INFLUENCE OF VEHICLE SHAPE AND STIFFNESS ON THE PEDESTRIAN LOWER EXTREMITY INJURIES: REVIEW OF CURRENT PEDESTRIAN LOWER LEG TEST PROCEDURE

Luis Martínez[1], Sabine Compigne[2], Luis J. Guerra[1]
[1] INSIA. Universidad Politécnica de Madrid. UPM.
[2] TOYOTA Motor Europe.

ABSTRACT

This study aims at evaluating the current pedestrian lower leg test procedure with respect to the human response in a pedestrian accident. The test procedure is examined for a variety of representative cars of the European fleet. The investigation is purely based on numerical simulations carried out using the regulatory lower leg impactor, as described in the Directive 2004/90/EC, and compared to simulations where the impactor is replaced by the human FE model THUMS (Total HUman Model for Safety). THUMS was developed in collaboration by Toyota Motor Corporation and Toyota Central R&D Labs.. It has been extensively validated under pedestrian impact conditions and its response has been proven to be biofidelic. Therefore, THUMS results are considered as reference for the analysis of the lower leg impactor simulation results.

THUMS and impactor responses are compared looking at:

- Their kinematics, accelerations, velocities and deflections,
- Their injury prediction,
- And finally, their contact forces with the vehicle structures.

This work was carried out in the frame of the sub-project Nº3 on "Pedestrians and Cyclist accidents" of the European funded project "Advanced PROtection SYStem", APROSYS.

Keywords: Pedestrians, Impactors, Regulations, Human Body FEM, THUMS.

THE RECENT EPIDEMIOLOGIC study conducted by APROSYS project has shown a 5.1% annual decrease of pedestrian fatalities in Europe from 1993 to 2001. However, many pedestrians still suffer from serious injuries: on average for UK, Sweden, Germany and Spain, 24% of pedestrian injuries were serious ones. Lower extremities are the most frequently and the second most severely injured after the head.

The current study focuses mainly on the evaluation of the lower leg impactor used in the European Directive. Many studies (Takahashi et al., Yasuki et al., EEVC WG 17) have already addressed this issue and the FLEX-PLI, developed by the Japan Automobile Research Institute (JARI), is one of the possible answers to current impactor biofidelity limitations.

In addition to these previous studies, the current one presents an evaluation based on a large range of cars using real car geometries and stiffnesses. In the near future, this study will be extended by testing proposed modifications of current impactor to the same car range.

METHODOLOGY

Numerical simulations of a pedestrian impact and of a lower leg impact, as defined by the European Directive 2003/102/EC, were performed under LS-DYNA code. Simulations were replicated for six car models from super mini cars to large off-road vehicles. In order to take into account the wide range of the European fleet, vehicle representative of the different segments were selected for this analysis. Main geometrical characteristics related to pedestrian impacts are summarised in Table 1. The front-end shapes of these vehicles are presented in Figure 1.

Table 1 – Vehicles selected in the different segments

Vehicle category	LBR /ground (mm)	UBR/ground (mm)	BLE/ground (mm)	BL/ground (mm)	MBH/ground (mm)
Super Mini Cars (SMC)	259	577	740	125	418
Small Family Cars (SFC)	253	554	755	105	404
Large Family Cars (LFC)	224	490	744	139	357
MultiPurpose Vehicles (MPV)	255	605	781	116	430
Small SUV (SSUV)	393	663	904	149	528
Large SUV (LSUV)	537	774	1104	150	656

(LBR: Lower Bumper Reference, UBR: Upper Bumper Reference, BLE: Bonnet Leading Edge, MBH: Middle Bumper Height calculated as $(LBR + UBR)/2$).

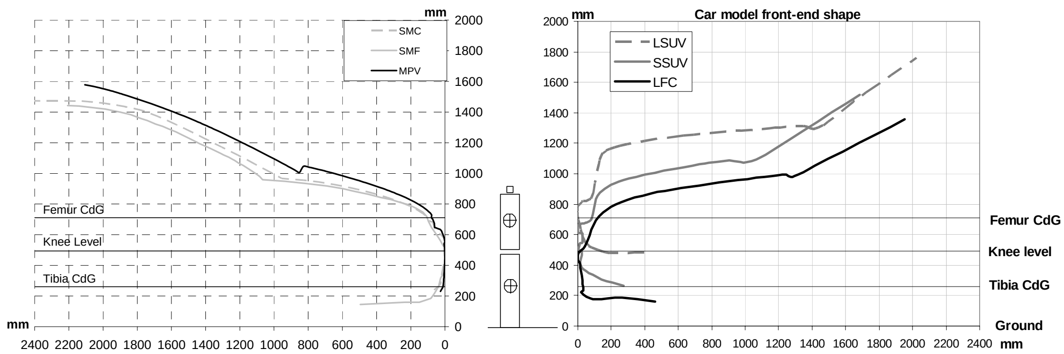


Figure 1 – Vehicle front-end shapes compared to impactor knee height

Two different approaches were followed for the development of these vehicle models:

- The SMC, SFC and MPV are multibody facet models created from 3D scan of cars and whose detailed mapping of stiffness was obtained from the analysis of Euro NCAP pedestrian tests (Martinez, 2007b),
- The LFC, SSUV and LSUV are FE models created referring to CAD geometry data.

FE models replicate the bumper and bonnet underlying parts geometry and stiffness but they require higher CPU time than facet models and can not be easily modified. The multibody facets car models represent the car outer geometry accurately and allow, as the FE models, a detailed interaction with the impactors and THUMS models. Multibody facets car models only replicate the general stiffness distribution over the car front-end, but adjustments of car front-end stiffness can be performed easily.

FACET VEHICLE MODELS were obtained through a 3D scanning, post processed in CAD and converted into a mesh. Their local stiffnesses were mapped using force-deflection contact characteristics according to a technique developed by Martinez (2007b). Martinez demonstrated that the Euro NCAP area colour had a stronger correlation with the car stiffness than the car segment. “Green”, “Yellow” and “Red” stiffness corridors were defined based on 425 Euro NCAP tests for each car area (bumper, BLE, child and adult head impact zones).

To implement these contact characteristics along the front end of the car, the rating map published by Euro NCAP was used to allocate a contact characteristic per Euro NCAP testing zone. Two approaches were analysed in order to select the stiffness to be used within this study.

The first approach would map the stiffness of the front end selected geometries with the average stiffness maps of the different car segments. These averages were calculated in Martinez, (2007a) taken into account all vehicles tested for pedestrian protection by Euro NCAP. These averages are not representative of current pedestrian protection offered by modern cars, since they are penalised by the high proportion of vehicle tested before Phase I came into force in 2005.

