

Development and Evaluation of an Upper Body Mass (UBM) for the Flexible Pedestrian Legform Impactor (FlexPLI) and for Incorporation within Improved Test and Assessment Procedures – Results from SENIORS

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Abstract Accounting for a stagnation of pedestrian fatality numbers alongside increasing injury severities, including lower extremities, the SENIORS project developed and evaluated a pedestrian torso surrogate for the lower legform impactor FlexPLI for incorporation in improved test and assessment procedures with focus on biofidelity and injury assessment ability. Based on APROSYS results, this next generation upper body mass (UBM) is attached via flexible rubber element to the FlexPLI, providing for time lag of the pedestrian's torso and hip joint rotation observed during vehicle-to-pedestrian impacts.

While human injury risk functions were analyzed, human body model (HBM) and FlexPLI-UBM simulations were correlated for a femur transfer function. FlexPLI vs. FlexPLI-UBM correlation studies contributed to knee and tibia assessment. Performance limits were proposed for revised Euro NCAP assessment procedures within the framework of a new rating scheme. Experimental tests served for evaluation.

The UBM significantly improves the impactor biofidelity and lower legform test and assessment procedures, including vehicles with distinctively angled outboard areas and high frontends. FlexPLI-UBM tests confirm good correlations with the HBM. For the first time, pedestrian lower extremity injury assessment is based on one unique test, replacing the lower (or upper) legform to bumper test as well as the test with the upper legform impactor to wrap around distance line 775.

Keywords injury mitigation, lower extremity Injuries, pedestrian component testing, test and assessment procedures, vulnerable road user protection.

I. INTRODUCTION

Because society is ageing demographically and obesity is becoming more prevalent, the EU-funded Safety ENhancing Innovations for Older Road users (SENIORS) project aimed to improve the safe mobility of the elderly, and overweight/obese persons, using an integrated approach that covers the main modes of transport as well as the specific requirements of this vulnerable road user (VRU) group.

Despite continued improvements to the passive vehicle safety of passenger cars, recent collision data showed a stagnation of pedestrian fatality numbers on European roads [2], facing the risk of not meeting the European Union goal of halving the number of road fatalities by the year 2020.

Based on collision data from the German In-Depth Accident Study (GIDAS) and the Swedish Traffic Accident Data Acquisition (STRADA) as reported in [3], the percentages of AIS1, AIS2, and AIS3+ injuries to the different body regions of the age groups 25-64 and 65+ are displayed in Fig. A-1 (Appendix A) exemplarily for pedestrians in collisions with passenger cars. Both databases agree regarding the most affected body regions for pedestrians and cyclists and their particular relevance for the elderly. In terms of pedestrian injury levels, head, thorax, pelvis and lower extremities are the most relevant body regions with the highest proportions of AIS2 and AIS3+ injuries, with the elderly sustaining severe injuries more frequently than younger pedestrians.

Test and assessment procedures for passive pedestrian protection which are based on developments by the European Enhanced Vehicle-safety Committee (EEVC) have been introduced for vehicle type approval more than a decade ago and harmonised to a large extent between world-wide regulations such as United Nations Global Technical Regulation No. 9 (UN-GTR9) [4] and consumer programmes like the European New Car Assessment Programme (Euro NCAP) [5][6]. Test tools and procedures significantly contributed to an improved

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safety of VRUs and mitigation of injuries during vehicle to pedestrian impacts; however, they also revealed clear limitations regarding their applicability. The flexible pedestrian legform impactor (FlexPLI), being the state-of-the-art tool for assessing lower extremity injuries, is equipped with strain gauges in the femur and tibia section and string potentiometers in the knee area. While performing well in the knee and tibia area, it lacks biofidelity in the femur, especially due to extraordinary high rebound during the assessment phase. Thus, a realistic assessment of femur injuries is not possible. Furthermore, data acquisition during impacts against angled surfaces leads to sometimes less meaningful results, mainly due to unrealistic high impactor rotation about the long axis of the legform [7]. Finally, the absence of the pedestrian upper body precludes the assessment of injury risks caused by high bumper and bonnet leading edge (BLE) vehicles [8]. These shortcomings were addressed by the development of an upper body mass (UBM) for the lower legform impactor alongside the modification of the test and assessment procedures.

The European research project SENIORS developed modified pedestrian test and assessment procedures and impactors with the aim to mitigate the injury severity of the most affected body regions of VRU towards a further improvement of passive pedestrian safety. Current pedestrian impactors were analysed regarding their ability to address recent collision scenarios and any remaining open gaps were closed with new impactor concepts. Based on results from APROSYS [1], this paper describes the steps undertaken towards the development of a pedestrian torso mass and its application to the lower legform impactor FlexPLI. The principal goals were to entirely assess lower extremity injury risks to the femur, the knee and the tibia, induced by passenger car frontends and to improve the current legform's injury assessment ability at angled surfaces as well as for vehicles with extremely high bumpers and leading edges such as SUVs and flat front vehicles (FFV).

