Development of a multibody 50th percentile model for Euro NCAP's Pedestrian Test Protocol

Juan M. Asensio-Gil, Richard Lancashire, Erik J. Dijkstra, Alberto Carnicero, Francisco J. López-Valdés

Abstract The aim of this study was to develop and validate a simplified ellipsoid-based human multibody model (s-HBM) of a 50th percentile pedestrian for Euro NCAP's Pedestrian TB024 certification protocol in Simcenter Madymo software. To this end, a three-step process was followed using Post Mortem Human Subject (PMHS) data as well as Simcenter Madymo Active Human Model simulation results without muscle activation as reference to ensure the biofidelity of the study model. First, a set of localized lateral and frontal blunt impact tests were performed to allow independent and precise modifications to the model. Then, overall behaviour and robustness were analyzed by means of a full-body lateral impact test against a rigid simplified vehicle. Finally, TB024 tests were simulated to verify whether the developed model complied with Euro NCAP's specifications. It was found that the s-HBM passed the certification protocol while showing biofidelic behaviour for all the tests performed in this study, although certain target trajectories of the protocol tests could be further improved in a future study. Additionally, further research is needed to properly address the lateral deflection of the thorax in this model.

Keywords Biofidelity, human body model, multibody, pedestrian, TB024.

I. INTRODUCTION

The World Health Organization states that approximately 1.35 million road users died as a result of road traffic accidents in the world in 2016, which implies almost 3,700 fatalities per day [1]. According to the same source, pedestrians accounted for 0.31 million (23%) of all road traffic-related fatalities, making them the second riskiest type of user after the occupants of 4-wheeled vehicles. In some regions, such as Europe or Africa, this percentage increases to 27% (2.51 deaths per 100,000 population) and 40% (10.64 deaths per 100,000 population), respectively. More recent data, provided by the Global Health Data Exchange [2], shows that between 387,000 and 503,000 pedestrians died in 2019, and that the Disability-Adjusted Life Years (DALYs) lost was between 20.5 and 26.2 million years.

Additionally, the European Transport Safety Council defines walking as the second riskiest mode of transport per population and travel distance after motorcycle, with a value of 6.4 deaths per 100-million-person-kilometers, and as the third riskiest per population and travel duration after motorcycle and bicycle, with 25 deaths per 100-million-person-travel hours [3]. The Mobility and Transport department of the European Commission states that 69% of total pedestrian deaths in 2020 involved a collision with a car [4].

For all these reasons, different concepts regarding the categorization of factors involved in road safety have been developed, with the most recent and accepted concept – which also aligns with the Vision Zero project – being the Safe System and its five pillars: safe roads and roadsides, safe vehicles, safe road use, safe speeds, and post-crash response [5]. The first of the pillars, the improvement of existing infrastructure, has proven to be one of the most effective measures in terms of both injury and cost-benefit rates with, for instance, the construction of pedestrian-only streets. Elvik *et al.* estimate that with this measure alone, pedestrian injuries can be reduced by up to 60% [6]. However, infrastructural interventions are not always possible due to, for instance, lack of sufficient physical space due to the presence of immovable objects, in which case vehicle design and Advance Driver Assistance Systems (ADAS) assume the most important role in pedestrian safety.

The TB024 or Pedestrian Test Protocol

The European New Car Assessment Programme (Euro NCAP) includes a whole section dedicated to the safety

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assessment of Vulnerable Road Users (VRUs) as part of its car consumer testing and rating. These tests include physical experiments with dummies and impactors, both for passive protection and for prevention.

Furthermore, Euro NCAP intends to implement extensive virtual testing in their Assessment Programme, as set out in their 2025 Roadmap [7], in order to add robustness as well as efficiency to its testing. Manufacturers will also benefit from this initiative because they will be using the same tool (and its advantages over physical testing) for their design process as the one with which they will be assessed. For these reasons, Euro NCAP has been developing the TB024 certification protocol [8] to ensure standardization of the behaviour obtained when using different existing human models (run with different simulation software) as that displayed by pedestrians in VRU tests. For its part, Siemens' Simcenter Madymo software has developed the TB024 pedestrian multibody model family [9].

The corridors and tolerances established by Euro NCAP that are found in the TB024 certification protocol tests (see Methods section, below) are based on advisory studies as part of the CoHerent project [10]. This project, led by Euro NCAP and Graz University of Technology, aims to improve the safety assessment of HBMs by considering the influence of body size and injury predictors, as well as by addressing multiple scenarios [11]. In this way, virtual testing can be implemented extensively in vehicle safety assessment and design in the future.

The advisory study [12] behind the TB024 bulletin of 2018, compares the outputs of the certification tests with up to 15 different TB024 HBMs from various companies, institutes, associations and universities that participated in the CoHerent project, and creates the corridors and tolerances from the average values of those outputs (after removing outliers). The models used are: three from GHBMC (different solvers and versions), five from THUMS (idem), JAMA pedestrian model, Honda HBM, Simcenter Madymo pedestrian models (MB solver coupled with FE solver), JLR humanoid FE model, and ESI PED 50 humanoid FE model. Madymo models are therefore the only multibody dynamics representation in this group (as the rest correspond to FE models).

A need for biofidelity validation

In addition to the model requirements of the previous certification, Simcenter Madymo simulation software aims to go a step further by developing a model that not only complies with TB024 protocol but that is also biomechanically validated. The fact that different reference points of a model happen to follow an experimentally established trajectory (see the Methods section for a more detailed explanation on Euro NCAP's certification protocol) does not ensure biofidelic behaviour in other scenarios, such as vehicle design, and therefore errors could be made in the steps taken towards pedestrian safety.