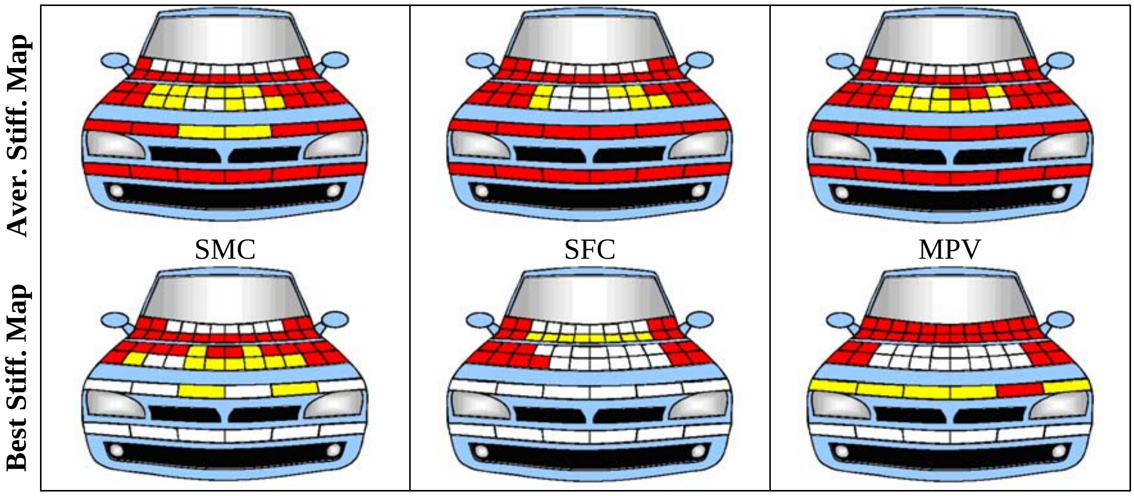


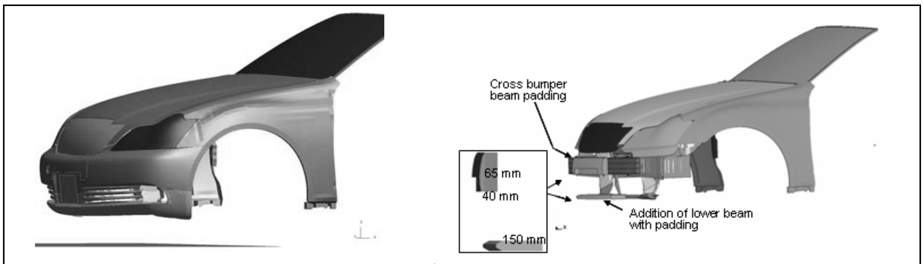
Figure 2 – EuroNCAP vehicle ratings for pedestrians

The second approach would map the stiffnesses of the front-end selected geometries with the best map of the different car segments (Bovenkerk, 2007). The different stiffness maps obtained from both approaches are presented in Figure 2 highlighting the improvement achieved nowadays in pedestrian protection. The second approach was preferred as it took into account the current improvements in pedestrian protection. Any conclusions would then be obtained considering pedestrian friendly cars. Therefore, each car segment vehicle model was developed with the vehicle stiffness map and stiffness functions of the most pedestrian friendly vehicle of its segment. The bumpers for the SMC, SFC and MPV would obtain the "Green" Euro NCAP score considering only the tibia acceleration. The BLE part would obtain a "Green" score for the SFC, a "Green" to "Yellow" score for the SMC and a "Yellow" score for the MPV.

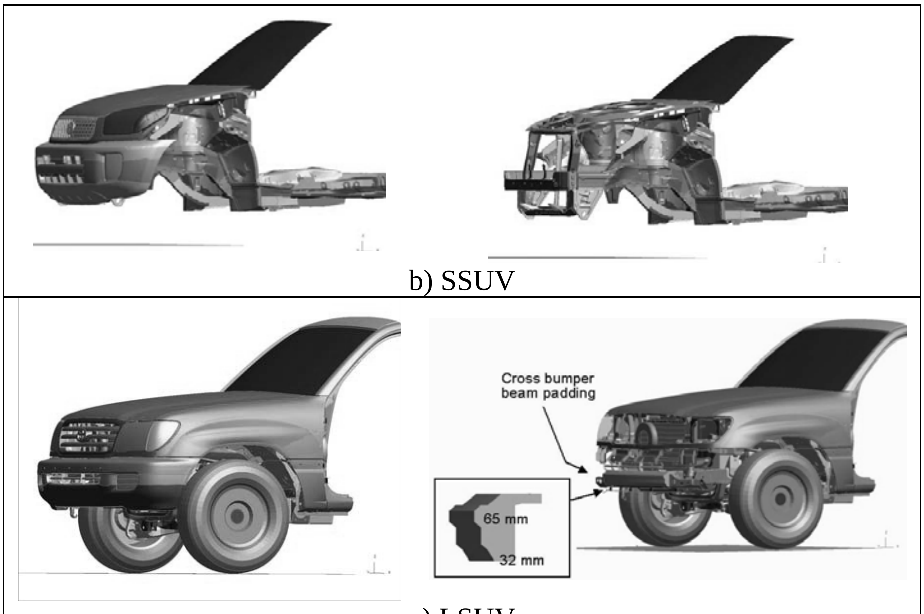
FE VEHICLE MODELS were created referring to CAD geometry data. Different detail levels were used for modelling these cars, but in all of them, major bumper and bonnet leading edge underneath structures were represented.

Padding on the cross bumper beam was added to the LFC and the LSUV to improve their performances in pedestrian impact. Except for the padding, the stiffnesses of the FE model structures are those of the real vehicles.

The vehicle models are shown in Figure 3. The LFC model and the LSUV models were validated using pedestrian impactor tests (without addition of cross bumper beam padding) (Yasuki 2006a, 2006b). The inclusion of these paddings made LFC bumper model become "Yellow" and the LSUV model "Green" compared with the APROSYS stiffness corridors defined for bumper area. It should be noted that the colour rating was attributed only considering tibia acceleration ("Red" score for tibia acceleration above 200 g, "Yellow" score for tibia acceleration in between 150 and 200 g and "Green" score for tibia acceleration below 150 g, Martinez et al. 2007b).



a) LFC



b) SSUV

c) LSUV

Figure 3 – Toyota FE vehicle models. a) LFC, b) SSUV, c) LSUV)

The THUMS (Total Human Model for Safety) was developed in collaboration by Toyota Motor Corporation and Toyota Central R&D Labs.. It reproduces anatomical geometry data and biomechanical properties of the human body of a 50th percentile male (height of 175 cm and weighting 77 kg) in a walking posture. It is made of approximately 60,000 nodes and 80,000 elements.

