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Development of a New Finite Element Model of the Abdomen for Impact

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ABSTRACT

The aim of the current study was to develop a finite element model of the abdomen for impact as part of the GHBMC coordinated effort. The model development and its validation were performed in multiple steps, from the selection of numerical formulations using isolated organ models, to the meshing of the full region and its validation against experimental data. The model response was found to be stable in 10 simulation setups, and close to literature data for most. However, the results highlighted the need to take into account geometrical differences between the model and the test subjects from the literature in order to be able to perform relevant comparisons. A sensitivity to the model mass was also observed for some of the test setups.

INTRODUCTION

The objective of the current study was the development of a new model of the abdomen as part of the Global Human Body Modeling Consortium (GHBMC) coordinated effort. The modeling of the abdomen for impact is a mechanically complex problem. Biological soft tissues are typically non-linear materials and the abdominal anatomy includes numerous organs and structures with irregular geometry and vasculature, and complex relationships. They can also be subjected to very high strains. Most importantly, injury mechanisms and internal response during impact are still not very well understood, in part due to limitations of experimental observation methods. As a consequence, past model have often suffered from numerical stability issues.

For such a complex problem, it was decided to define pragmatic performance targets for the future model:

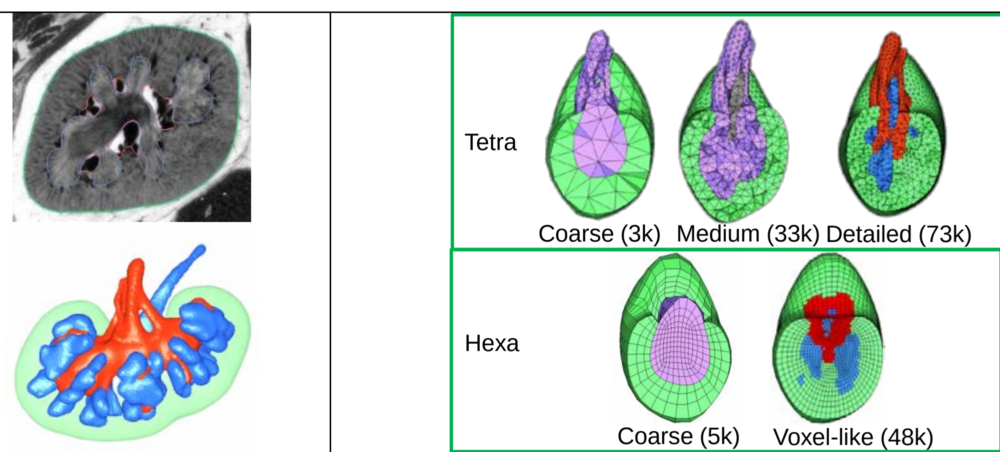
1) *Stability*: the model that is defined should be stable even when subjected to strenuous impacts in order to ensure that the complete model can run
2) *Contribution to the full human model*: the abdomen contribution to the full human model should be appropriate in terms of kinematics, mass etc.
3) *Load penetration response*: the model should have an acceptable response (compared with PMHS testing) for impacts in multiple directions and locations.
4) *Injury prediction*: in the absence of internal validation data at the beginning of the work program, it was decided to focus on solid organs and compare the model response with previous test results.

In order to attempt to achieve these targets – especially with regards to the stability, a bottom up approach from organ to whole region was selected for the development. The current manuscript focuses on the first and third points.

METHODS

Organ modeling: detail and numerical options

Solid organ modeling. In order to select stable numerical options for the organ models, a preliminary organ model based on the geometry from the Visible Human Dataset (Visible Human Project) was developed. It is a kidney model that was segmented manually from the high definition Visible Human Dataset. The model includes cortex, pyramids, major vessels and fat. It was then meshed into five models after simplifying its geometry by smoothing algorithms (Figure 1). The model included three tetrahedral models (coarse, medium and detailed) and two hexahedral models (coarse and voxel-like). For the model called voxel-like, a high density hexahedral mesh was first generated and then parts were defined based on the location of the elements. The models were implemented in the three codes defined by the GHBMC: Ls-Dyna3D (LSTC), which was the baseline code, Pamcrash (ESI) and Radioss (Altair Engineering). Linear viscoelastic and/or Ogden based hyperelastic viscous properties were selected for the different parts. Material parameters were adapted based on an initial selection from Schmitt et al. (2005). The models were then subjected to an impact using a free flying pendulum as described in Snedeker et al. (2005). The global responses of the models in terms of force vs. displacement, and the Von Mises stress distributions in small spheres defined in the cortex and the medulla were compared for various models, codes, and element formulations. The simulation setup and sphere locations are illustrated in Figure 2.