Advantages of the multibody models for this purpose

A characteristic of multibody models such as the one developed in this study is that they follow the Multi-Body Dynamics method (MBD) instead of the finite element method (FEM) that is present in other models available to researchers and industry. This implies, among other things, that the different components of the system do not deform but penetrate each other. Its main advantage over FEM is its computational cost given that it is a much faster numerical method, allowing for extensive design space exploration, Design of Experiments (DOE) and optimization. This aligns with the concept of virtual testing, as one of its key objectives is efficiency. Hence, the use of a biomechanically validated pedestrian multibody model in both certification protocol and OEMs' vehicle design processes could vastly improve pedestrian safety and complement the results obtained with FEM.

II. METHODS

An existing ellipsoid-based and simplified human multibody model (s-HBM) of a 50th percentile pedestrian was selected as baseline [13].

The process of biomechanical validation and subsequent TB024 certification protocol followed in this project was divided into three main steps, which are described in detail later in this section. First, a set of tests of localized impacts on the s-HBM was simulated so that precise and independent modifications could be made on the model. Second, the overall behaviour of the modified s-HBM was analyzed by means of a full-body lateral impact simulation against a simplified rigid car. Finally, Euro NCAP's TB024 certification tests were performed. In case the s-HBM did not pass these certification simulations, possible causes were analyzed, and the consequent modifications were made in the first step, restarting the process.

Madymo models used in this study

The following models are referred to throughout this study:

1. Simplified human multibody model (s-HBM): composed of 70 ellipsoids, 52 joints, and 52 rigid bodies, and based on the TB024 model [9] with characteristics modified in a previous internal study [13]. This s-HBM is the one being modified and validated within the current study (Fig. 1).

2. Madymo Active Human Model (AHM): facet-based complex model that mimics the human anthropometry by including the skin, bone structure and muscles, and by implementing controllers for the neck, spine, shoulders, elbows, hips and knees, which allows it to maintain the initial position under the influence of external loading. It is composed of 182 rigid bodies, 8 flexible bodies, and 191 joints [14]. This model has been previously validated against PMHS and volunteer data. In the tests conducted in this study, muscle activation was not applied as the simulation results of this model were used as reference together with PMHS data (Fig. 2).

The contact characteristics defined in both models use non-linear force-based definitions. Such contact models exercise a resistant force depending on the penetration distance (N vs m) and distinguish both loading and unloading conditions under different hysteresis models, with the possibility of elastic limit definition. They also consider damping effects, by means of either a coefficient or a penetration-velocity-dependent function (N vs m/s); a force-dependent amplification function (amplification factor vs N) can be applied to either of these methods.

Joint restraint definitions in the s-HBM behave similarly to the contact characteristic for each degree of freedom of a specific joint (from Q1 to a maximum of Q7), with the only difference being that the resistant force applied depends on the displacement of said joint.

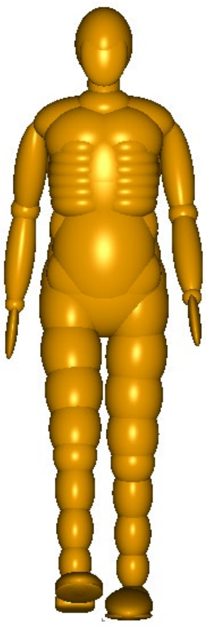

Fig. 1. Simplified Human Multibody Model (s-HBM).

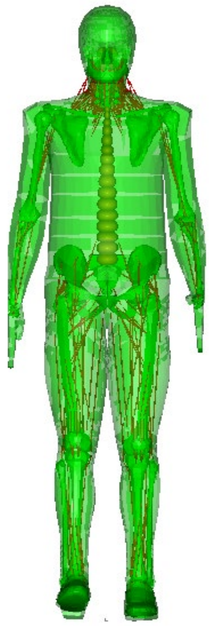

Fig. 2. Active Human Model (AHM).

Localized impact tests

As previously mentioned, the baseline s-HBM model was initially simulated under several different frontal and lateral blunt impact conditions. These were benchmarked against existing Post Mortem Human Subject (PMHS) data for the purpose of comparing the load registered by the impact cell as well as the acceleration or deflection of different body parts, depending on the test, and therefore assessing the biofidelity of the s-HBM model. They were also benchmarked against the Madymo Active Human Model (AHM) 50^{th} Facet Q Model version 3.3 for the overall behaviour correlation of the s-HBM model.

These impact conditions were selected from existing PMHS experiments and involved localized impacts at different velocities ranging from 2.00 m/s to 6.66 m/s with impactors of 23.40 kg. The majority of these tests were lateral impacts since the pedestrian certification protocol, the aim of this study, also involves a lateral impact. Specifically, these tests were: frontal impact on head [15], lateral impact on shoulder [14-15], and two tests involving lateral impact on thorax and lateral impact on pelvis [18] (Table I).

TABLE I
LOCALIZED IMPACT TESTS

PMHS experiment	Velocities (m/s)	Impactor mass (kg)	Subject position	Reference
Frontal impact on head	2.00 & 5.50	23.40	Seated	[15]
Lateral impact on shoulder	4.50	23.40	Seated	[16]
Lateral impact on shoulder	5.50	23.40	Seated	[17]
Lateral impact on torso	3.30 & 5.90	23.40	Seated	[18]
Lateral impact on pelvis	3.46 & 6.66	23.40	Seated	[18]

The simulations corresponding to these tests were performed distally to proximally (head, shoulder, torso, and pelvis), and with boundary conditions as accurately representative of the PMHS experiments as possible. This order follows the kinematic chain of the HBM from the first link located in the pelvis to the final link located in the head (the subject was seated for all tests) in order to avoid, as much as possible, dependencies among the different tests. Meanwhile, an optimization procedure was carried out for each test using Simcenter HEEDS software, where the target responses were gradually fitted inside validation corridors. Then the kinematics and trajectories were, within established acceptance criteria, matched with those of the corresponding AHM model simulation. Correlation was optimized by adjusting, within a ±15% margin of the initial definitions found in the baseline modified TB024 model (which in turn were based on established literature detailed in [9]), the contact characteristics, the joint definitions and, to a limited extent, certain geometries that were relevant to each test. Moreover, additional simulations at higher speeds were performed to check contact stability as well as numerical robustness.