THUMS lower leg dimensions are in accordance with anthropometric studies such as the one performed by UVA (Crandall, 1996). In this study, tibia heights were collected on 39 male cadavers of 50th percentile mass (77.5 ±18.3 kg cm) and stature (175 ±7.6 cm) (Table 2).

THUMS tibia height (459 mm) is within the range of the average tibia height measured by Crandall et al. (470 ±40.7 mm). On straightened legs, with shoes, THUMS knee height is almost the same than the impactor (498 mm for THUMS, versus 494 mm for the impactor).

Therefore, in the case of a pedestrian impact, impact points on THUMS and on the impactor with respect to their knee are comparable.

Table 2 – Comparison between human and THUMS tibia and ankle dimensions

		Human 50th male (Crandall, 1996)	THUMS	Impactor
F	Medial malleolus height from head to floor Medial malleolus height from tip to floor	83 ±11 mm 83 ±13 mm	90 mm	
G	Lateral malleolus height from head to floor Lateral malleolus height from tip to floor	72 ±2.9 mm 69 ±12 mm	69 mm	
O	Tibial height from distal heel to tibial medial margin	470 ±40.7 mm	459 mm	494 mm

The lower limbs of THUMS model include the cortical and spongy parts of the bones (femur, tibia fibula and patella), as well as the meniscus, the knee ligaments, the ankle ligaments and the Achilles tendon. THUMS model response was extensively validated at different levels: bone level, lower extremity level and full-scale pedestrian impacts (Maeno 2001, Iwamoto 2002, Omori 2002, Yasuki 2006, Nagasaka 2003).

Material properties of bone are based on the study of Yamada (1970). Long bones were validated in quasi-static three points bending tests as carried out by Yamada. Figure 4 shows that simulations were in good agreement with the tests.

THUMS knee was validated under pedestrian impacts using various sources of experiments (Kajzer (1993, 1997, 1999), Levine (1984), Ramet (1995)). Validation of knee joint response under bending load is presented in Figure 5. A comparison of THUMS and lower leg impactor knee bending response is shown in Figure 7. The impactor knee records much higher bending moment than THUMS and cadavers.

Full THUMS kinematics was also validated comparing with pedestrian cadaveric tests performed by Schroeder et al. (2000) and Ishikawa et al. (1993) (Figure 6).

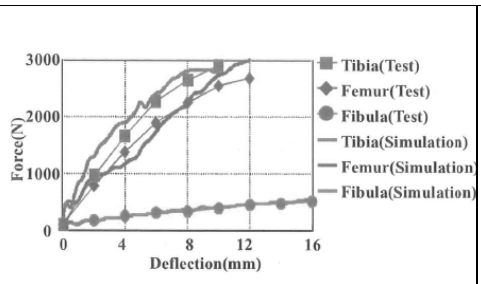


Figure 4 – Bone property validation according to Yamada's quasi-static bending tests

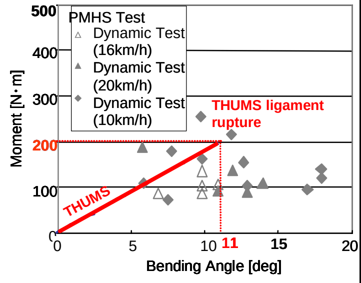


Figure 5 – Validation of knee response in shearing and bending (Yasuki, 2005)

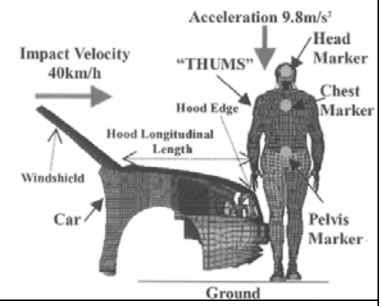


Figure 6 – Full body kinematics validation (Maeno, 2001)

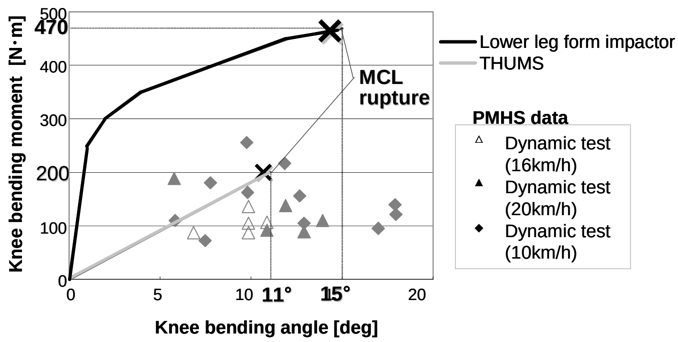


Figure 7 – Comparison of impactor and THUMS knee bending characteristics (Yasuki, 2005)

Failure criteria for the bones and the ligaments are summarised in Table 3.

Table 3 – Failure criteria for THUMS lower extremities

Part	Failure criteria
Cortical bone	Failure based on plastic strain (3%) (Failure for a total elongation of > 3%).
Ligament	Failure based on von Misses stress or plastic strain: 2% plastic strain (failure for a total elongation of around 15%).
Trabecular bone	No failure

Additionally, in order to compare with lower leg impactor knee injury criteria (bending angle and shearing displacement), "D" and "θ" values were measured on THUMS using the nodes as shown in Figure 8.

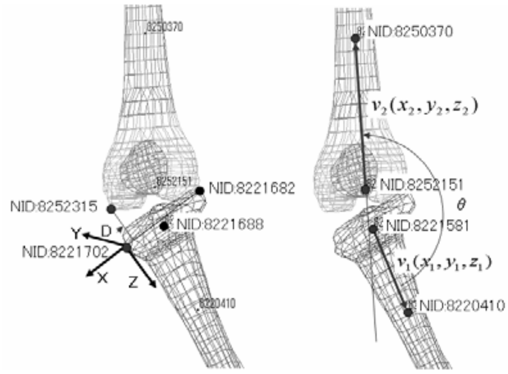


Figure 8 – Shear displacement and bending angle measured on THUMS

THE SIMULATION MATRIX consists of twelve different scenarios (two impact simulations per vehicle):

1) Lower leg impacts (according to Directive 2003/102/EC): the car is standing still and the lower leg impactor is propelled at an initial velocity of 40 km/h into the bumper of the car. The centre line of the impactor is aligned with the car centre line and the bottom of the impactor is set at ground level to keep the same impact height with respect to the knee joint as in THUMS model,
2) THUMS impacts: THUMS model is standing still in front of the vehicle with its left impacted leg stretched and the other one slightly bent and positioned backwards in a walking posture. The vehicle is moved at an initial velocity of 40 km/h without any braking applied.

