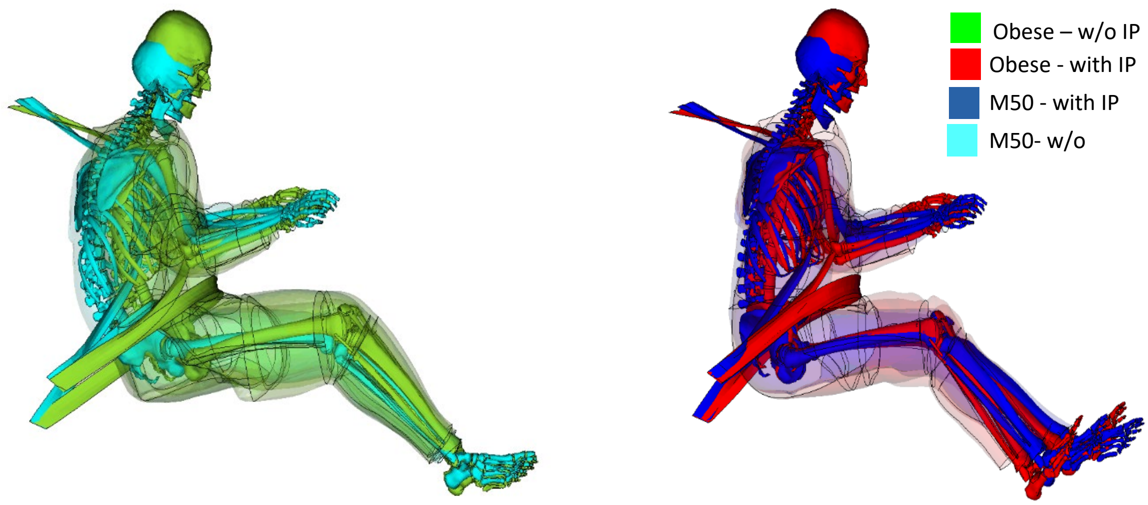

Fig. 4. Skeleton kinematics comparison of obese and reference occupant model, (Left: without IP Right: with IP).

In the presence of IP, the legs would soon hit on its surface thus no significant motion of the pelvis was allowed for both BMIs. This resulted in increased stresses across the lower extremities' of the obese occupant as the impact on the IP was more severe due to the increased trend of the legs moving forward. This seems to be in line with previous FE [8] [35] and statistical studies [12] [15] that show increased injury risk on the lower extremities of obese occupants when compared to non-obese. Similarly, M95 occupant exhibited high stress increase due to the limited space available to the IP as a result of its longest limbs (Fig. 5).

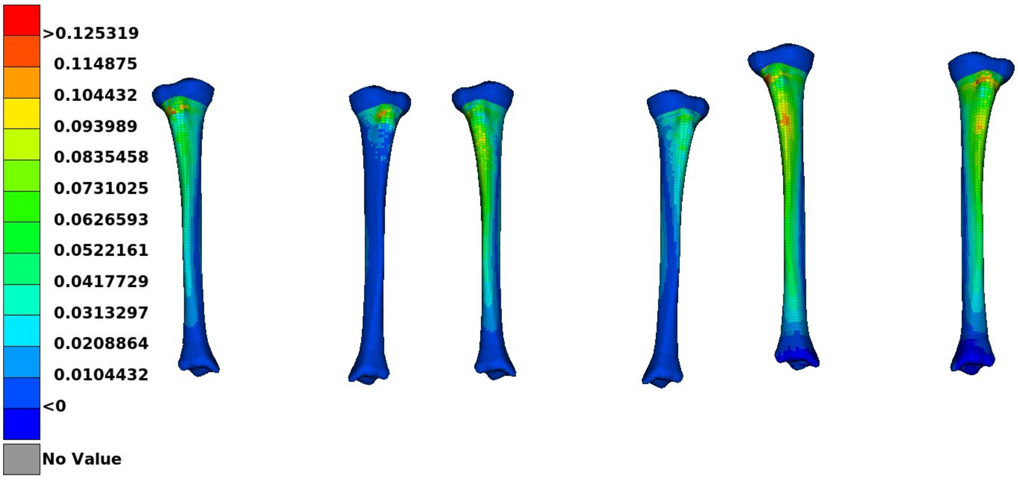

Fig. 5. Distributions of Von Mises (VM) Stress (GPa) on Tibia per variant (Left: M50 Middle: Obese, Right: M95).