Full-body impact tests

After the modifications made in the s-HBM to satisfy the PHMS results from localized blunt impact tests, the model was exposed to lateral impacts at velocities ranging from 25 kph to 39 kph against a simplified rigid non-deformable vehicle (Fig. 3). These impacts allowed the overall behaviour of the model to be checked in a less computationally expensive simulation before proceeding to the Euro NCAP certification protocol.

The outcome of the s-HBM in these tests was compared to that of the PMHS vehicle tests included in Ishikawa *et al*. [19], as well as with the output of the AHM simulation. Furthermore, an additional test at 55 kph was performed, to verify the model stability and robustness.

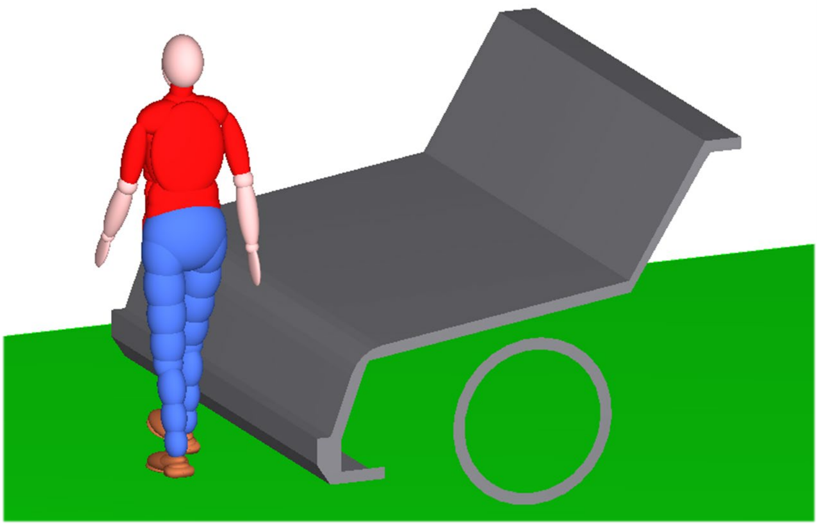

Fig. 3. Full-body lateral impact.

Euro NCAP's TB024 certification protocol tests

Once the s-HBM model was biomechanically validated, the certification tests were carried out following the TB024 protocol of November 2019 [8], by means of a coupled simulation of Madymo with Radioss. Thus, the s-HBM model was simulated in several impact tests involving certified generic vehicles models in Radioss of three different types: Family car, Roadster, and Sport Utility Vehicle (SUV), at 30 kph, 40 kph, and 50 kph (see Table II). This TB024 certification protocol validates, for the simulations involving those car segments and speeds, the trajectories of the model for center of gravity of the head (HC), the center of the T12 vertebral body (T12, equivalent to a specific coordinate system relative to the Lumbar-Up ellipsoid body for the s-HBM model), and the midpoint of the line that joins the center of the left and right acetabula (AC, corresponding to the H-point for the s-HBM model).

In addition, other characteristics, listed below, regarding the validity of the numerical simulation are also required by Euro NCAP's protocol:

1. Contact force is zero at the start of the simulation.
2. Total energy remains constant within a 15% tolerance.
3. Hourglass energy is less than or equal to 10% of the total energy.
4. Contact energy at the start of the simulation is less than or equal to 1% of the total energy.
5. The artificial energy (contact and hourglass types) is less than or equal to 15% of the total energy.
6. The artificial mass increase for moving parts is less than or equal to 3%.
7. HIT (Head Impact Time) is within tolerance.
8. Contact force is within corridors (visual check).

TABLE II
TB024 PROTOCOL TESTS

Car Type	Velocities (km/h)
Family Car	30.0, 40.0, 50.0
Roadster Car	30.0, 40.0, 50.0
SUV Car	30.0, 40.0, 50.0

III. RESULTS

For all the graphs present below, the s-HBM curves correspond to the final optimized s-HBM model (including all final modifications made to it in this study), the corridors or experiment curves to the ones defined in the PMHS tests, the baseline curves to the original s-HBM model (previous modifications), and the AHM curves to the response of the Active Human Model's passive response. Furthermore, information regarding the main modifications made in each localized impact test is provided.

Frontal impact on head

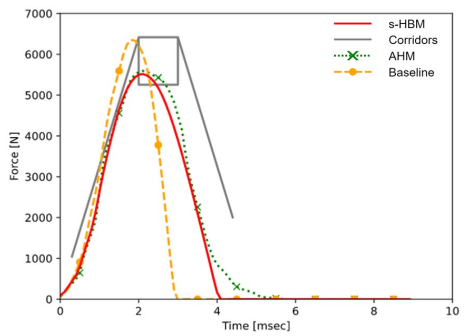

Fig. 4. Frontal impact on head 2.00 m/s.

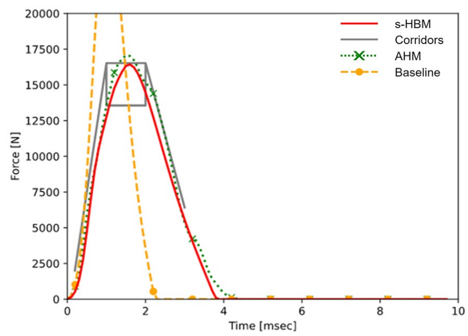

Fig. 5. Frontal impact on head 5.50 m/s.