RESULTS

For all the simulation cases the following set of results were obtained:

- A kinematics of the impact in both scenarios, making a detailed analysis of the THUMS leg and the impactor through the tracking of several nodes along the THUMS leg during the impact (graphs of Figure 10). In these graphs, a point defined as knee centre in the impactor and THUMS were made coincident and used as reference (zero height). The global displacement of the knee was subtracted in order to highlight the relative bending between femur and tibia. The vehicle profile was also superimposed to visualise impact point on the car with respect to the knee. From these graphs it is of course not possible to evaluate the global vertical movement of THUMS and the impactor.
- An injury prediction based on Directive 2003/102/EC using knee shearing displacement, knee bending angle and tibia acceleration with their associated injury thresholds. These three parameters were also assessed in THUMS, following the definition of these parameters as mentioned in Figure 8. Table 4 summarises the peak values or the value reached at failure for these parameters. In the case of ligament or bone fracture, THUMS and lower leg impactor parameters could not be compared anymore since the impactor was not designed to break.
- A strain distribution in the bones and ligaments, highlighting when plastic strains are reached.

Table 4 – **Summary of the injury criteria values**

Vehicle type	Criteria	Impactor	2003/102/EC thresholds (Phase I – Phase II)	THUMS
SMC	Knee shearing displacement	1.39 mm	6 mm-6 mm	*5 mm
	Knee bending angle	25º	21º-15º	*9º
	Tibia acceleration	121 g	200 g-150 g	222 g
SFC	Knee shearing displacement	1.5 mm	6 mm-6 mm	*4.8 mm
	Knee bending angle	21º	21º-15º	*9º
	Tibia acceleration	99 g	200 g-150 g	175 g

(Table 4 continuation)

LFC	Knee shearing displacement	2 mm	6 mm-6 mm	*5.8 mm
	Knee bending angle	14.6º	21º-15º	*19º
	Tibia acceleration	127 g	200 g-150 g	155 g
MPV	Knee shearing displacement	1.45 mm	6 mm-6 mm	*1.75mm
	Knee bending angle	20º	21º-15º	*10º
	Tibia acceleration	100 g	200 g-150 g	165 g
SSUV	Knee shearing displacement	5 mm	6 mm-6 mm	*0.4 mm
	Knee bending angle	38º	21º-15º	*12º
	Tibia acceleration	180 g	200 g-150 g	195 g
LSUV	Knee shearing displacement	4.5 mm	6 mm-6 mm	*3 mm
	Knee bending angle	22º	21º-15º	*26º
	Tibia acceleration	132 g	200 g-150 g	331 g

*: Value of the injury criterion when the first ligament breaks, Maximum knee shearing displacement in absolute value.

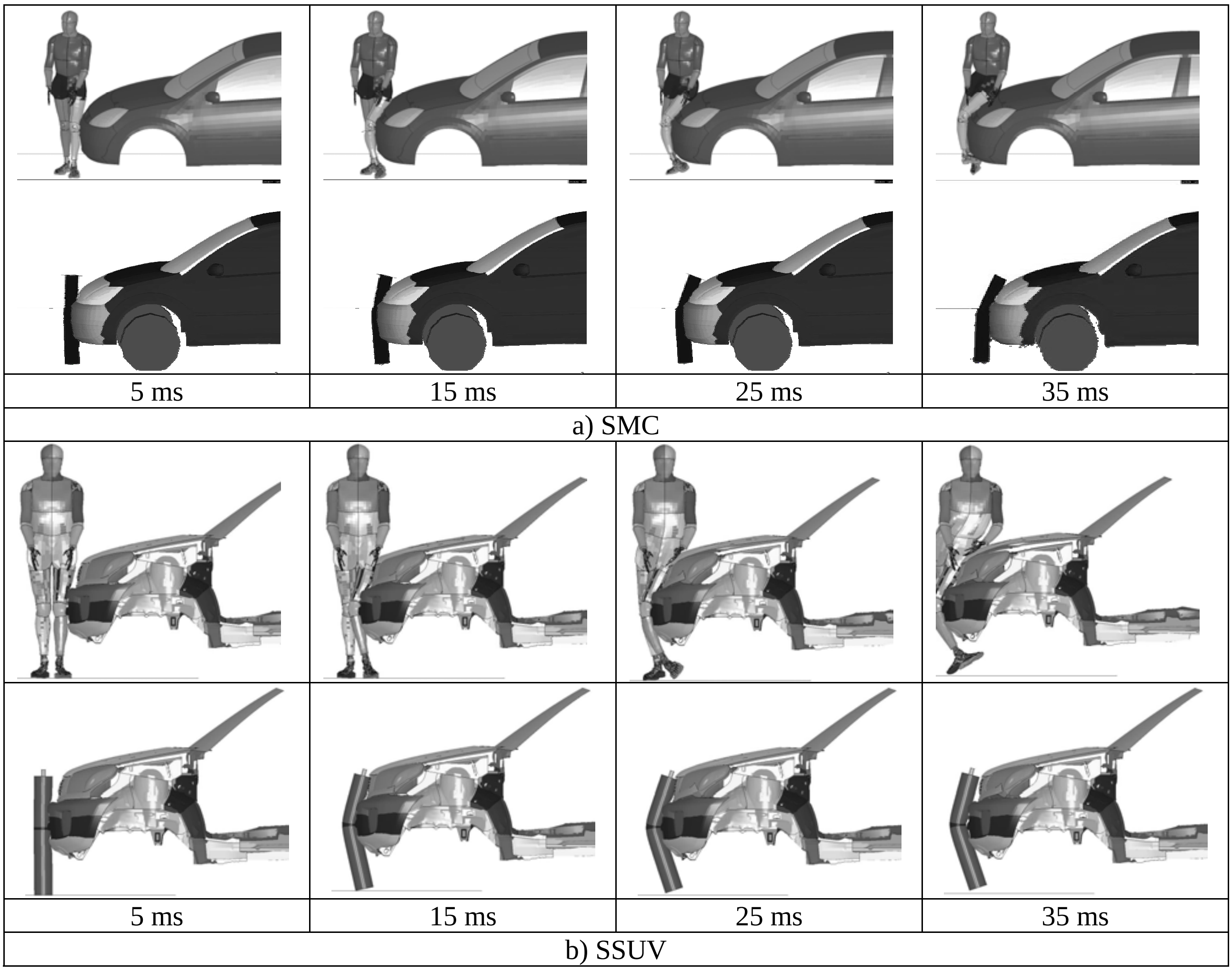


Figure 9 – **Kinematics of the impact in two selected cases. a) SMC, b) SSUV).**

DISCUSSION[1]

KINEMATIC ANALYSIS. The superimposition of the kinematics of THUMS impacted leg and of the impactor in the case of the six vehicle types is presented in Figure 10.

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[1] Leg means the lower extremity from hip to ankle.