a) Segmentation and polygonal geometry

b) The five models with different levels of detail and mesh densities

Figure 1: Geometry based on the Visible Human Dataset and meshes of the kidney

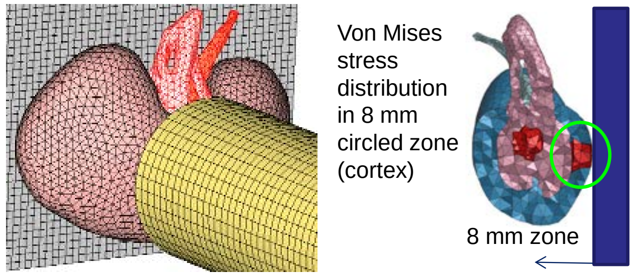


Figure 2: Kidney impact simulation. Left: impact test setup. Right: zones considered for the Von Mises stress analysis.

*Hollow organ modeling***.** A similar approach (which is not described in detail in the current manuscript) was used for the selection of numerical options for modeling of hollow organs: a stomach model was first defined based on the Visible Human Dataset. Then, modeling approaches tested included solid elements or monitored volumes for the organ contents, and solid or shell elements for the membrane. In the absence of reference data for whole organ impact, the model was subject to accelerations and hit by a rigid bar. Only the stability and cost were compared.

Regional model

*Model development***.** For the regional model, only the Ls-Dyna3D version was developed as of now. The geometry provided by the Full Body Model COE from the GHBMC was used. For the meshing constraints, a particular attention was paid to limit the amount of free space between components as this space could affect the proper transmission of loads from component to component. The element type and formulations were based on the results from the organ level evaluations. An overview of the components is provided in Figure 3. The solid organs were modeled using tetrahedral elements surrounded by a membrane shell to simulate the capsule. For the hollow organs, formulations based on monitored volumes with high bulk moduli were used for the organ contents while the membranes were modeled using shell elements (mostly quad). The same approach was used for the major vessels of the abdomen. They include branches connected to the solid organs internally by coincident nodes. Since the lumbar spine was not targeted for injury prediction, it was modeled using rigid bodies (one per vertebra) connected using 6 d.o.f. joints. Muscles were modeled using tetrahedral formulations, and the geometry was altered to provide a continuous mesh near the linea alba. Since gaps were present between these components after meshing, a component was defined to fill the largest gaps (i.e. more than 1 to 2 mm thick). This component was obtained by Boolean operation between the abdominal cavity and the organs. It was meshed with tetrahedral elements and called fat. The anatomical relationships between organs were simulated based on anatomical descriptions: springs were used for ligaments that are well described, and tied or sliding contacts were defined on all surfaces.

The mesh of the abdominal region includes approximately 270,000 deformable elements. All elements met the quality guidelines defined by the consortium. It uses only elastic, viscoelastic and hyperelastic viscous types of material laws, with properties adapted from the literature. The lumbar spine model parameters were selected based on Demetropoulos (2001) isolated lumbar spine tests. The model time step for the assembled region was above 1μs.