As seen in Fig. 4 and Fig. 5, the baseline s-HBM model was overly stiff, which resulted in a response lying outside the corridors. Hence, the head contact characteristic was modified by scaling down the force values for the loading and damping amplification functions, and by reducing the damping coefficient. As a result, the response was fitted within validation corridors in both cases.

Lateral impact on shoulder

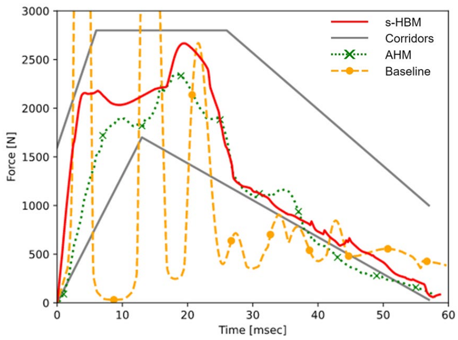

Fig. 6. Lateral impact on shoulder 4.50 m/s.

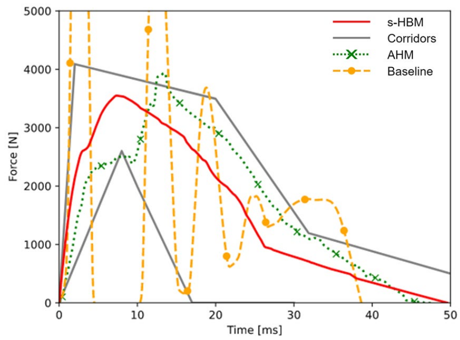

Fig. 7. Lateral impact on shoulder 5.50 m/s.

For this set of lateral impact tests on the shoulder, the baseline response was highly unsteady and almost completely outside corridors. This was caused by an especially aggressive and underdamped contact interaction. Also, when compared to the AHM simulation, the kinematic behaviour was significantly different. Hence, the baseline model's shoulder definitions were considered as non-biofidelic, both from a dynamic and kinematic point of view. For this reason, different degrees of freedom of the joint restraints present in both clavicle (universal-translational joint) and shoulder (universal joint) were modified until the kinematics matched the AHM model. The upper shoulder contact characteristic was also modified to fit the response within the experimental corridors (Fig. 6 and Fig. 7). In particular, damping coefficient was increased for shoulder Q2, and clavicle Q1, while the clavicle Q3 load and unload functions' force values were scaled up. Regarding the upper shoulder contact characteristic, damping coefficient value was increased, factor values were scaled up in the damping amplification function, and loading and unloading function force values were scaled down.

Lateral impact on torso

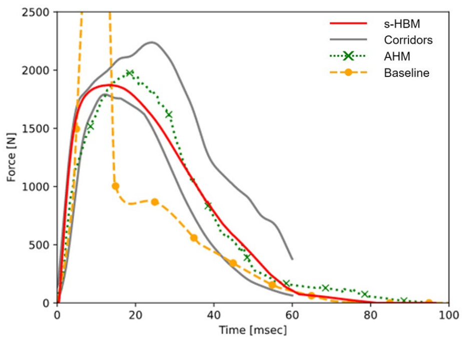

Fig. 8. Impactor force in test 3.30 m/s.

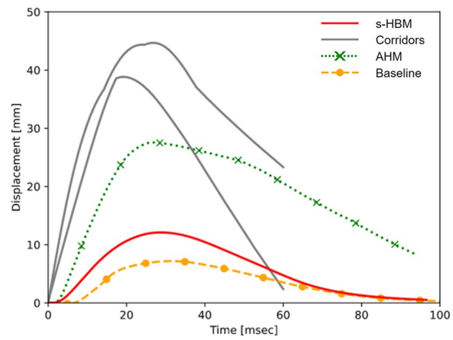

Fig. 9. Lateral thoracic deflection in test 3.30 m/s.

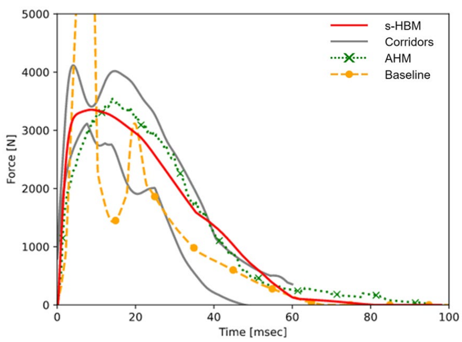

Fig. 10. Impactor force in test 5.90 m/s.

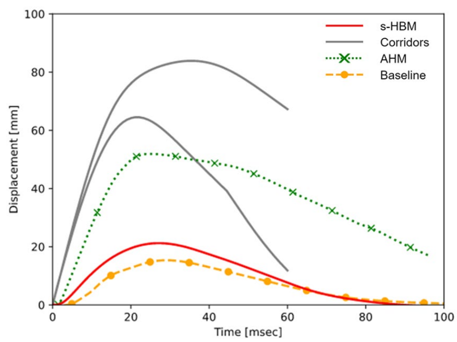

Fig. 11. Lateral thoracic deflection in test 5.90 m/s.

The original s-HBM model presented a stiff response followed by a sudden fall in the contact force recorded by the impactor. This was caused by the geometry of the ellipsoid present in the upper back part of the torso, which forced the impactor to hit the s-HBM model in a non-biofidelic way. Reducing the shortest axis of the relevant ellipsoid by 2 mm solved the issue. Furthermore, the stiffness of the response was diminished by modifying the contact characteristics of several ellipsoids present in the lateral part of the torso, so that the force response was fitted inside validation corridors (Fig. 8 and Fig. 10). Specifically, the force values of the loading functions of the back and lateral torso ellipsoids were scaled down, while the factor values of the damping amplification function of the lateral torso ellipsoid were scaled up. However, as seen in the thoracic lateral deflection graphs of Fig. 9 and Fig. 11, the responses are far from their respective corridors.