II. METHODS

Starting point was an upper body mass based on impactor and HBM FE simulations, reflecting a variety of load cases as combination of vehicle type, impact height, mass weight, mass position and impactor angle [1]. This first generation FlexPLI-UBM showed good correlation to HBM results as well as good performance and durability on the FlexPLI leg parts. However, it was rigidly attached to the upper end of the femur part of the FlexPLI and therefore neither accounted for the time lag seen during an impact with the human torso before loading the leg nor for the damping effect like in the flesh of a human or the hip rotation as observed during a human full body impact. These effects were addressed by a next generation UBM with relatively short rubber element attachment and a rotation point forward to the rubber part centreline. The moderate additional mass of 7kg and low CoG were mainly chosen in line with the APROSYS results on the one hand and to maintain durability and technical feasibility regarding the application to existing launcher systems on the other hand.

The overall methodology for the development of the FlexPLI-UBM is depicted in Fig. B-1 (Appendix B). Paired human body model and impactor simulations were performed against generic, i.e. universal vehicle frontends representing different vehicle categories, and actual vehicle frontends. Several loops including variations of impact parameters and UBM specifications gave essential input to impactor fine tuning and resulted in transfer functions for implementation within updated assessment procedures with modified threshold values. Validation tests against generic and actual vehicles under the modified test procedures concluded the project.

FE Simulations

Paired simulations with version 4 of the Human Body Model (HBM) Total HUMAN Model for Safety (THUMSv4) and impactor models of the FlexPLI with and without applied UBM, were performed against the generic vehicle Buck of the Society of Automotive Engineers (SAE Buck) and two derivatives, SUV and Van/MPV [9], as well as against four actual vehicle models, representing the categories Sedan, SUV and Van/MPV. The aim was to investigate correlations that could be used for the subsequent establishment of lower extremity injury criteria. A revision and fine tuning of the UBM, its attachment to the upper femur and the test parameters contributed to improved correlations in terms of kinematics as well as impactor and HBM loadings.

An example of the THUMSv4 simulation setup is displayed in Fig. C-1 for the SAE Buck. THUMSv4 was positioned with the struck side leg vertically to the ground while the other leg was inclined at 20 degrees in forward motion, as defined in [10]. This setup was intended to replicate the vertical FlexPLI position and to avoid interaction between the two legs. With regard to the impact height, to account for the thickness of the shoe sole, THUMSv4 was positioned with the struck side leg 25mm above ground level.

Subsequent to THUMSv4 simulations, a variety of FE simulations with the original FlexPLI (FlexPLI Baseline) and equipped with both investigated UBMs were carried out against four actual vehicle models, the SAE Buck and two derivatives. An overview of all simulations including ambient conditions is illustrated in TABLE I.

TABLE I
OVERVIEW OF HBM AND IMPACTOR SIMULATIONS

Surrogate/ Vehicle	THUMSv4			FlexPLI Baseline			FlexPLI-UBM _{rigid} (PK1.1)			FlexPLI-UBM _{rubber} (WS)		
Impact height [mm] for	-EoB	y0	EoB	-EoB	y0	EoB	-EoB	y0	EoB	-EoB	y0	EoB
Actual Van/MPV	0	0	0	75	75	75	25	25	25	25	25	25
Actual SUV	0	0	0	75	75	75	72	50	72	72	50	72
Actual Sedan I	0	0	0	75	75	75	25	25	25	25	25	25
Actual Sedan II	0	0	0	75	75	75	25	25	25	25	25	25
SAE Buck (Sedan)		0			75			25			25	
SAE Buck (SUV)		0			75			40			40	
SAE Buck (Van/MPV)		0			75			25			25	

FE impactor simulations were performed with the FlexPLI Baseline, with applied UBM with rigid connection (FlexPLI-UBM_{rigid}) [1] and a protection kit (PK1.1) for avoiding unintended interaction between mass and vehicle frontend, and with applied UBM via flexible connection (FlexPLI-UBM_{rubber}), consisting of rubber material identical to that used for the neck of the WorldSID dummy (WS) [11]. All Tests against actual vehicle models were aimed at vehicle centreline (y0) and at either end of the bumper beam (left hand side (LHS): -EoB, right hand side (RHS): EoB). Due to its two-dimensional front, tests against the SAE Buck were performed at y0, only.

Considering the sometimes extraordinary rotation of the FlexPLI in particular when tested against angled surfaces, perpendicular impactor testing alongside vehicle rotation around its z-axis was further investigated. Since the current test area is limited by either the EoB or the corners of bumper, as defined by the contact points between a vertical plane [5] or corner gauges [8] making an angle of 60 degrees to the vertical longitudinal vehicle centreplane, the vehicle was rotated by a yaw angle of 30 degrees for the impact, as illustrated in Fig. C-6 in Appendix C. With this setup, additional simulations were carried out against the actual Van/MPV representative at both EoBs, using both the FlexPLI-UBM_{rigid} and the FlexPLI-UBM_{rubber}.

While the HBM simulations were performed with THUMS always positioned at identical standing height, the FlexPLI Baseline impacted the vehicle frontend with the bottom of the legform at 75mm according to UN-R 127.01 [12] as well as Euro NCAP [5]. The FlexPLI-UBM_{rigid} and the FlexPLI-UBM_{rubber} followed considerations leading to the draft test procedures described in [13]: Based on anthropometric data and the geometries of the FlexPLI (see Fig. C-2 in Appendix C), the impact heights for the FlexPLI-UBM_{rubber} and the FlexPLI-UBM_{rigid} were defined by the bonnet leading edge (BLE) height as well as the height of the lower bumper reference line (LBRL). According to the anthropometric data from [14] (and THUMSv4 respectively), an assessment of femur injuries is recommended to be undertaken only for vehicles with a minimum height of the BLE exceeding 535mm (and 521mm respectively). Second, the height of the LBRL should not exceed 1,045mm (and 970mm resp.), which is always the case with all M1 passenger vehicles currently on the European market.