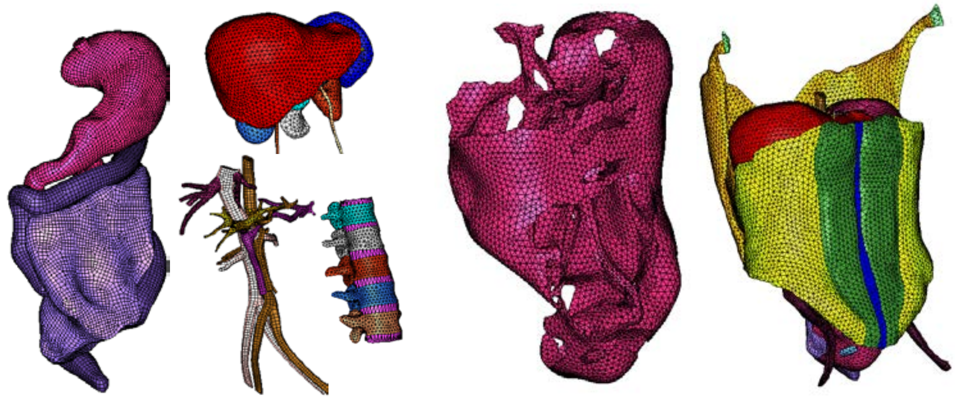


Figure 3: Overview of the components meshed for the abdominal region. From left to right: hollow organ models, vessels, solid organs, lumbar spine, fat component and abdominal muscles

*Model global validation***.** Because of the anatomical relationships with the thorax and the pelvis, exercising the abdomen model on its own would have little meaning. The model was exercised first based on a simplified and preliminary assembly with the other regions. Then, the full model provided by the Full Body Center and described in a paper also available in the current proceedings was used. The description of the full model will not be repeated here.

In order to exercise the model, various tests setups from the literature were used. Since some of these are fairly strenuous (e.g. 48kg bar impact at 9m/s to the mid abdomen from Hardy et al., 2001), they also served as stability tests for the model. The list of test setups is the following:

- Rigid bar impacts to the mid abdomen at 6m/s and 9m/s (pendulum test, 48kg) from Hardy et al. (2001)
- Rigid bar impact to the upper abdomen (pendulum test, 48kg, 6m/s) from Hardy et al. (2001)
- Low speed belt abdominal loading to the mid abdomen (pendulum test, 32kg, 3m/s) from Hardy et al. (2001)
- Airbag surrogate loading to the mid abdomen (high speed low mass test) from Hardy et al. (2001)
- Belt pretensioner loading to the mid abdomen from Forster et al. (2006)
- Belt loading at two different speeds (MHA and PRT) from Lamielle et al. (2008)
- Pendulum impact to the upper abdomen in oblique direction from Viano et al. (1989).

This led to nine impact conditions simulated with the full model. Two additional setups from Kremer et al. (2011) were also modeled later when they became available in the literature but are not presented here. While these setups can provide an overview of the model response to impact, none of them really exercises the lumbar spine over a large range of motion. However, large lumbar flexions could be critical for the interaction between organs. An additional test setup was added solely to check the kinematics of the model in lumbar flexion. It is loosely adapted from the Hybrid III torso flexion calibration test: the pelvis was fixed and a cable attached near C7 was used to pull the model into flexion. The speed was selected to lead to a complete flexion in 200ms. Since there are no reference data for this setup, only the stability of the simulation was considered.

All simulations were run using LsDyna3D HP-MPI Linux X86_64 version on a cluster of hexacore Xeons connected by an Infiniband switch. Typical elapsed times for 100ms of simulation on 48 cores (4x2x6) were 3 hours for the simplified model (time step: 1µs) and 15 hours for the fully deformable model (time step: 0.3µs). The time steps remained constant throughout the simulations.

RESULTS

Organ modeling

First, basic simulation work was performed in LsDyna3D with the five models of different detail and element formulations. With the same material properties, the global responses of different models were very similar (Figure 4, left), and the differences were small compared with the width of the reference corridor based on literature data. When comparing the average stress in the cortex zone previously defined (Figure 4, right), similar responses were also obtained despite very different number of elements in the zone of comparison. The maximum difference of the average stress was highlighted on the figure. Overall, considering the large uncertainty on material parameters, all responses were considered as admissible.

Then code comparisons were performed using the tetra medium kidney model. Equivalent formulations were found for the three codes. For equivalent material properties, the global responses of the three models were very close from one another, and the LsDyna3D and Pamcrash responses were virtually indistinguishable (Figure 5). Similarly, the responses of the cortex sphere were almost identical.

Finally stability testing was performed by increasing progressively the impact energy. A model was said to be more stable than another one if a simulation terminated without error for higher impact energies. The results will only be summarized here:

- Tetra models were more stable than corresponding hexahedral models, and coarse models were more stable than detailed models,
- Formulations with single integration point were more stable than formulations with multiple integration points
- Some stability issues were observed in the hyper-elastic viscous law of the Radioss code (time step drop, energy balance issues).