In general, THUMS lateral bending is greater than that of the impactor. This can be explained by the higher flexibility of THUMS with respect to the impactor (impactor “bones” are rigid and impactor knee is stiffer than THUMS’s one (Figure 7)) but also by THUMS upper body mass which keeps the leg in contact to the car front-end whereas the impactor escapes more easily (LFC and LSUV cases). In addition, THUMS higher flexibility is given by its ability to simulate bone fracture and ligament rupture.

From the six kinematics obtained for the studied vehicle range, it can be seen that the impactor kinematics varies from THUMS kinematics when the MBH moves away from the THUMS and impactor knee level. In the case of the LFC, which has the lowest mid bumper height (357 mm), the impactor tibia moves upwards (clockwise in the graph) and starts to escape from the car front-end after 15 ms. For the SFC (403.5 mm) and SMC (418 mm), which have the next MBH in increasing order but at a height still under the knee level, the impactor tibia also moves upwards, as in the case of the LFC, but during the rebound phase. In the first 15 ms and 20 ms for the SFC and the SMC respectively, the tibia of the impactor moves first downwards underneath the car. This latter movement can also be observed for THUMS.

For the MPV, of 430 mm MBH, only the motion towards the car is seen (no rebound upwards). In the cases where the MBH is higher than the knee height (SSUV: 528 mm, LSUV: 656 mm), the impactor tibia has a large motion underneath the bumper, similarly to THUMS lower leg. For the SSUV and the LSUV, the kinematics of the impactor and THUMS differs in the femur area.

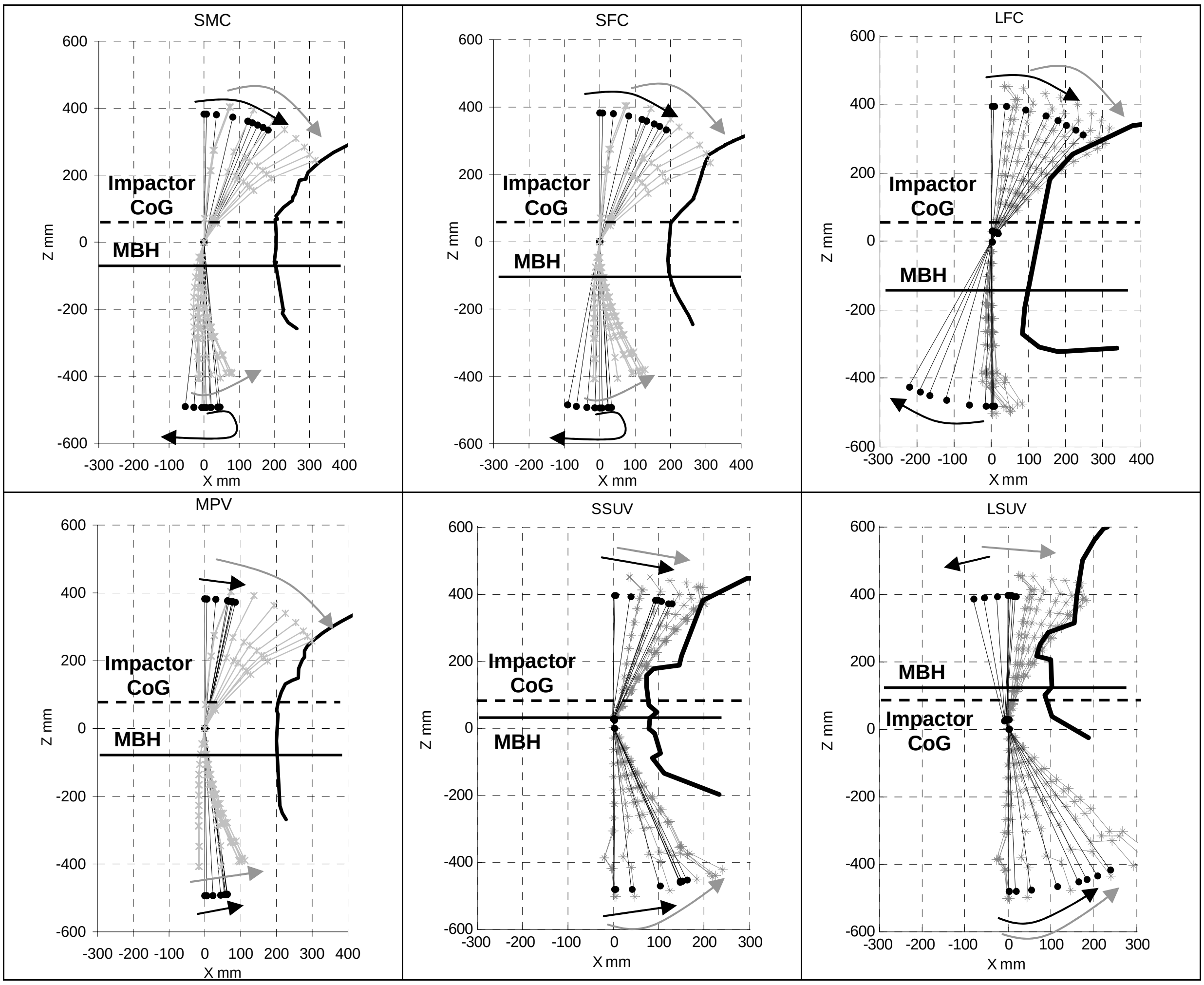


Figure 10 – Kinematic analysis of THUMS impacted leg and lower leg impactor for the different car models. From 0 to 40 ms with a time step of 5 ms (in gray: THUMS, in black: Legform impactor, in bold black: car front-end).

Opposite kinematics are observed for the impactor and THUMS in the case of the LSUV. The femur of the impactor moves anti-clockwise and losses rapidly contact with the car front. As the center of gravity of the impactor is 550 mm above the ground, the MBH hits the legform above it whereas the center of gravity of THUMS is still higher than the MBH.

Inertial effect of upper body mass of THUMS keeps for a longer time its leg in contact with the car front. The flexibility of THUMS bones reinforces this tendency whereas the rigid legform, where no inertial load is applied from an upper body mass, can easily moves away from the car. The difference of flexibility in THUMS and impactor increases the difference in their tibia kinematics in the case of lower bumper (LFC, SMC and SFC). Due to the bumper lead and THUMS upper body mass, the femur moves towards the bonnet pulled by the pelvis when it leans onto the bonnet.

For bumper impacts, the MBH could be considered as a good approximation of the load concentration applied by the car on the pedestrian lower extremity. The impactor kinematics depends mainly on the distance between its centre of gravity and the MBH.

INJURY CRITERIA. Impactor and THUMS injury assessments were conducted for all car models through measuring the knee bending angle, the knee shearing displacement and the tibia acceleration. Additionally, the femur acceleration was measured in THUMS model for comparison with the tibia acceleration.