Lateral impact on pelvis

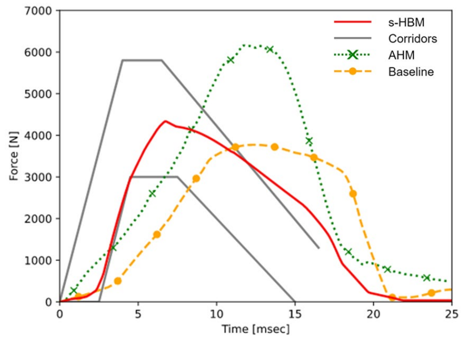

Fig. 12. Impactor force in test 3.46 m/s.

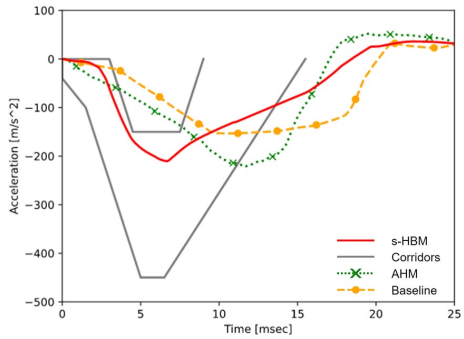

Fig. 13. Sacrum acceleration in test 3.46 m/s.

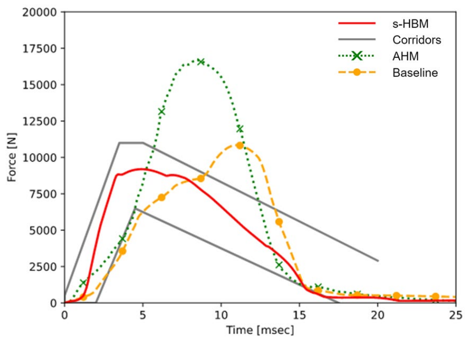

Fig. 14. Impactor force in test 6.66 m/s.

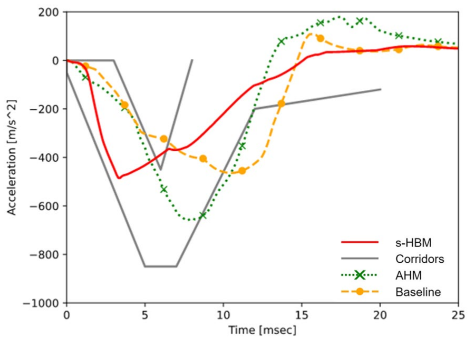

Fig. 15. Sacrum acceleration in test 6.66 m/s.

In this case, the s-HBM response needed to be initially stiffer than that of the baseline in order to fit inside the PMHS corridors. Pelvis and hip contact characteristics were modified so that the initial response would also be similar to that of the AHM model, while still being inside corridors. Said modifications included increasing the damping coefficients and scaling up the force values of the loading functions as well as the factor values of the damping amplification functions, for both the hip and pelvis contact characteristics. These intermediate results are not shown above, since only the final model responses are included in this study. As previously explained, additional changes had to be made due to the model not initially passing Euro NCAP's certification protocol in two of the tests. Pelvis and hip contact characteristics were further modified (Fig. 12 to Fig. 14), so that the initial response was not as stiff as the AHM model's, but more similar to that of the baseline model and, overall, still inside corridors. These smaller changes consisted of scaling down some of the first force values of the loading functions of the pelvis and hip contact characteristics.

Full-body lateral impact

Full-body lateral impact tests were based on and validated with the Ishikawa *et al*. PMHS vehicle tests. These tests were performed at 25 kph, 32 kph, and 39 kph. Results shown below correspond to the 39 kph test but can be considered representative of the results for 25 kph and 32 kph. The z axis defined in Fig. 17 to Fig. 22 corresponds to the vertical axis (normal to the ground plane), whereas the x axis is coincident with the direction of displacement of the simplified rigid vehicle (Fig. 3).

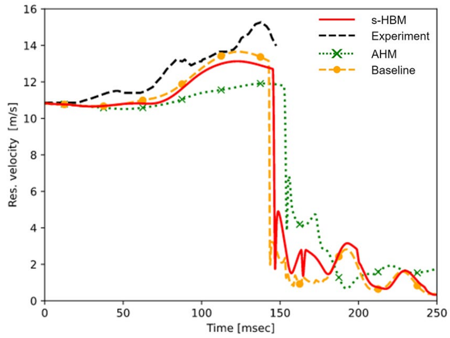

Fig. 16. Resultant velocity of the head in test 39 kph.

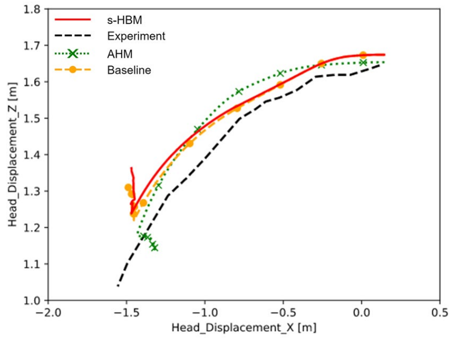

Fig. 17. Displacement of the head in test 39 kph.

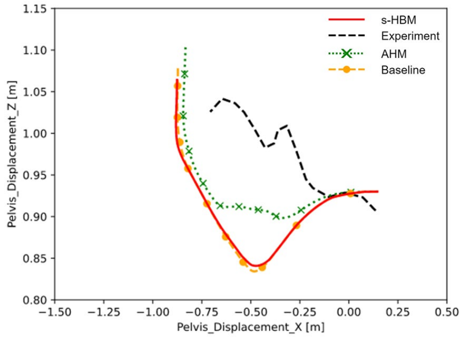

Fig. 18. Displacement of the pelvis in test 39 kph.