Overall, tetrahedral elements did not show specific limitations in terms of response and stability for the three codes tested. Based on these observations, it was decided to use tetrahedral elements for subsequent developments as high quality tetra models seem easier to generate than hexa counterparts for a complex geometry.

Similarly, after performing simulations on the stomach in LsDyna3D and Pamcrash, a formulation using monitored volumes and deformable shell membranes were selected due to cost and stability considerations.

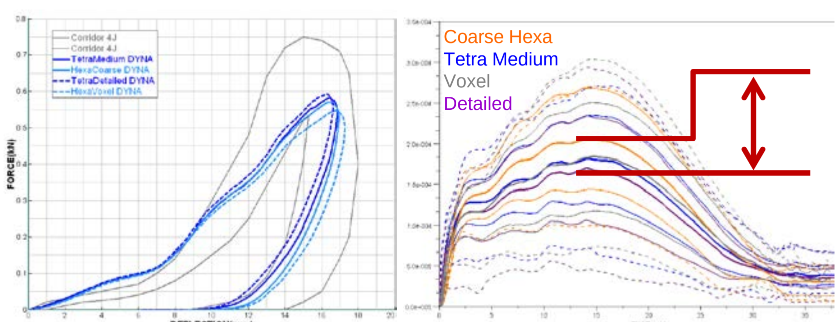


Figure 4: Left: global responses of the five LsDyna3D models with different details and formulations. Right: Average (thick solid lines) plus or minus one standard deviation (thin solid line) and extreme values (dashed lines) obtained within the cortex zone

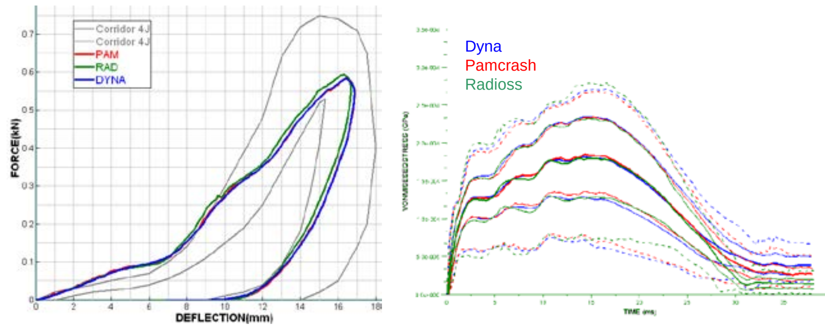


Figure 5: Left: global responses of the tetra medium model in the three codes vs. experimental data. Right: Average (thick solid lines) plus or minus one standard deviation (thin solid line) and extreme values (dashed lines) obtained within the cortex zone

Regional model

Overview of simulation results: all tested setups terminated successfully with the full deformable human model, even for the most strenuous impact condition (9m/s setup from Hardy et al. 2001). Illustrations of the model responses for three of the validation setups are provided in Figure 6.