The FlexPLI-UBM impact height was finally derived from human anthropometry as well as the dimensions of the FlexPLI as illustrated in Fig. 1:

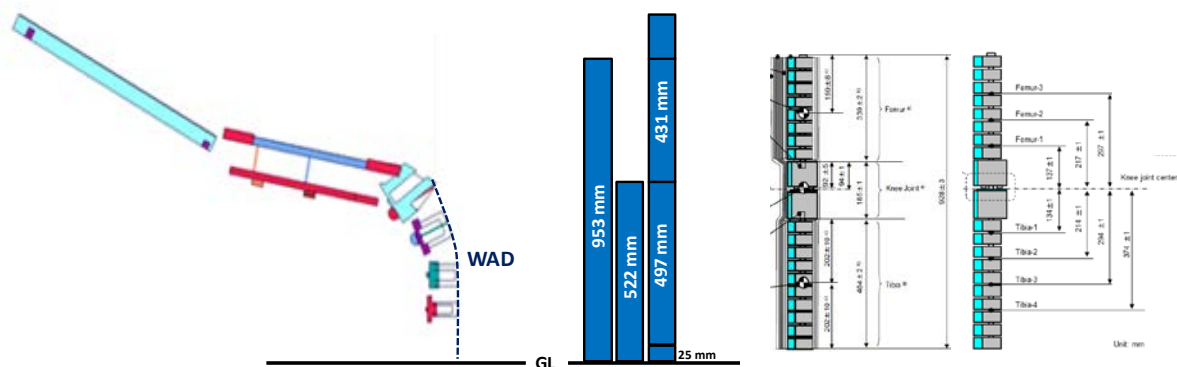


Fig. 1. Test conditions and impact heights for FlexPLI-UBM_{rubber}.

According to the geometry of the FlexPLI-UBM_{rubber/rigid}, the wrap around distances (WAD) of the BLE should not be lower than 522mm since this value marks the lower end of the femur when the FlexPLI_{rubber/rigid} is tested at an impact height of 25mm. For WAD of the bonnet leading edge reference line (BLE-RL) up to and including 953mm, the impact height is 25mm above ground level. In case of the BLE-RL WAD between 954mm and including 1,000mm, the FlexPLI-UBM impact height is further increased between 1mm and 47mm, resulting in an impact height between 26mm and 72mm. In case of the BLE-RL WAD greater than 1,000mm, the FlexPLI-UBM impact height remains unchanged at 72mm.

Test Procedures

Based on a broad variety of FE simulations with HBM and impactor models against generic front-ends as well as actual vehicle models, updated lower extremity test and assessment procedures were drafted and subsequently applied to physical component tests, using the FlexPLI with and without UBM. In terms of the test area to be assessed with the FlexPLI-UBM, reference was made to the bumper test area as defined by Euro NCAP [5]. From anthropometric considerations, the impact height of the FlexPLI-UBM was defined as follows:

BLE-RL WAD $\leq$ 953: GL + 25mm

954 $\leq$ BLE-RL WAD $\leq$ 1,000: GL + 26mm ... 72mm

BLE-RL WAD $>$ 1,000: GL + 72mm

Experimental Testing

For validation of the simulation results and the draft test and assessment procedures, a test tool for entirely predicting lower extremity injuries to the femur, the knee and the tibia was prototyped (see Fig. 2) and tested according to modified procedures.

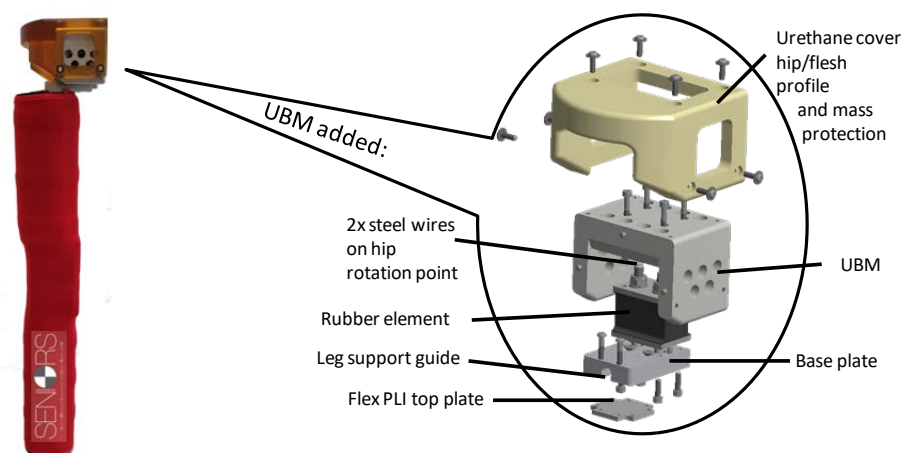


Fig. 2. FlexPLI with UBM with flexible connection (FlexPLI-UBM_{rubber}).