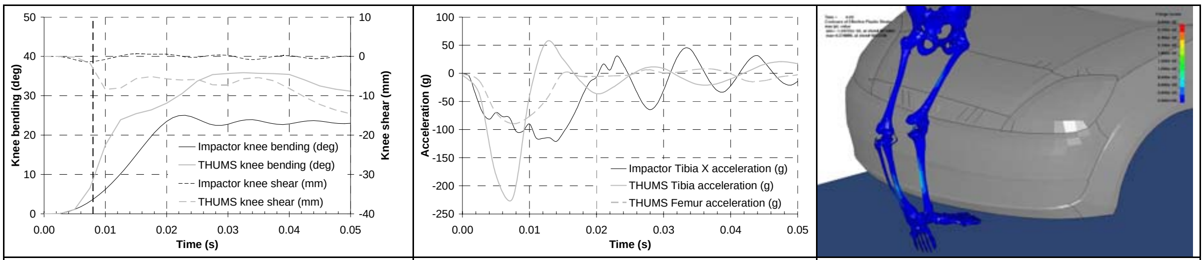
SMC case (Figure 11):

The results of the knee measurements for the SMC are shown in Figure 11a. The knee bending angle results suggest that both bending and shearing are rather higher in the THUMS model than in the sub-system impactor. The impactor was developed with rigid components. Despite of the lack of flexibility, it can still simulate the tendency of lower extremity responses in a gross point of view until injury occurrence.

Different ligaments failed in THUMS simulations. The first one to fail was the medial collateral ligament (MCL), which fails at 9 ms as a consequence of the direct impact with the car. Later, also the anterior cruciate ligament failed as they are not able to withstand the load needed to keep the tibia and femur condyles together without the MCL. When the first ligament rupture occurs in the model, THUMS knee bending angle was 9º, whereas the impactor knee angle was less than half of this value at the same moment. However, the maximum value obtained in the impactor (25º) although it was later in time (22 ms), does suggest that a knee injury may appear according to the limits of the Directive 2003/102 Phase I (21º) and Phase II (15º).

Regarding the accelerations measured in the THUMS leg (Figure 11b), it can be seen that tibia acceleration almost doubles the femur acceleration with a peak value above 200 g. The acceleration measured in the impactor only reaches a peak value of 121 g, being below the limits from the Directive phase I (200 g) and phase II (150 g). The AIS2+ tibia injury risk is estimated as 10-30% for this case, based on the injury risk curve defined by EEVC WG17 (1998).

In the THUMS leg, although the acceleration is higher, there is no tibia bone fracture. In this case, the impactor does predict correctly the absence of tibia fracture. The stress distribution in THUMS leg shows areas where the bones do reach their yield stress and therefore, plastic strain arises, but without reaching plastic strain failure limits implemented in the model. The stress distribution reaches similar maximum values in the femur and the tibia, whereas no stress concentration is observed in pelvic region. In the SMC case, all the loads are transferred to the leg, the pelvis just slides over the bonnet with low loads.



a) Knee bending angle and shearing displacement	b) Leg accelerations	c) THUMS plastic strain distribution in lower extremities

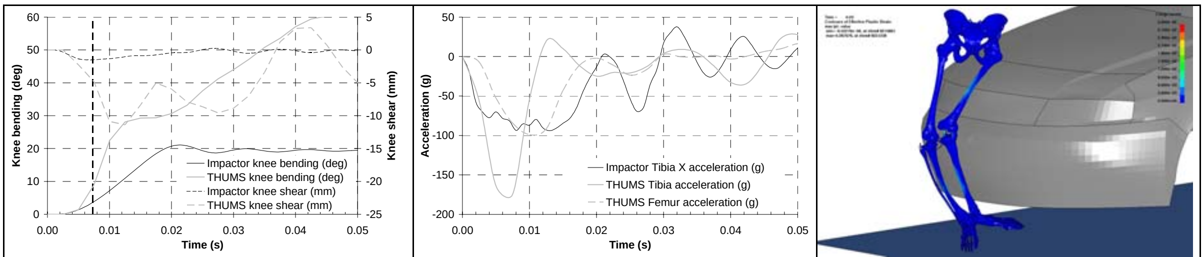
Figure 11– Injury criteria for the SMC

<u>SFC case (</u>Figure 12<u>):</u>

Due to the geometry of the SFC bumper, the force application is concentred on the knee region. In this case, THUMS tibia acceleration just reaches 175 g peak acceleration and the femur stays close to 100 g. In this case, the tibia acceleration of the impactor does not exceed the limits (99 g) from the Directive Phase I (200 g) and Phase II (150g). The AIS2+ tibia injury risk is estimated as 5-24% for this case, based on the injury curve defined by EEVC WG17 (1998).

Regarding knee injuries, it can be seen that the impactor predicts injuries because the maximum bending angle in the knee is 21º, which equals to the maximum value allowed in Directive Phase I (2003/102/EC) but exceeds the value of Phase II.

Injuries are indeed found in THUMS simulations too. The MCL ligament fails at the beginning of the impact (7 ms). At this time, THUMS knee bending is equal to 9º, whereas the knee bending angle of the impactor is just equal to 4º. THUMS knee bending increases continuously along the whole impact while the one for the impactor reaches and maintain its maximum at 20 ms. Later, also the cruciate ligaments fail as they are not able to withstand the load needed to keep the tibia and femur condyles together without the MCL. After failure occurred in THUMS, comparison with the impactor becomes meaningless since the impactor is not designed to replicate failure. Therefore, its values do not further increase after reaching their maximum. The evaluation of the stress distribution in THUMS legs shows that the load distribution starts in the knee region and extends for both femur and tibia bones. In this process, there are areas where the bones do reach plastic strains but still stay below the plastic strain limits. The load transferred to the pelvis remains low.

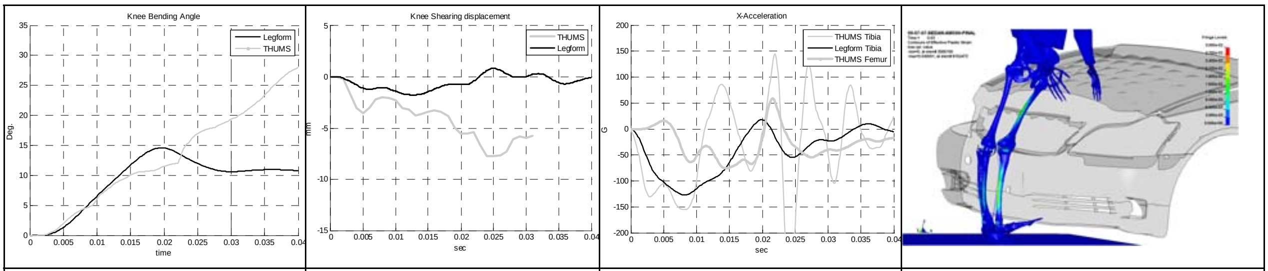


a) Knee bending angle and shearing displacement	b) Leg accelerations	c) THUMS plastic strain distribution in lower extremities