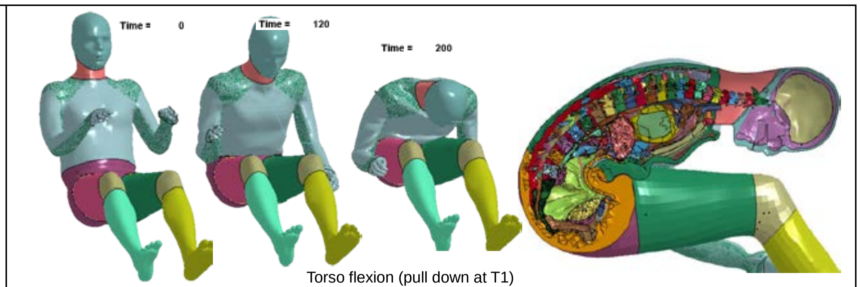
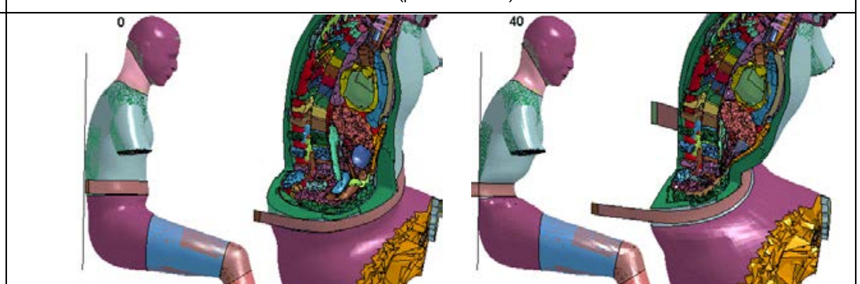
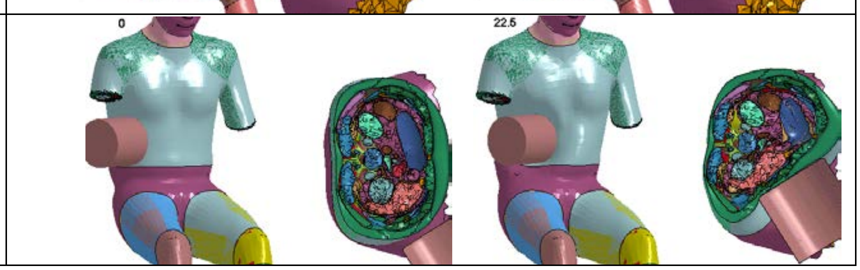
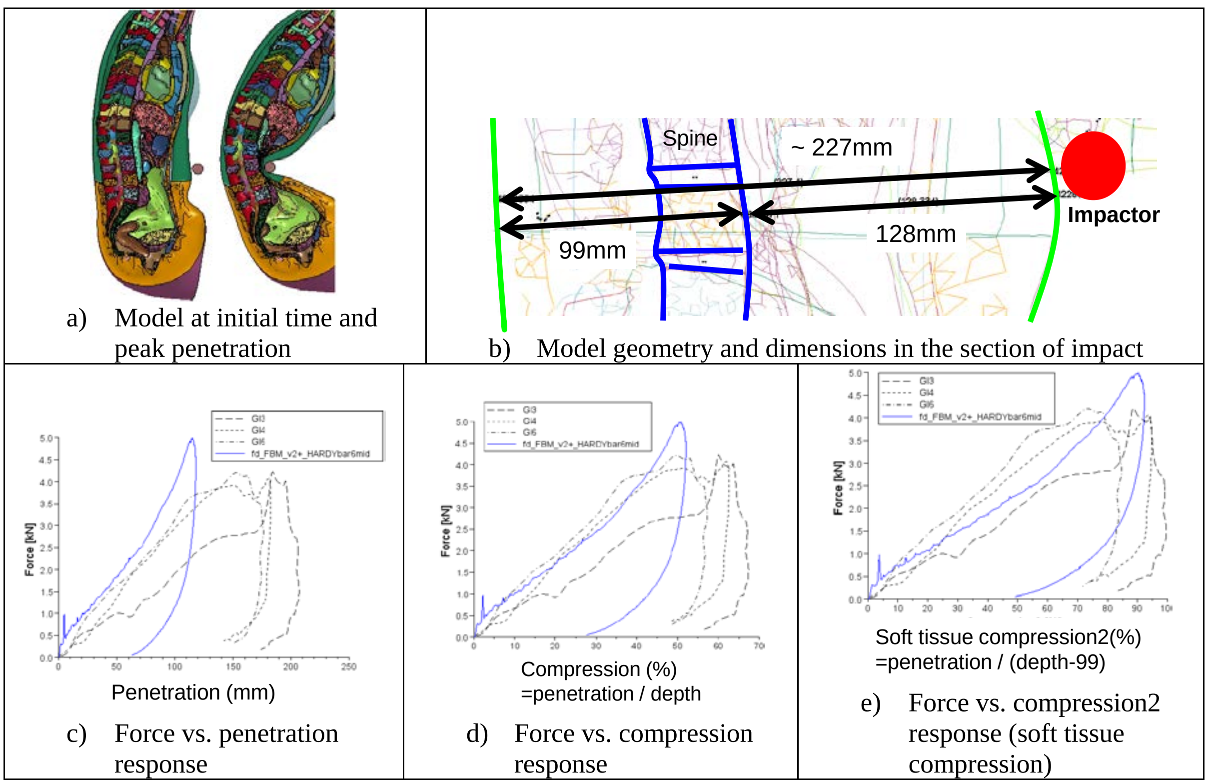
Torso flexion test setup	
Belt loading setup (MHA) from Lamielle et al (2008)	
Pendulum impact setup from Viano et al. (1989)	

Figure 6: Overview of three simulation setups. Illustrations are provided in initial position and near peak deflection.