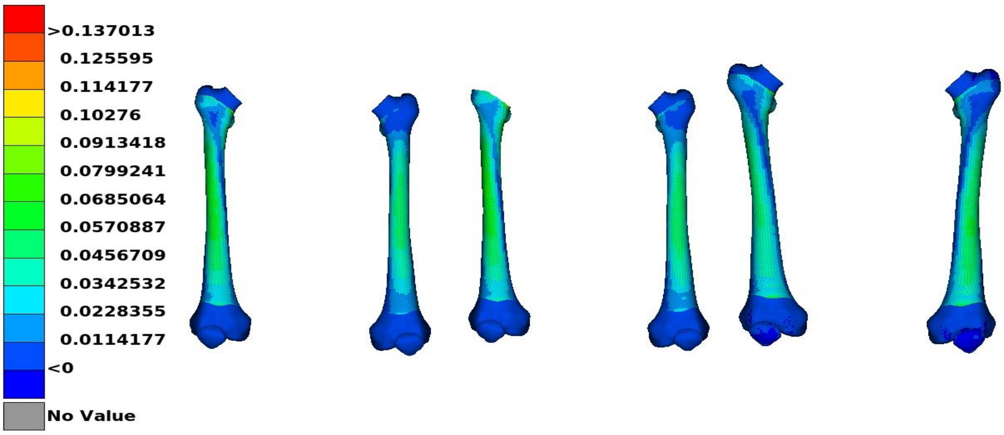

Fig. 6. Distributions of Von Mises (VM) Stress (GPa) on Femur per variant (Left: M50 Middle: Obese, Right: M95).

Higher stresses were also observed on the spine of the obese occupant compared to M50 and M95, specifically between the thoracic T12 and lumbar L1 vertebrae [12] [39]. Stresses on the clavicle and around the sternum are higher for M50 which potentially explains fractures observed in the same regions on belted occupants of BMI<30 compared to obese [40]. Higher stresses were also observed on the clavicle of the M95. The loading profile of the spine for M95 seems closer to the reference model without a significant stress increase between the corresponding vertebrae as in the obese variable.

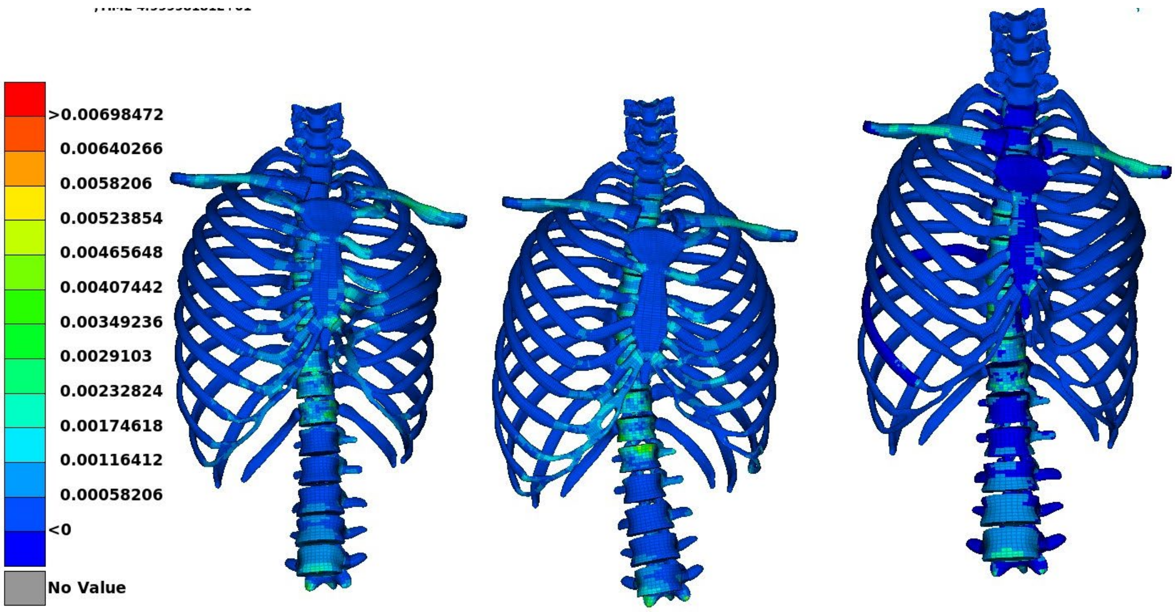

Fig. 7. Von Misses Stresses (GPa) on spine and rib cage (Left: M50 Middle: Obese Right: M95).