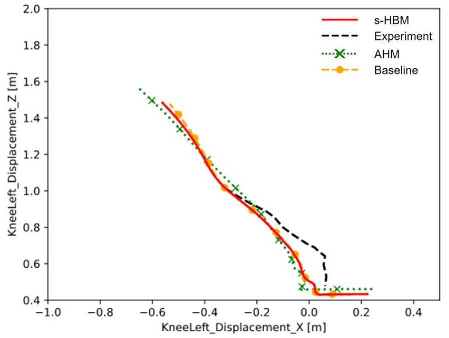

Fig. 19. Displacement of the left knee in test 39 kph.

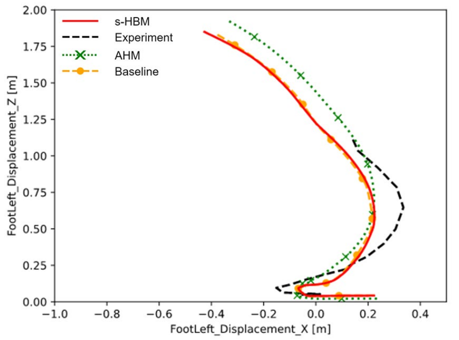

Fig. 20. Displacement of the left foot in test 39 kph.

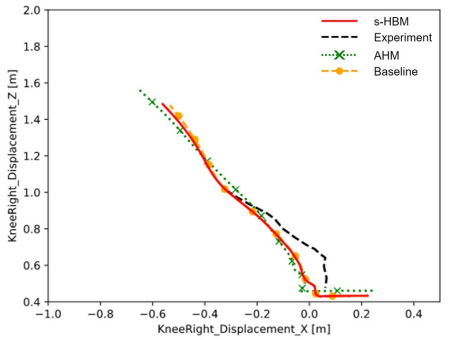

Fig. 21. Displacement of the right knee in test 39 kph.

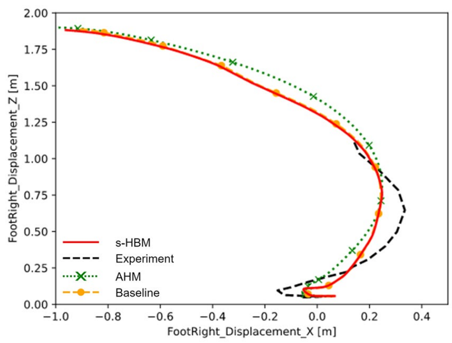

Fig. 22. Displacement of the right foot in test 39 kph.

Overall, the response of the s-HBM model is similar to that of the Active Human Model and close to the experimental PHMS sled tests results curves (Fig. 16, Fig. 17 and Fig. 19 to Fig. 22). Head impact time is 145.4 ms, which results in a 2.1 ms variation from the experimental data (147.5 ms). On the other hand, the head's resultant velocity at impact is 12.7 m/s, while the PHMS data recorded 13.9 m/s (8.6% variation). In the case of Fig. 18 (z-displacement vs x-displacement of the pelvis), the difference between the optimized s-HBM, the AHM and the experimental curve is significant but not relevant for the purpose of this test, as explained in the Discussion section.

Euro NCAP's TB024 Certification Protocol

Results shown below correspond to Euro NCAP's 50 kph impact tests for the segments of Family car, Roadster, and SUV. Tests for 30 kph and 40 kph were also performed, but the tests with the highest velocity were found to be the most restrictive ones. Additionally, Table II shows the percentage of deviation of the HIT time for each test with respect to Euro NCAP's reference.

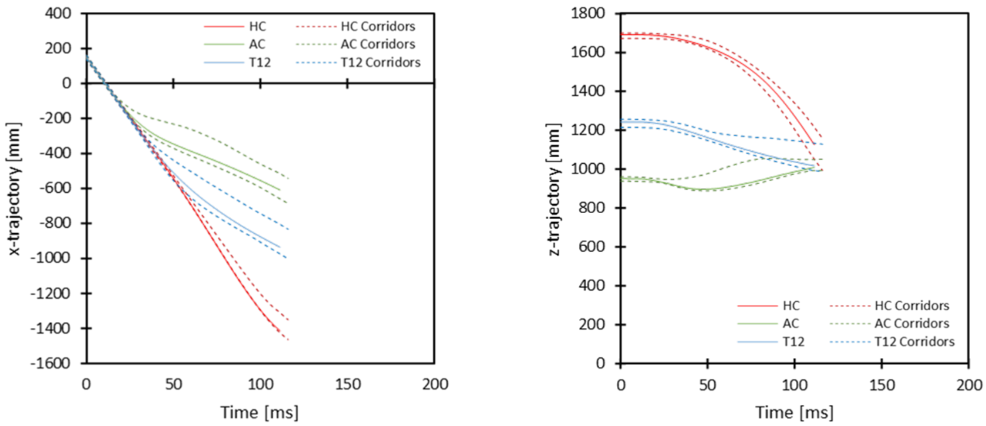

Fig. 23. Family car test 50 kph.

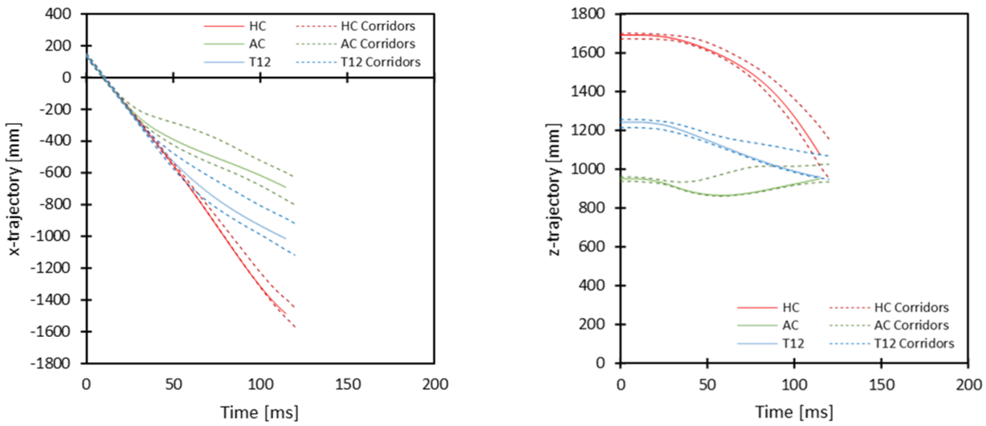

Fig. 24. Roadster test 50 kph.