Quantitative comparison with the literature data: at first, results were compared directly between simulation and test data using force and deflection/penetration results. However, this approach quickly showed its limitations. For the 6m/s bar impact from Hardy et al. (2001), peak forces seemed to be matching the experimental responses but the penetration predicted by the model was much smaller than the test results (Figure 7c). Scaling the results using equal stress equal velocity approach did not solve the issue. However, further increase of the penetration in the simulation was not possible (Figure 7a) due to geometrical constraints: the impactor was already very close to the spine and the penetration obtained in the tests would place the impact behind the spine in the simulation. After investigation, it appeared that the PMHS tested had larger abdominal depths than the model, and that matching the penetration would not be possible. As a consequence, two alternate metrics were defined for comparisons. The compression, which is the penetration divided by the abdominal depth, is commonly used in abdominal studies. It was calculated for both model and experimental datasets and the responses using this metric are provided in Figure 7d. This still led to predictions of the displacement variable that were too low for the model, with no possibility of increase. A second metric was then defined based on the idea that the impactor may also have stopped near the spine in the test, and that spine contact should be taken into account for the normalization. This metric, called compression2 or soft tissue compression in the plots, correspond to the compression of the soft tissues anterior to the spine (the only ones able to compress). Since the spine thickness was not available for the literature results, it was assumed that the spine would have similar dimensions as the spine in the model, and that most of the thickness variability came from the soft tissues. Using this variable suggested that the impact could also have stopped near the spine in the tests (Figure 7e). This variable was used for subsequent results and a summary of the model vs. experimental responses is provided in Figure 8.



a) Model at initial time and peak penetration

b) Model geometry and dimensions in the section of impact

c) Force vs. penetration response

d) Force vs. compression response

e) Force vs. compression2 response (soft tissue compression)

Figure 7: definition of comparison metrics using the Hardy et al. (2001) 6m/s bar impact for illustration