In both IP configurations, obese drivers exhibited greater upper shoulder belt stresses than non-obese[15] and M95. This could be attributed to the higher moment of inertia that is linked to the extra mass of the obese occupant torso and its higher tendency to move forwards [1] [39][41-42].

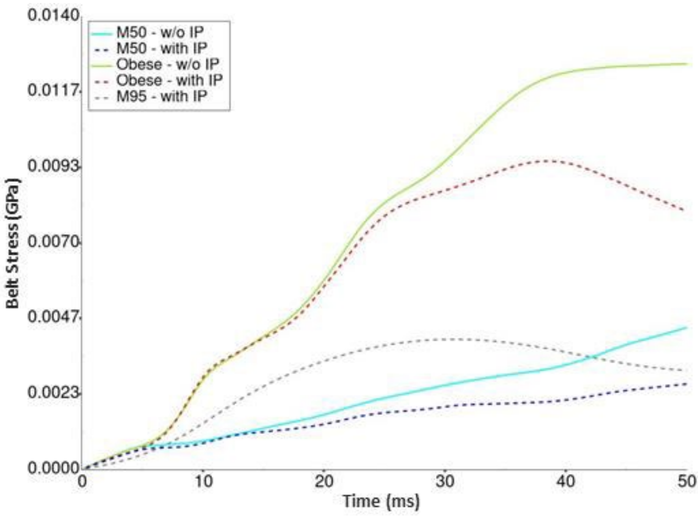

Fig. 8. Belt Stress (GPa).

Upper Abdominal Loading

SED on the liver is substantially higher compared to the corresponding values for the spleen for all scenarios which can explain the higher frequency of injuries observed on the liver in [13] for occupants in the driver's seat. The higher SED of the liver can be attributed to the higher compression of the left side of the driver's ribcage by the belt as it passes over this area. This mechanism could also explain the higher injury risk of the spleen for front seat passengers on the right side of the vehicle [8] where the belt is shifted to be attached on the right side of the vehicle's frame.

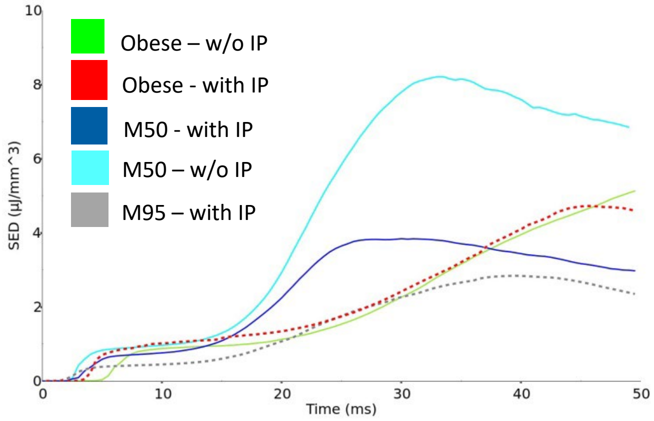

Fig. 9. Liver's SED for the four impact scenarios.

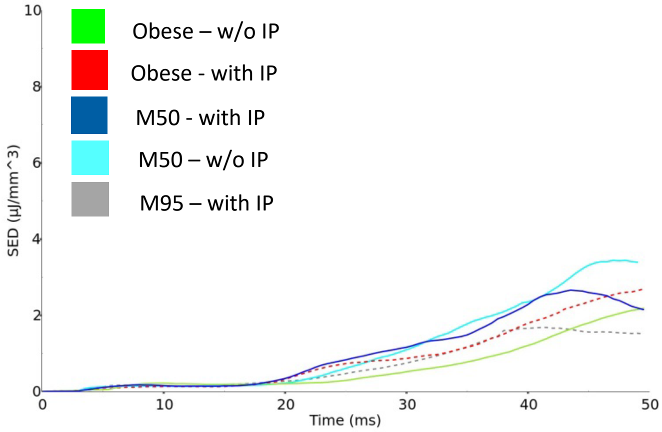

Fig. 10. Spleen's SED for the four impact scenarios.