Figure 12 – Injury criteria for the SFC

<u>LFC case (</u>Figure 13<u>):</u>

In the case of the LFC, the impactor values are below the thresholds defined in the Directives 2003/102/EC Phase I and Phase II: below 15°, 6 mm and 150 g for knee bending angle, knee shearing displacement and tibia acceleration respectively. THUMS exceeds all thresholds and the increasing knee bending angle and knee shearing displacement suggest relatively higher risk of injuries to the knee. Looking into more details to the model, rupture of the MCL on the struck leg is observed at 30 ms. Fractures of the tibia plateau and femur distal condyle of both legs appear around 20 ms, which creates the peak on the tibia acceleration. The plastic strain value of 3% defined for cortical bone failure was reached in the tibia plateau and femur distal condyle areas. The elements in these zones were deleted during the calculation. The plastic strain values in the tibia and femur shafts remain below the failure limit (around 2% of plastic strain).



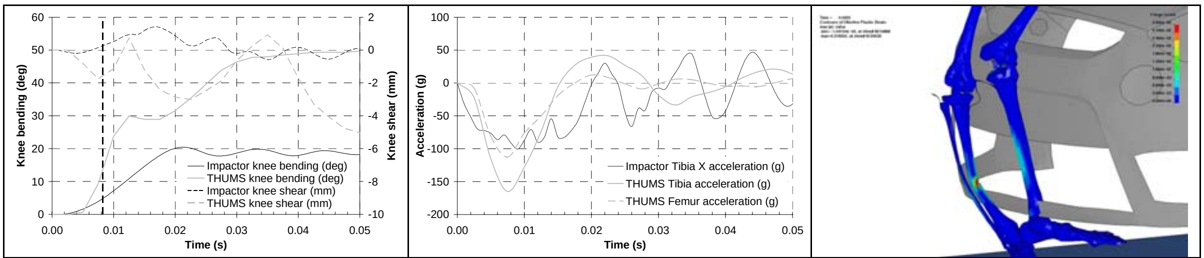
a) Knee bending angle	b) Knee shearing displacement	c) Leg accelerations	d) THUMS plastic strain distribution in lower extremities

Figure 13 – Injury criteria for the LFC

<u>MPV case (</u>Figure 14<u>):</u>

Both bending angle and shearing displacement are higher in THUMS model than in the sub-system impactor. In THUMS, the MCL fails at 9 ms and is followed a bit later by the failure of the cruciate ligaments. When the first ligament rupture occurs in THUMS, its knee bending angle is 12º. At the same time, the impactor knee angle is around 5°, indicating a higher stiffness of impactor knee compared to THUMS (Figure 14a). Considering the maximum value of the knee angle (20°), the impactor predicts a high risk of knee injury if the limit of the Directive Phase II (15º) is considered whereas opposite conclusion will be given if Directive Phase I (21°) is considered.

Similar tibia and femur accelerations (around 100 g) are measured in THUMS (Figure 14b). Both are well below the Directive Phase I (200 g) and Phase II (150 g) limits. No tibia bone fracture is seen in THUMS. For the impactor, the tibia acceleration maximum value (165 g) corresponds to a risk of AIS2+ injury of 5-24% (according to EEVC WG17, 1998). The impactor and THUMS both indicate the absence of risk of tibia fracture. Figure 14c shows the distribution of plastic strain. At the beginning of the impact, the load is evenly distributed between the tibia and the femur. In a later stage, the lower part of the tibia impacts the lower bumper cross beam and exceeds its yield stress, but without reaching the plastic strain failure limit implemented in the model. A bone fracture is observed on the non-struck leg. An acceleration peak of 200 g is recorded (not shown in the graph).

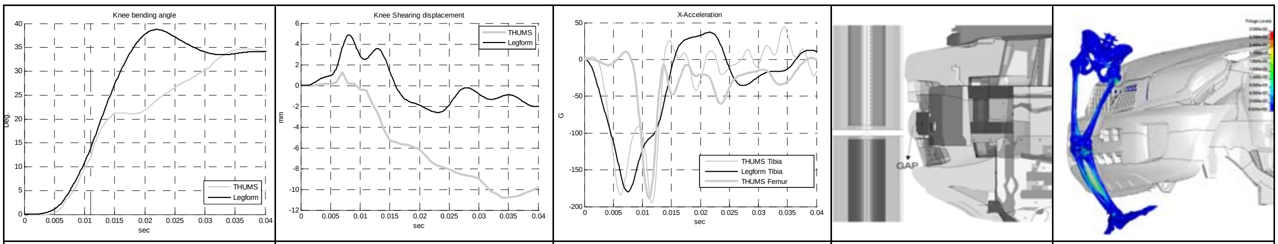


a) Knee bending angle and shearing displacement	b) Leg accelerations	c) THUMS plastic strain distribution in lower extremities

Figure 14 – Injury criteria for the MPV

<u>SSUV case (</u>Figure 15<u>):</u>

In the case of the SSUV, THUMS and impactor tibia X-acceleration values exceed the 150 g limit defined by the Pedestrian Directive phase II. Bone fractures are predicted at the tibia plateau, femur distal condyle, femur neck and pubic rami. In THUMS case, two peaks can be observed whereas only one is seen in impactor case. The first peak in THUMS case is seen when the flesh is compressed and the leg starts to move. The second peak is seen when the gap behind the bumper cover is compressed and the hard structures start to load the leg. This second event also corresponds to the single peak observed for the impactor. MCL fails in THUMS at 11 ms which explains the high value for knee bending angle. Failure criteria are reached at several areas: MCL, tibia plateau, femur distal condyle, femur neck, pubic rami. The impactor knee bending angle value is even higher (37°) because all the flexibility of the leg is concentrated in the knee are.