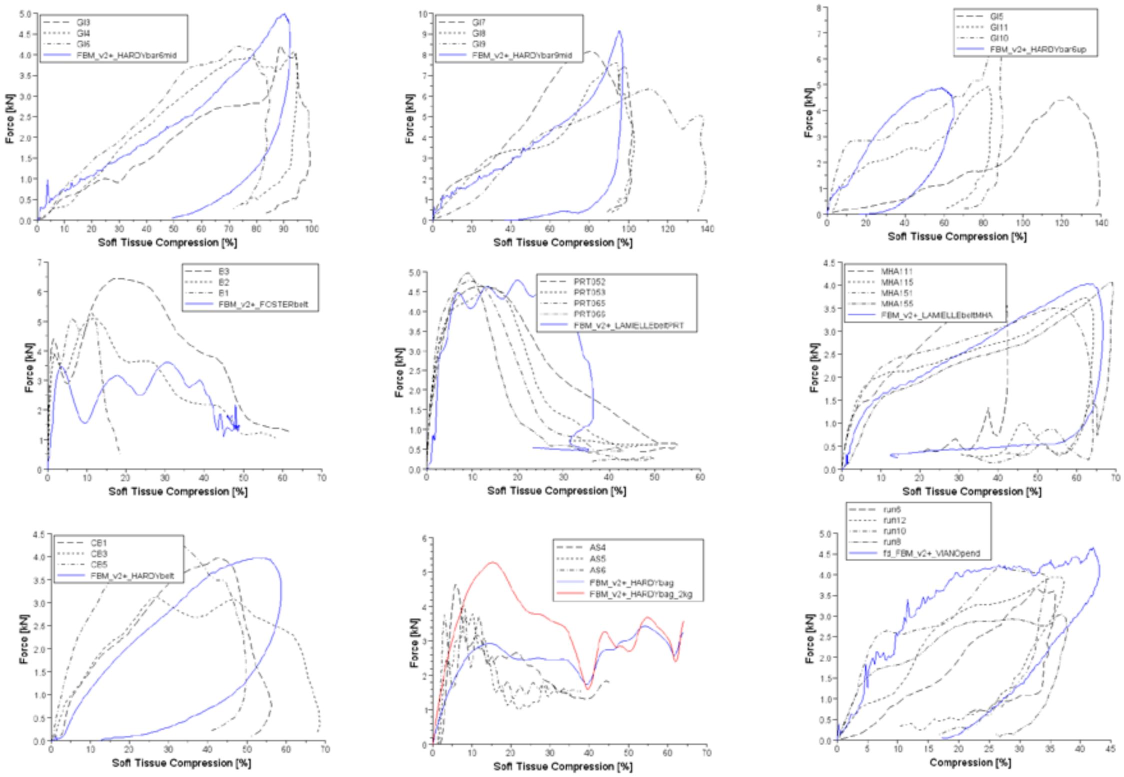


Figure 8: overview of the model responses for the nine experimental setups. The test setup name is provided in the legend of every graph

Overall the model response was close to the literature data in most cases. There were larger differences with the upper abdomen bar impact from Hardy et al. (2001), the belt loading from Forster et al. (2006), the PRT condition from Lamielle et al., (2008) and the surrogate airbag condition also from Hardy et al. (2001). These cases were further investigated. For the upper abdomen impact, it was found that numerous rib fractures had occurred during the tests. The lack of fracture simulation in the version of the model that was used was considered to be the likely cause of the soft tissue compression mismatch in that setup. For the other cases, it was found that the setups were sensitive to abdominal mass variations as illustrated in Figure 9. In other words, a limited mass mismatch between model and experiment could be a possible cause of the response differences. It was also verified that other setups (e.g. 6m/s bar impact to mid abdomen) were largely insensitive to such small mass variations.

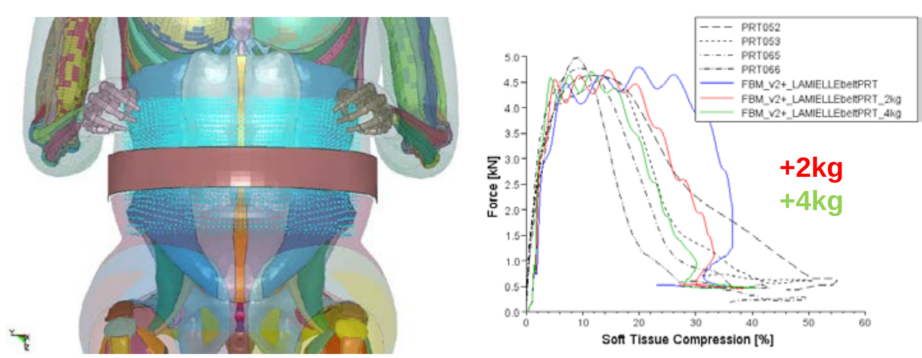


Figure 9: effect of added abdominal masses on the model response.

DISCUSSION AND CONCLUSIONS

A full finite element model of the abdomen was developed as part of the GHBMC coordinated effort. The model uses formulations that were previously evaluated on organ models. These evaluations suggested that equivalent formulations can be found for the simulation of soft tissues in the three leading finite element codes using for crash simulation if common material laws are used. More importantly, they suggest that, for soft organs and the configurations that were tested, tetrahedral elements had better or equivalent performance in terms of stability, response and cost when compared with hexahedral elements. This has implications regarding the ability to describe the complex shape of organs with high quality finite element meshes as the hexahedral meshes typically used for impact simulation are difficult to generate.

After coupling with models of other regions, the abdominal model was subjected to nine impact setups and one kinematic condition. The simulations terminated without error in all cases even for the most strenuous conditions. This suggested a good stability performance, which was the first of the objectives. The model responses were shown to be close to the experimental results for most test setups after a few adjustments (third objective). First, geometrical differences between model and experimental subjects were found to affect the comparison, with most PMHS having higher abdominal thickness than the model. A metric representing the percentage of soft tissues anterior to the spine was defined to be able to perform meaningful comparison. However, the external mismatch may be associated with an internal mismatch and it is unclear how this will affect the results when looking at injury prediction. Second, this scaling of dimensions does not account for the corresponding mass present in the PMHS that could affect the response. Some of the tests setup were found to be particularly sensitive the mass of the abdominal regions, and this factor should be considered in future studies.

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