A series of tests with the FlexPLI Baseline, the FlexPLI-UBM_{rigid} and the FlexPLI-UBM_{rubber} was performed against the SAE Buck (Sedan) and three actual vehicles. Besides the principal goal of checking the applicability of the test procedures, further aims were to: compare the correlation between the different derivatives of the FlexPLI, investigate the correlation between FE simulation and physical testing, study the repeatability of test results, crosscheck the impactor behaviour at the end of (-EoB / EoB), close to (EoB-84mm) and beyond (EoB+84mm) the bumper beam and the possibility of compensating extraordinary high rotation of the impactor by turning the vehicle accordingly (EoB 30°, see Fig. C-6). TABLE II summarises the different types of tests that have been performed with the FlexPLI, FlexPLI-UBM_{rigid} and FlexPLI-UBM_{rubber}:

TABLE II
TEST MATRIX FOR PHYSICAL TESTS WITH THE FLEXPLI AND ITS DERIVATIVES

Surrogate/ Vehicle	FlexPLI Baseline			FlexPLI-UBM _{rigid}			FlexPLI-UBM _{rubber} (WS)					
Impact height	75mm			25...72mm			25...72mm					
Actual Van/MPV	-EoB	y0	EoB (2)	-EoB	y0	EoB	-EoB -84	-EoB 30°	-EoB (3)	y0 (3)	EoB (3)	
Actual SUV	-EoB	y0	EoB	-EoB	y0	EoB			-EoB (3)	y0 (3)	EoB 30°	EoB +84
Actual Sedan I	-EoB	y0	EoB -84				-EoB -84	-EoB	-EoB +84	y0 (2)	EoB -84	EoB (2) +84
SAE Buck (Sedan)		y0			y0			y0 (3)			y0+30° (3)	

III. RESULTS

FE Simulations

To study the effect of a pedestrian's torso on the kinematics and loadings of lower extremities, FE impactor simulations with the FlexPLI, FlexPLI-UBM_{rigid} and FlexPLI-UBM_{rubber} were compared with the results obtained with THUMSv4. For the FlexPLI-UBM_{rigid}, the correlation of the mass center of gravity (CoG) for four different adjustable positions was investigated in a previous study [15]. Here, the centre lower position turned out to have the best correlation to HBM simulation results and thus, this position was chosen for subsequent validations.

To summarise the peak bending moment and elongation results of THUMSv4, impacts on the LHS of the bumper beam (-EoB) produced, apart from very few exceptions (Medial Collateral Ligament (MCL) vs. actual SUV, femur and Anterior Cruciate Ligament (ACL) vs. compact car) lower readings than those on the opposite end of beam (RHS, +EoB) [16]. Besides, in most cases ACL was lower than PCL output. The knee geometry and HBM orientation are exemplarily depicted for the Van/MPV derivative of the SAE Buck in Fig. C-3 in Appendix C. On the contrary, the FlexPLI Baseline as well as both versions of the FlexPLI-UBM demonstrated during a great number of simulations a symmetrical performance of most vehicle frontends on both ends of the bumper beam [16]. This phenomenon – asymmetrical behaviour of the HBM vs. symmetrical behaviour of the impactor – needs to be taken into account during subsequent studies of the FlexPLI-UBM biofidelity as a function of correlation between HBM simulations and impactor tests.

Prior to a comparison of the loadings on femur, tibia and the knee ligaments regarding kinematics, peak results as well as the corresponding time histories, it has to be considered that - as done for the impactors - the waveforms of the femur and tibia bending moments of the human body model were entirely generated with the moments around the y-axis (My) only, in order to demonstrate the direction of the primary load and indicate correct sign conventions. As a consequence, the actual resultant moments are not reflected in these plots. This however has a remarkable effect on the time histories during angled impacts only, when a moment around the x-axis (Mx) is induced, as e.g. at the end of the bumper beam of the actual Van/MPV, see left plot in Fig. C-4 (Appendix C). At vehicle centreline, the vast majority of bending is induced around the y-axis, as shown in the right plot of Fig. C-4. All further diagrams, other than the time history curves, include the actual resultant moments of the HBM.

Fig. 3 illustrates the impactor kinematics during the FE simulations against the SUV Buck at different timings. The illustrations clearly show the impact kinematics of the FlexPLI-UBM_{rigid} and FlexPLI-UBM_{rubber} superior to the FlexPLI Baseline. While the initiation of the rebound of the FlexPLI has already started at 30ms, its UBM derivatives are still in the assessment interval (AI), as defined in United Nations Economic Commission for Europe (UNECE) [12] with very similar kinematics to THUMSv4. At 48ms, when the FlexPLI Baseline is already in complete rearward motion with negative moments, the UBM impactors are still sustaining biofidelic loads. At 69ms, after release of the UBM impactors from the vehicle front the HBM is still being loaded, which indicates that the duration of impact cannot be fully addressed by including the upper body surrogates. The kinematics of the human leg during SUV impacts is thus better represented by the FlexPLI-UBM.