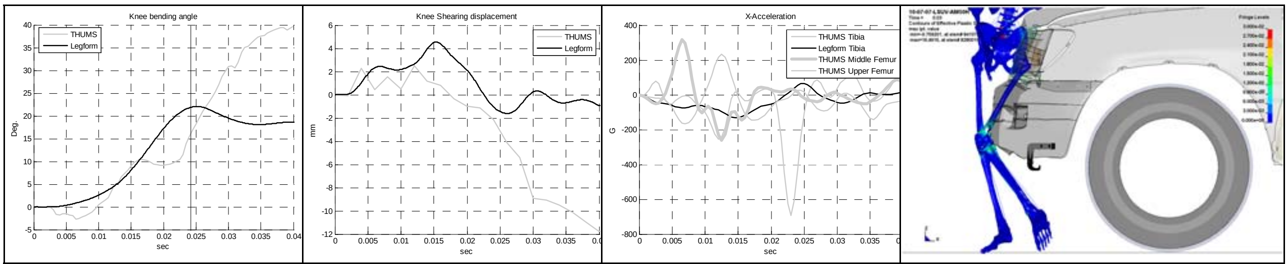


a) Knee bending angle	b) Knee shearing displacement	c) Leg accelerations	d) Underneath bumper structures	e) THUMS plastic strain distribution in lower extremities

Figure 15 – Injury criteria for the SSUV

<u>LSUV case (</u>Figure 16<u>):</u>

In the case of the LSUV, the impactor tibia X-acceleration was below 150 g whereas high peaks were recorded on THUMS certainly due to various bone fractures created during the impact and a dislocation of the knee joint. Knee bending angles reach 26° in THUMS before the MCL rupture and 20° in the impactor. The car front loads the femur and the pelvis, areas where no injury assessment is provided by the impactor. The shearing displacement is first positive as the contact happens first at the femur level. It becomes then negative and reaches 11 mm in THUMS case.



a) Knee bending angle	b) Knee shearing displacement	c) Leg accelerations	d) THUMS plastic strain distribution in lower extremities

Figure 16 – Injury criteria for the LSUV

CONTACT FORCES. Figure 17 shows the contact forces between the cars and the impacted THUMS leg or impactor in the cases of the LFC, the SSUV and the LSUV. Tibia contact forces, combined with knee ones, femur and pelvis contact forces are recorded. In all the cases, the impactor forces are higher and have shorter pulses than THUMS ones. The impactor is stiffer than THUMS and often losses contact with the car front-end earlier than THUMS due to its reduced flexibility and lack of upper body mass.

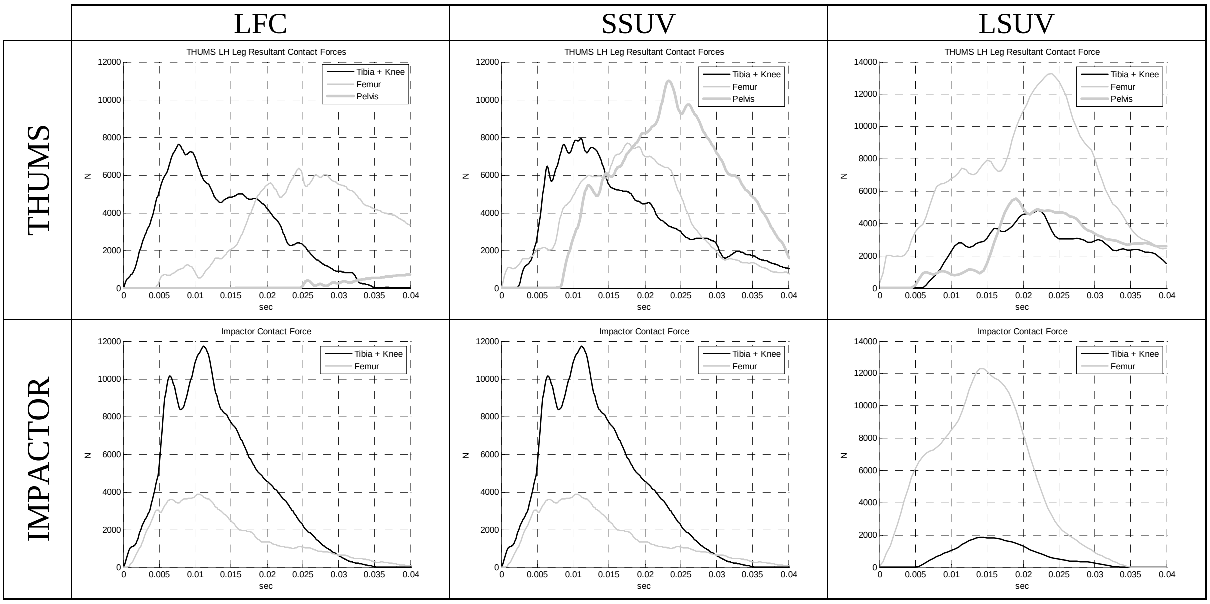


Figure 17 – Contact forces for THUMS and impactor for selected vehicles.

CONCLUSIONS

This study evaluates the response and capability of current pedestrian lower leg impactor to predict lower leg injuries. Similar simulations with the impactor and the human model THUMS are carried out in which THUMS response and injury prediction are considered as the reference. The analysis considers a wide range of cars covering the different categories found in the European fleet nowadays. Compared to previous studies, detailed geometry of car front-ends are considered here.

The results highlight the different kinematics of the impactor and THUMS. Lacks of flexibility and lack of upper body mass in the impactor are identified as the two major drawbacks. The differences in the kinematics increase as more difference exists between the impactor centre of gravity (CoG) and the mid bumper reference height. The kinematics differences appear when the impactor CoG is below or over the mid bumper reference height. These differences in height cause the impactor to lose contact with the car much earlier than it happens in THUMS either in the tibia or in the femur area.

To assess high bumper car front-end, the addition of an upper body mass might be necessary. Indeed, the mass plays a significant role in keeping the femur in contact with the car front. The impactor femur can more easily release from the car which may influence the value of the knee bending angle. With an added mass, it might be possible to increase the range of cars tested with the lower leg impactor (for LBR> 500 mm). Further validation will be needed, but this result is in accordance with previous study conducted by Takahashi et al. (2001).

In general, higher contact forces on the car structures are generated with the impactor due to its rigidity and the time of contact is often shorter than with THUMS. In the case of active pedestrian protection systems triggered by the contact with pedestrian, the impactor might appear as unsuitable.

For all the cases, injuries are observed in knee ligaments (mainly MCL) with the THUMS model. The impactor knee bending predicts some of them by using the 15º limit proposed in the Phase II of the Directive 2003/102/EC. It fails in predicting injuries for the LFC case. The knee shearing displacement can not predict any injury even when cruciate ligament failure is observed on THUMS. The tibia acceleration criterion, established to predict tibia fractures, is only reached in the SSUV case. The tibia accelerometer was aligned in this case with the lower bumper support creating this peak on the acceleration. For the other vehicles, the tibia acceleration is always below the 150 g limit.

Despite some limitations recognised in THUMS model, such as the need for refinement in the knee area to predict injury more accurately and the need to implement active muscle activation, THUMS has been proven as able to predict pedestrian global kinematics and main injuries. The lower leg impactor failed many times to predict injuries observed in THUMS. Its biofidelity is limited (no flexibility of tibia and femur parts, stiff knee behaviour in bending load) and the lack of an upper body mass prevents its use in the case of high bumper vehicles for which it will have incorrect kinematics and therefore an incorrect prediction of the knee bending angle.

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