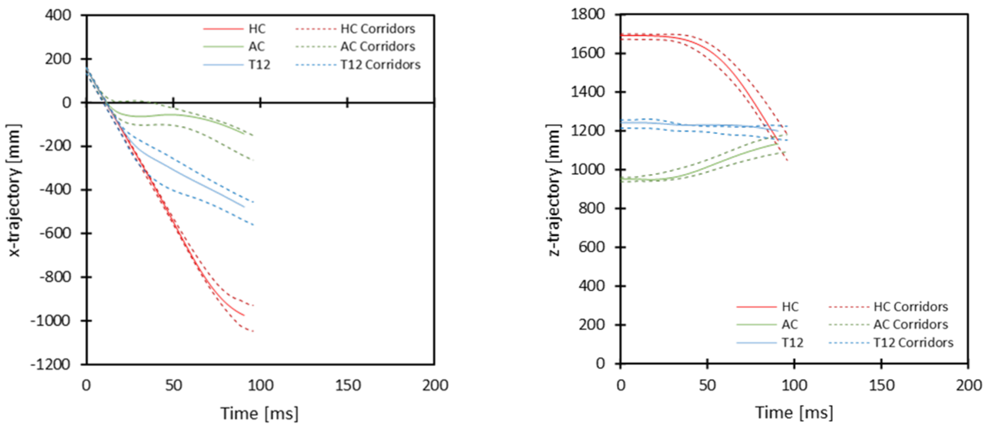

Fig. 25. SUV test 50 kph.

All segment cars at all velocities, and all outputs, including the trajectory of the pelvis (H-point), were found to be within tolerance or corridors (Fig. 23 to Fig. 25) and passed all visual checks. Some of the trajectories of the different tests (which are detailed in the Discussion section) were, for small periods of time, outside corridors but within Euro NCAP's established tolerance.

TABLE III
S-HBM HIT RESULTS IN THE TB024 TESTS

Car type	Test velocity	Value (ms)	Target (ms)	Tolerance (ms)	Resultant deviation (%)
	30 kph	169.2	167.3	[157.2,177.4]	+1.10
Family Car	40 kph	135.3	135.55	[129.1,142.0]	-0.20
	50 kph	111.3	112.6	[108.5,116.7]	-1.20
	30 kph	175.3	175.0	[163.5,186.5]	+0.20
Roadster Car	40 kph	140.9	141.0	[134.3,147.7]	-0.10
	50 kph	114.6	117.25	[112.9,121.6]	-2.30
	30 kph	143.1	137.1	[127.4,146.8]	+4.40
SUV Car	40 kph	109.9	109.3	[103.2,115.4]	+0.50
	50 kph	90.7	92.65	[88.3,97.0]	-2.10

The Head Impact Times (HIT) of the developed s-HBM model obtained from the different certification tests have been found to be not only inside tolerances but very similar to the established reference, with a range of percentage of deviations from the reference of -2.30 to +4.40, an average of +0.03, and a standard deviation of 1.40.

IV. Discussion

The methodology followed in this study involved the comparison of the model being developed against three different sets of data, moving from more simple scenarios based on localized impacts to more complicated ones, including full vehicle vs. pedestrian scenarios. The initially developed s-HBM did not pass the TB024 certification because the T12 trajectory fell outside validation corridors and their respective tolerances at the 40 kph and 50 kph tests that involved an impact against a SUV vehicle, and this required a second round of modifications. After analyzing the results, the pelvis contact characteristic definition was found to be solely responsible, since the vehicle's height caused it to impact directly on this part of the model.

Thus, the lateral impact on pelvis test was re-simulated, focusing on modifying different parameters of the pelvis' contact characteristic to a less stiff behaviour while still displaying biofidelic kinematics and dynamics. Afterwards, the rest of the localized impact tests were also repeated to verify that no other responses were altered due to this adjustment, even though the tests were considered as independent from each other (see the Methods section). Full-body lateral impact test was also re-simulated as a final confirmation or validation phase before the TB024 protocol tests.

On this second attempt of Euro NCAP's certification tests, all trajectories were found within corridors and all additional requirements (see Methods section) were met. Nevertheless, this study believes that the x-coordinate trajectories for HC in the 50 kph tests for Family cars and Roadsters, as well as the z-coordinate trajectories for T12 in the 40 kph and 50 kph tests for SUV cars, could be improved with further development (with for instance, a specific DOE and optimization process) given that these signals are currently, for the s-HBM developed in this study, outside corridors for a small period of time (but within Euro NCAP's established tolerances).

Regarding the significant difference found between the pelvis trajectory response and the experimental data of the full-body lateral impact test (see Fig. 8 in Results section), a possible explanation for this could be the simplifications made regarding the 4-wheel vehicle used. Instead of a deformable car, such as in the Euro NCAP's certification protocol, a rigid one was utilized, giving results that can be analyzed only as an overall verification. Therefore, no exact results or conclusions were meant to be obtained, but rather an insight of the s-HBM's general behaviour and robustness, to evaluate whether it was coherent to simulate the (much more computationally expensive) TB024 tests, or whether more modifications to the model were required. Hence, and considering the whole set of the results for this test, the TB024 certification protocol was expected to pass for all segment cars and velocities in both of the attempts made in this study, although in the first attempt the s-HBM did not pass it for two velocity sets of the SUV vehicle tests (as explained above).