In the presence of the IP, obese occupants showed higher values of SED both for the liver and the spleen compared to non-obese [13]. In absence of the IP, the situation reversed with non-obese occupants exhibiting higher values of SED for the same organs. As the IP is not present to constrain the motion of the obese driver and prevent his forward motion along the seat, the difference in kinematic tendencies was much more pronounced. In absence of the IP, the deflection of the non-obese driver chest was much higher in the area of the liver and spleen, specifically on the entire right and the back of the left thoracic area where the two organs are located, respectively (Fig. 11 Lower). This potentially results from the fact that the response of the non-obese driver to the belt's loading has to be much more rapid, as there is no subcutaneous fat to cushion the loading on the ribcage nor allow some additional movement of the pelvis, thus increased loading and deformation of the ribcage are applied. This could also be evident by the fact that SED curves for obese occupants seem to have longer build-up time towards their maximum potentially due to the cushioning effect. [42] [5]

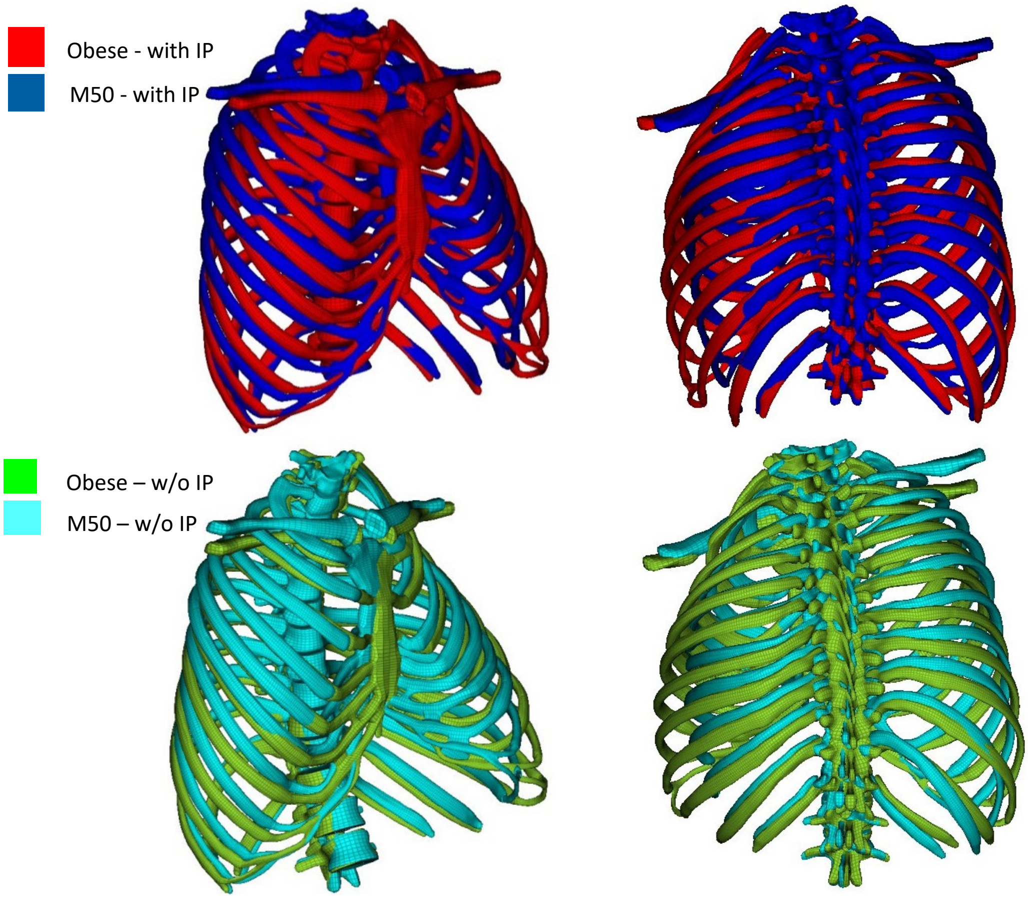

Fig. 11. Chest maximum deflection of obese and M50 drivers in presence of IP (Upper) and in its absence (Lower).

Finally when comparing the loading on each variant in presence and absence of IP, it was observed that both for obese and non-obese occupants, the liver's SED was higher in the absence of IP. This can be attributed to the fact that the severity of abdominal injuries largely depends on pelvic displacement as in the absence of IP the pelvic displacement was higher [8].