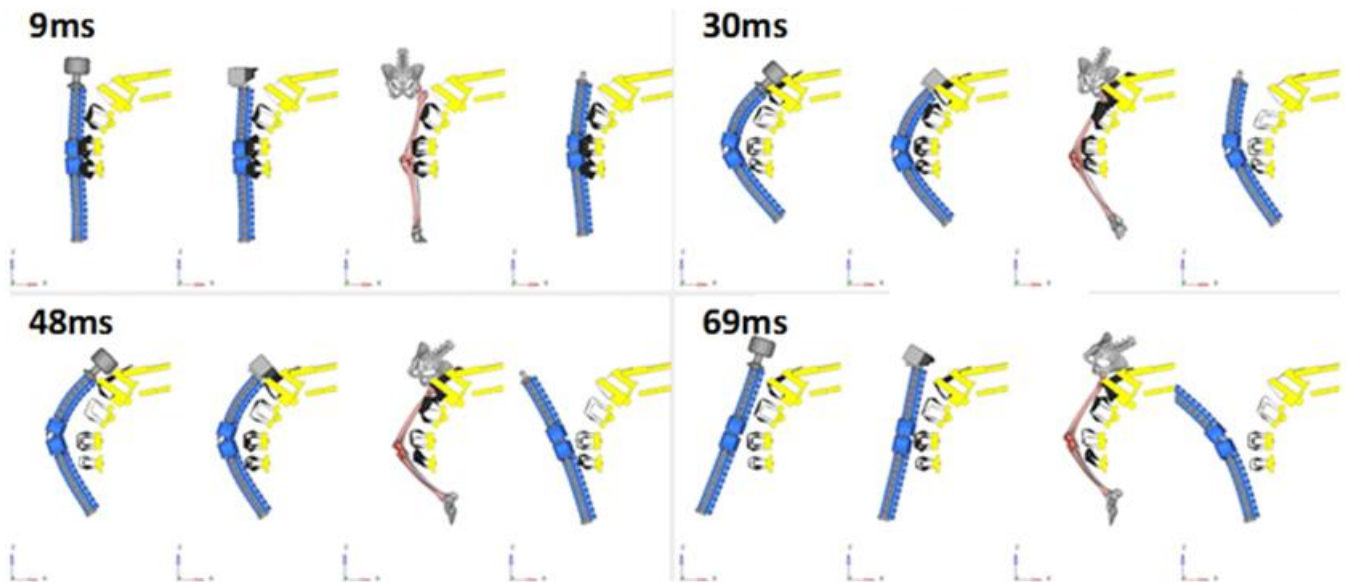


Fig. 3. Impact kinematics during simulations against SAE Buck (SUV).

The described phenomenon, as exemplarily depicted for the time history curves of Femur-2 (mid upper position, Fig. 4 (left)) was detected during all simulations with the SAE Buck, where the combination of low impactor mass (compared to the Buck) with the properties of the load path covers of the SAE Buck (material: polyethylene 300) resulted in the observed spring effects, especially for the FlexPLI Baseline and during later points in time of the impact phase.

Also during simulations against actual vehicles the impact duration of THUMSv4 was not fully reflected by the FlexPLI-UBM (compare Fig. 4 (right)); however the time histories got closer to those of the HBM.

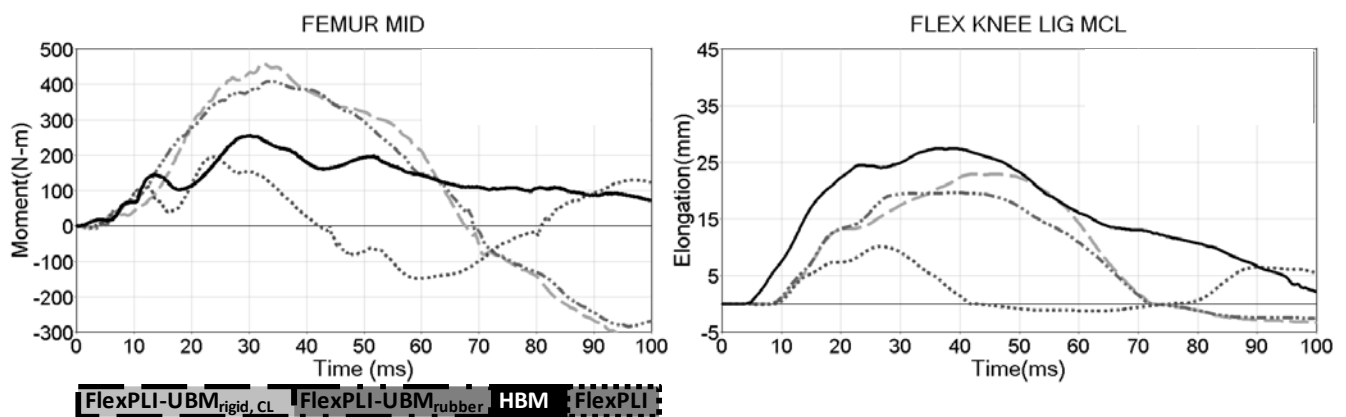


Fig. 4. Time histories of Femur-2 bending moment (HBM: M_y) during simulations against SUV Buck (left) and of MCL elongation during simulations against actual Van/MPV (right).

With UBM, the qualitative correlation between HBM and FlexPLI was also improved at EoB, as illustrated by the impact kinematics of THUMSv4 and the FlexPLI-UBM in Fig. C-5 (Appendix C). A comparison of the FlexPLI with and without applied UBM and the HBM shows an overprediction of the human lower extremities rotation by the FlexPLI but often good representation by the FlexPLI-UBM, though with a certain time lag.