On the other hand, the lateral deflection response for the lateral impact on torso test of the localized impact tests section was found to be clearly dissimilar to that of the PMHS data. A possible explanation for this is an unsuitable definition of the thoracic lateral deflection in the s-HBM model. Due to the nature of multibody dynamics, this deflection is measured as a displacement of a particular point of the model or as a relative displacement between a specific point and a reference located in the sagittal plane of the model, rather than as the deformation itself, as in FEM dynamics. Several attempts were made in this study to redefine this parameter to obtain more coherent and robust results, but without success. For this reason, further study on this matter is still necessary to properly address lateral deflections of the thorax in this model.

Finally, it is necessary to contextualize the biofidelity of the model developed in this study within the field of existing TB024 HBMs. The models used as reference for the development of the requirements found in the TB024 certification protocol [12] have been broadly validated in terms of biofidelity using different data sources, references, methodologies, simplifications, and even solvers. Thus, if a specific model is found within corridors (that is, its response is very similar to that of the rest of the models), it can be concluded that it presents a biofidelic kinematic response at a global level. This is the case of the s-HBM developed in this study, but also of the model that was used as baseline for it. Therefore, the contribution of this new s-HBM lays on a significantly improved kinematic and dynamic biofidelity at a local level without compromising the global level's kinematic behaviour and robustness, thanks to small and precise adjustments (see Methods section) that could be considered to be within human variability.

V. CONCLUSIONS

A biofidelic pedestrian 50th percentile multibody model was developed for Madymo software from an existing simplified Human Body Model (s-HBM) that complied with Euro NCAP's TB024 certification protocol; and using as reference Post Mortem Human Subject (PMHS) data and Active Human Model (AHM) simulation results. This was achieved by means of modifying certain contact characteristics, joint definitions and restraints and, to a limited extent, geometries. The model was first benchmarked against localized blunt impact tests to improve its response, and then verified in a full impact test for overall behaviour and robustness, before simulating Euro NCAP's certification protocol tests (to confirm that the newly developed model still followed Euro NCAP's TB024 protocol). In a second attempt of this three-step process, the model managed to pass the TB024 tests while manifesting biofidelic responses in all tests present in this study.

VI. REFERENCES

[1] World Health Organization (WHO) (2018) *Global status report on road safety 2018.*

[2] Global Health Data Exchange (GHDx) (2019) *C.1.1.1 Pedestrian Road Injuries,* 2019. Internet: http://ghdx.healthdata.org/gbd-results-tool, Retrieved 01/03/2022.

[3] European Transport Safety Council (ETSC) (2003) *Transport Safety Performance in the EU - A Statistical Overview*, 2003.

[4] European Commission | Mobility and Transport (2021) *Road safety: European Commission rewards effective initiatives and publishes 2020 figures on road fatalities*.

[5] Parliamentary Advisory Council for Transport Safety (PACTS). *Safe System*.

[6] Elvik, R., Høye, A., Vaa, T. & Søresen, M. (2009) *The Handbook of Road Safety Measures: Second Edition,* pp.397–732. Emerald Group Publishing Limited, Bingley, UK.

[7] Euro NCAP (2017) *2025 Roadmap: In pursuit of vision zero*.

[8] Euro NCAP (2019) Pedestrian Human Model Certification Version 2.0. TB024, November 2019.

[9] Simcenter Madymo (2021) Human Body Models Manual Release 2022.1, 2021.

[10] Klug, C., Feist, F., Sinz, W., Ellway, J. & van Ratingen, M. (2018) Presentation of the CoHerent Project IWG-DPPS*.* Presented at SAE Government/Industry Meeting.

[11] Klug, C., Feist, F., Sinz, W., Ellway, J. & van Ratingen, M. (2017) A Procedure to compare kinematics of Human Body Models for pedestrian assessments. IRCOBI Conference, 2017, Antwerp*.*

[12] Klug, C., Feist, F., *et al.* (2019) Development of a Certification Procedure for Numerical Pedestrian Models. ESV Conference, 2019. Eindhoven, The Netherlands.

[13] Power, J. (2020). Virtual Testing for Pedestrian Safety: Biomechanical Validation of a Human Body Model. Trinity College Dublin.

[14] Simcenter Madymo (2021) Active Human 50th Facet Q Model, version 3.3 (R2021.1). *Quality Report Release Update QAHM50-210601*, 2021.

[15] Melvin, J. W., King, A. I. & Alem, N. M. (1985) AATD system technical characteristics, design concepts, and trauma assessment criteria. Task EF final report. University of Michigan & Highway Safety Research Institute, 1985.

[16] ISO TR9790 (1997) Road vehicles - anthropomorphic side impact dummy - lateral impact response requirements to assess the biofidelity of the dummy. *ISO/TC22/SC12/WG5 (Anthropometric Test Devices)*, 1997, Document N455, Revision 4.

[17] Meyer, E. & Bonnoit, J. (1994) Le choc latéral sur l'épaule: mise en place d'un protocole expérimental en sollicitation dynamique. Mémoire de DEA. *Laboratoire de biomécanique appliquée, faculté de Marseille*, 1994.

[18] Bouquet, R., Ramet, M., Bermond, F. & Cesari, D. (1994) Thoracic and pelvis human response to impact. *Proceedings of the Fourteenth International Technical Conference on Enhanced Safety of Vehicles*,1994, pp.100–109.

[19] Ishikawa, H., Kajzer, J. & Schroeder, G. (1993) Computer simulation of impact response of the human body in car-pedestrian accidents. *SAE paper*, 1993, 933129.