In all cases, the M95 exhibited lower SED on the liver and spleen compared to the other two variants despite the fact that the M95 is heavier than the M50, and potentially a behaviour similar to the obese variant could be expected. As the M95 has the maximum thoracic volume, followed by the obese variant and then the M50, it seems that the volume of the inner-ribcage cavity plays a dominant role allowing greater compression of the variant's ribs with lower loading of the underlying organs, thus reduced SED. This could potentially be a reason for the increased build-up time of the M95 SED curves, similar to the obese variant.

Middle-Lower Abdominal Loading

In order to investigate if the differences on the loading profile of the Middle-Lower region between occupants of different BMIs can be quantified and depicted by the current study, VM stresses were measured on the solid (pancreas, gallbladder) and hollow (duodenum, small intestine, colon, rectum and bladder) organs of the region. The quantification is mainly performed for reasons of comparison however further experimental research is needed to confirm the magnitude of the results. Loading on the lower area of the abdomen was more prominent for obese occupants in comparison to non-obese [8-9], with the rectum (17-30 ms) and bladder (46-50 ms) exhibiting the highest loading when compared to the rest of the organs in presence of IP, and the rectum undertaking stresses almost equal to the area's maximum in the absence of IP (33-37 ms) (Fig. 12). This was not the case for non-obese and M95 occupants where stresses on the rectum and bladder were significantly lower than the stresses in the rest of the hollow organs of the middle-lower abdominal region.

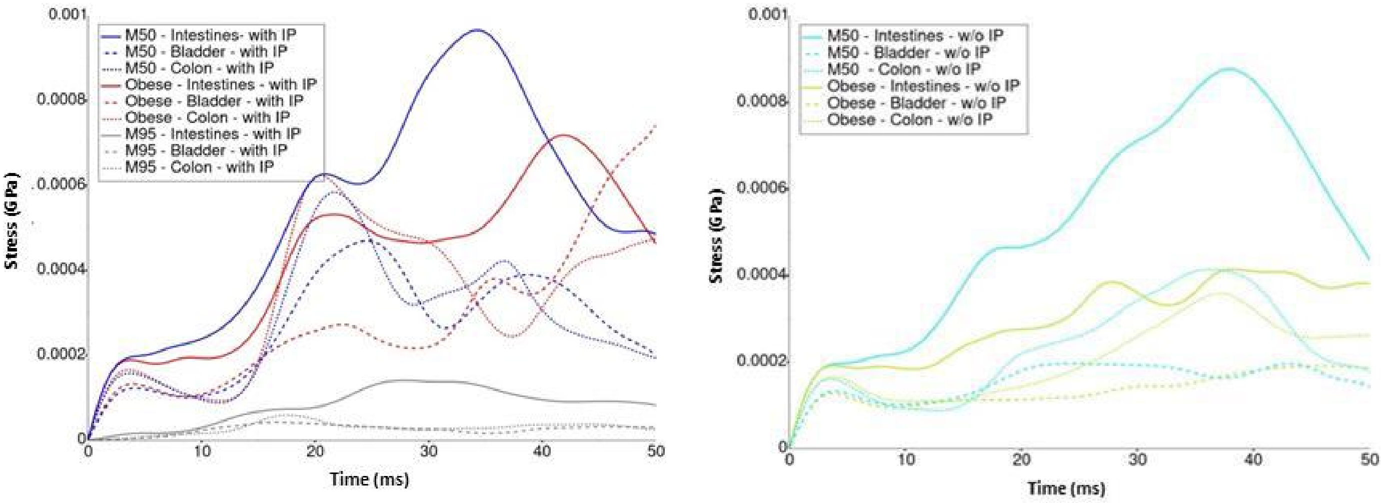

Fig. 12. Von Misses stresses in the middle-lower Abdominal area (Left: with IP Right: w/o IP).

Stresses on the pancreas seem to follow a pattern similar to the liver's SED. In the presence of the IP, stresses seem to have a higher build up time for obese occupants as they are lower during most of the impact, but eventually reaching higher values compared to non-obese. In the absence of the IP, the situation was reversed with non-obese occupants exhibiting higher values of VM stresses. This could be attributed to the mechanism already described in the previous paragraph. No significant differences were observed between the different scenarios for the loading to the gallbladder.

IV. Discussion

In this study, a novel method of creating HBM variants is presented. The main novelties of the method is that it takes into account the dependence of the abdominal organs morphology on the occupant's BMI and mesh quality criteria for the subcutaneous fat, through the creation of new elements. In order to verify its potential, a series of variants is created to assess the quality of the generated mesh, the accuracy of the finally achieved volume per abdominal organ with respect to the target volume and the overall functionality and behaviour of the generated HBM during a frontal crash in LS-DYNA. The generated models showed good element quality in terms of element length and jacobian value, deviation below 10% between the target and finally achieved volume while the kinematics of the model seems to be aligned with previous statistical and FE studies.