Given the previously described reasons, additional simulations were performed with the vehicle rotated by 30°. As shown Fig. C-7 (Appendix C), after reaching its maximum bending during the impact against the non-rotated vehicle, the impactor was transferred into z rotation. In contrast, no additional impactor rotation was noticed when impacting the rotated vehicle. A perpendicular impact is thus expected to contribute to reducing unrealistic impactor rotation. The improved impactor kinematics due to vehicle rotation is however not always reflected by the corresponding time histories.

Altogether, a comparison of the simulation results reveals the significantly superior behaviour of the FlexPLI-UBM_{rigid} and FlexPLI-UBM_{rubber} over the FlexPLI Baseline in terms of kinematics and loadings, with further

advantages of the FlexPLI-UBM_{rubber}, such as simulating the hip rotation, time lag and the ability of the mass to maintain its impact direction longer relative to the leg, as illustrated in Fig. C-8 (Appendix C), a sometimes later UBM interaction with the vehicle front, leading to more humanlike results, and an occasionally slightly more realistic shape of the waveforms. This, however, was only partially reflected by the maximum loadings.

A study of peak result correlations was carried out with the final goal to establish transfer functions between THUMSv4 and the FlexPLI-UBM. Under consideration of all impacts as summarised in TABLE I, the good correlations between HBM and FlexPLI-UBM in terms of kinematics and time histories could not be confirmed by the correlation of peak values. Reasons can be seen amongst other things in the differences of material properties, thicknesses and geometries of the bones, measurement locations and timings between impactor and HBM. Besides, the results from SUV (upper femur and tibia) impacts with unintended or premature interactions between the UBM and the vehicle frontend during the impact phase, contributed to low correlations with incorrect timings and magnitudes of the peaks especially for the femur bending moments. Correlations were altogether thus not satisfactory when including all simulations, but significantly increased when categorising vehicles, particularly for the upper and middle femur segments, Femur_{max} and the two lower tibia segments during the Sedan impacts with FlexPLI-UBM (rigid and rubber), resulting in coefficients of determination (R^2) between 0.76 and 0.85. Please note that the individual number of data points of SUV and Van/MPV categories did not further allow a reliable correlation of their peak results.

For Sedan simulations, i.e. simulations without unintended interaction between UBM and the vehicle front, good transition equations for the femur bending moment could be established, see Fig. 5 (left). The regression lines not going through the datum point indicate the applicability for defined ranges of loadings.

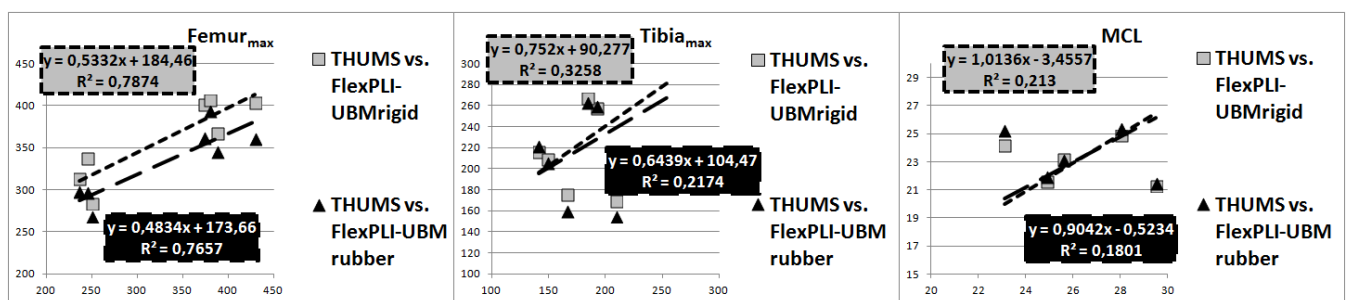


Fig. 5. Correlation of maximum femur and tibia peak bending moments and MCL elongation (THUMSv4 vs. FlexPLI-UBM) on the Sedan category.

Since correlations of THUMSv4 vs. FlexPLI-UBM maximum tibia bending moments and MCL elongations were unsatisfactory (compare Fig. 5, middle and right) while good quantitative correlations between a human FE pedestrian lower limb model and FlexPLI were found with R values of 0.91 (Tibia_{max}) and 0.56 (MCL) during the development and evaluation of the FlexPLI [17], further investigations in this area were focused on a comparison of FlexPLI vs. FlexPLI-UBM output values. Here it could be found that the tibia and MCL results correlated well with coefficients of determination of 0.62 and 0.57 for the FlexPLI-UBM_{rubber}, however with comparatively small slope of the regression lines, see Fig. 6:

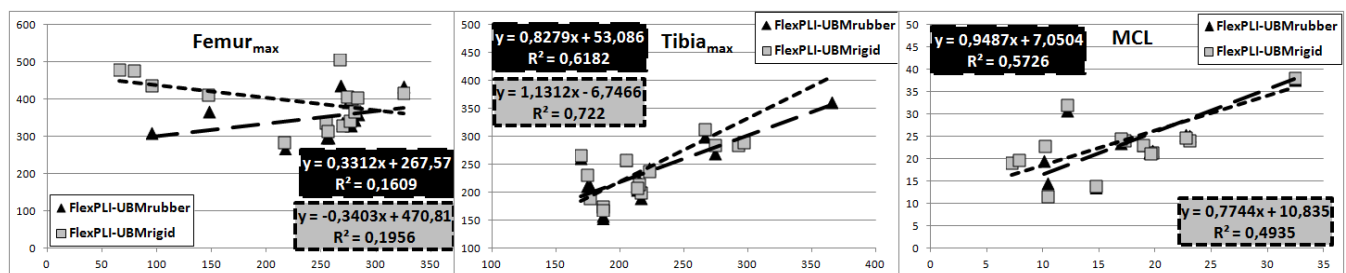


Fig. 6. Correlation of maximum femur and tibia peak bending moments and MCL elongation (FlexPLI Baseline vs. FlexPLI-UBM_{rubber/rigid}).

Taking into account the abovementioned correlations and the purpose of the FlexPLI-UBM test as a unique replacement of the upper or lower legform to bumper as well as the upper legform to WAD 775 test, the

FlexPLI-UBM_{rubber} impactor thresholds are derived by a transformation of human femur injury criteria and FlexPLI Baseline tibia and MCL limits.

No relationship between degree of quantitative correlation and impact angles could be found; thus both the FlexPLI-UBM_{rigid} as well as the FlexPLI-UBM_{rubber} reveal improvements compared to the FlexPLI Baseline when tested on areas at or around EoB.

Assessment

In most cases the performance of the FlexPLI-UBM_{rigid} and the FlexPLI-UBM_{rubber} were similar regarding time histories and quantitative correlation. Due to the superior impact kinematics of the FlexPLI-UBM_{rubber} this derivative was finally chosen for prototyping. Upper (UPL) and lower (LPL) performance limit definitions for the assessment were thus derived from transition equations based on correlations with FlexPLI-UBM_{rubber}.

Femur: Injury risk functions for the thigh and the bare femur loaded at mid shaft and distal third in three point bending tests were developed, as reported in [18]. These injury risk functions were based on test data that was normalised to represent the 50th percentile, average height pedestrian. The normalisation was based on the geometry of a single, commercially available anatomical femur model that was described as 50th percentile. However, review of the data suggested that the mean height of the subjects was already close to the 50th percentile: The mean height of subjects in the reported data was 1.765m, with 29 out of 34 subjects being male and five female. The mean height for males and females in the UK is 1.755m and 1.734m [19], which suggests that the mean height for the reported proportion of males and females would be 1.735m (or 1.7% less than the mean height of subjects in the reported data). A similar height is also defined for 50th percentile male ATDs. Therefore, the normalisation should have had very little effect on the mean bending moment for fracture. However, the reported normalisation reduced the mean bending moment for fracture from 465 Nm to 386 Nm, a reduction of 17% - which is far greater than the 1.7% difference in height between the reported sample and the 50th percentile. Therefore, it was decided to reanalyse the data in [18] to remove the normalisation. This analysis was performed using R-scripts developed in the SENIORS project, building on the survival analysis method published in ISO TS 18506:2014 and adding the capability to use multiple covariates. The analysis indicates that subject age, mass, impact location and presence of flesh (cf. bare femur bone) were all significant covariates. The reported data contained 34 data points and ideally more than 40 data points would be used for four covariates (for exact data, such as that available in the data set in [18]), giving more confidence in the parameterisation. Fig. 7 depicts the resulting injury risk function for mid-shaft femur fracture, based on a Weibull distribution, for a 65-year-old with a mass of 75kg. TABLE III shows the mid-shaft femur bending moment associated with different risks of femur fracture for the same data.

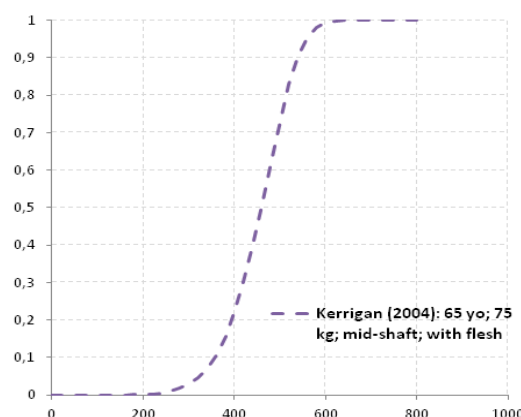


Fig. 7. Injury risk curve for femur bending moment, based on analysis of data from [18].

TABLE III
FEMUR FRACTURE RISK FOR 65-YEAR-OLD, 75 KG THIGH LOADED AT THE MID-SHAFT, BASED ON ANALYSIS OF
DATA FROM KERRIGAN ET AL. (2004)

Femur fracture risk (%)	10	20	25	30	35	40	45	50	60	70	80	90
Mid-shaft bending moment (Nm)	356	394	408	420	430	440	450	459	476	494	514	539

The FlexPLI represents the pedestrian's lower extremities including a simulation of the human flesh. Therefore, the relevant thigh fracture risks at mid shaft derived from the reanalysed data of 394Nm (for the 20% risk, as upper performance) and 459Nm (for the 50% risk, as lower performance) respectively were taken into account. Finally, the application of the transition equation in Fig. 5 resulted in the following FlexPLI-UBM_{rubber} femur bending moment UPL and LPL:

$$Femur_{max,UPL}(FlexPLI - UBM_{rubber}) = (0.4834 * 394 + 173.66)Nm = 364Nm \quad (1)$$

$$Femur_{max,LPL}(FlexPLI - UBM_{rubber}) = (0.4834 * 459 + 173.66)Nm = 396Nm \quad (2)$$