Specifically, various frontal impact scenarios were reproduced to verify the kinematics of the model and gain insights into the loadings exerted on the occupant by the belt and the IP. As the data on exact abdominal loadings are very scarce in the literature with studies mostly focused on overall injury patterns, the study tries to confirm these patterns as a means of model validation. The quantitative results generated serve the purpose of better comparison between the different variants in the study although experimental results are needed to give deeper insights on the investigated loadings. However that would require the creation of a testing rigs and equipment that goes beyond the scope of this study.

The injury patterns found in literature are indeed aligned with the findings of this study. Specifically, in the absence of IP, greater forward motion was observed in the case of the obese occupant [1][39][41-42] as greater occupant mass has been linked to increased kinetic energy, thus inducing forward hip/pelvis movement before adequate safety belt restraint. This contributes to the observation of increased lower extremity stresses and injuries in frontal crashes, resulting from increased hip excursion and a higher knee impact against the lower instrument panel [4][12][39][42][43]. Forward pitch of the torso was decreased for both cases for the obese occupant [40] and increased stresses on the spine were observed compared to non-obese and M95 [12] [39]. A different loading profile of the internal organs was observed with an overall translation of the stresses on the hollow organs towards the lower abdomen area (rectum, bladder) for the obese occupants. Solid organs showed also different loading profiles depending on the presence of IP. Specifically, obese occupants experienced increased loadings on the liver and spleen in the presence of IP, which is closer to the driving space of a vehicle’s cabin than that of non-obese in the absence of IP, which is closer to sled testing procedures. The M95 exhibited relatively lower loadings compared to the other two variants.

These observations give an additional value to the need of creating models for obese occupants, as currently available HBMs of higher mass seem not to be sufficient surrogates for occupants of higher BMI. They also put some additional focus on the testing procedures of frontal impacts, as they pin point the differences in occupant kinematics and loadings within the restrained environment of vehicle cabins, compared to the more simplified created by testing set-ups during the frontal sled test certification procedure.

Even though the method is applied on GHBMC M50 as a case study, it could further be extended on other percentiles or HBMs in order to create additional subcutaneous fat layers and increase the volume of the abdominal organs. To that purpose, HBMs of low and very low BMI that go beyond the range of BMIs investigated in this study could be used, as long as measurements on the organs’ volume are available.

A limitation of the method is that no morphing of the skeleton is applied. As a result the organs cannot be morphed to BMIs higher than 33.3 as their volume would increase to the point that they would contact the bones of the ribcage. However there is no such issue for the external surfaces of the body thus further expansion of subcutaneous fat is possible for the creation of variants of even higher BMI while the volume of the abdominal organs maintains its maximum value (which corresponds to BMI 33.3).

It should be mentioned, that in this study the presence of lumped internal organs is not taken into account as it is based on the 6th version of GHBMC where such conditions are not present. However it could be extended to such cases if a library of volume measurements and MRI scans per BMI is created as there is no limit to the geometry of the surface that is used as target during morphing.

Finally it should be noted that this study is focused on the mesh generation and morphing method for rapid development of HBMs, without performing direct validations of the morphed models against cadaver test results. As the mesh morphing method is proven to be robust to generate a large number of HBMs, the next step of our study is to validate and apply these models in crash conditions.

V. CONCLUSIONS

A new methodology for creating HBM variants is presented and utilised to investigate the loading on the abdominal area and skeleton of three different HBM variants subjected to frontal sled tests. In all scenarios the same IP model was used. The study was able to verify the different loading conditions observed between occupants of different BMIs on the respective areas. Loading of the upper abdomen exhibited more time to build-up on the obese and M95 occupant model compared to M50. On the other hand the bladder and rectum on the lower abdomen exhibited higher stresses. Furthermore the lower extremities of the obese and M95 occupant seem to have higher probability of injury as higher stresses on a greater area of the femur and tibia are observed with shorter time to impact onto the dashboard. Different loading profiles were observed also on the spinal cord which aligns with accident statistics.

This study and the process presented could aid further experimental research and measurements on the loadings of the abdominal region and organs. The process presented in combination with such measurements could lead to further enhancements of HBMs.

VI. ACKNOWLEDGEMENT

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