Tibia: The current FlexPLI performance limits were derived from two tibia fracture risk curves based on different underlying sets of PMHS data, anthropometric databases, scaling methods and statistical procedures [20]. As tibia UPL, a 20% risk for human tibia bone fracture was chosen by Euro NCAP. According to the FlexPLI vs. HBM transition equation the FlexPLI threshold was determined calculating the average value based on both risk curves (300Nm and 244Nm respectively for 20% human injury risk):

$$\text{Risk curve I:} \quad Tibia_{max}(FlexPLI)_I = 1.259 * Tibia_{max}(HBM) - 72.798 = 305Nm \quad (3)$$

$$\text{Risk curve II (cons. of 10% scatter):} \quad Tibia_{max}(FlexPLI)_{II} = (1.259 * Tibia_{max}(HBM) - 72.798) * 1.1 = 258Nm \quad (4)$$

$$Tibia_{max,UPL}(FlexPLI) = MV(Tibia_{max}(FlexPLI)_I + Tibia_{max}(FlexPLI)_{II}) = 282Nm \quad (5)$$

For the LPL, the type approval requirement [12] was chosen. Here, the 30% human bone fracture risk at 330Nm human tibia bending moment of Risk Curve I was directly transferred into the corresponding FlexPLI threshold:

$$Tibia_{max}(FlexPLI)_{LPL} = 1.259 * Tibia_{max}(HBM) - 72.798 = 343Nm \quad (6)$$

This value was rounded to 340Nm.

Under consideration of the transition equation from Fig. 6 and the current Euro NCAP UPL and LPL for the FlexPLI (282Nm and 340Nm), FlexPLI-UBM_{rubber} tibia bending moment UPL and LPL result as follows:

$$Tibia_{max,UPL}(FlexPLI - UBM_{rubber}) = 0.8279 * 282 + 53.09)Nm = 287Nm \quad (7)$$

$$Tibia_{max,LPL}(FlexPLI - UBM_{rubber}) = 0.8279 * 340 + 53.09)Nm = 335Nm \quad (8)$$

Knee: For a calculation of the risk of injuries to the human knee, the averaged 50% risk derived from two human injury risk functions [21][22] resulted in a knee bending angle (KBA) of 19°. This value was transferred to a corresponding elongation of the medial collateral ligament (MCL-EL) of 15.9mm under the assumption of the human KBA being equal to the HBM KBA [23].

According to the FlexPLI vs. HBM transition equation [17] and incorporating the effect of muscle tone [24], the FlexPLI threshold was calculated to 21.1mm.

Comparing with the legal requirement of 19° for the EEVC WG 17 legform impactor [25], it was found that the calculated MCL requirement, correlating to an EEVC WG 17 legform impactor KBA of approx. 17°, would have a higher stringency. Therefore, and since this value was also in line with a correlation study between FlexPLI MCL elongation and EEVC WG 17 legform impactor KBA [26], the FlexPLI threshold in [12] was set to 22mm.

Euro NCAP defined this value as LPL. For the UPL, an MCL elongation correlating to 15° KBA of the EEVC WG 17 legform impactor, as the previous Euro NCAP UPL, was chosen at 19mm according to the abovementioned correlation study.

Inserting the current Euro NCAP UPL and LPL of 19 and 22mm into the MCL transition equation from Fig. 6 results in FlexPLI-UBM_{rubber} MCL elongation UPL and LPL as follows:

$$MCL - EL_{max,UPL}(FlexPLI - UBM_{rubber}) = 0.9487 * 19mm + 7.0504mm = 25mm \quad (9)$$

$$MCL - EL_{max,LPL}(FlexPLI - UBM_{rubber}) = 0.9487 * 22mm + 7.0504mm = 28mm \quad (10)$$

Rating scheme and test synthesis: For the development of a VRU rating scheme, the relevance of VRU femur and tibia injuries relative to all addressed injuries (i.e. femur, tibia and knee) is estimated at 30% each while the

relevance of VRU knee (MCL) injuries is estimated at 40%. Here it needs to be added, that ACL/PCL ruptures in lateral vehicle to pedestrian accidents are expected to be covered by the protection of the MCL [27]; thus, a separate injury criterion seems redundant.

When maintaining the current rating scheme of Euro NCAP with a maximum score of 1 point per grid point, a maximum score of 0.3 points each for femur and tibia assessment and a maximum score of 0.4 points for MCL assessment can be achieved per impact point in case of meeting the UPL. When exceeding the LPL, no femur (tibia, MCL) points are awarded to the test point. Between UPL and LPL, a sliding scale is applied, as illustrated in Fig. D-1 (Appendix D).

An overview of FlexPLI-UBM assessment is given in Fig. 8. The total score for each impact location is calculated by adding the scores for femur and tibia bending moment and the MCL elongation. All grid point scores are summed up to the total grid point score. As before, the total lower legform score will be calculated by scaling the total grid point score to the maximum number of achievable points for this subsystem test (12 in the example of Fig. 8):

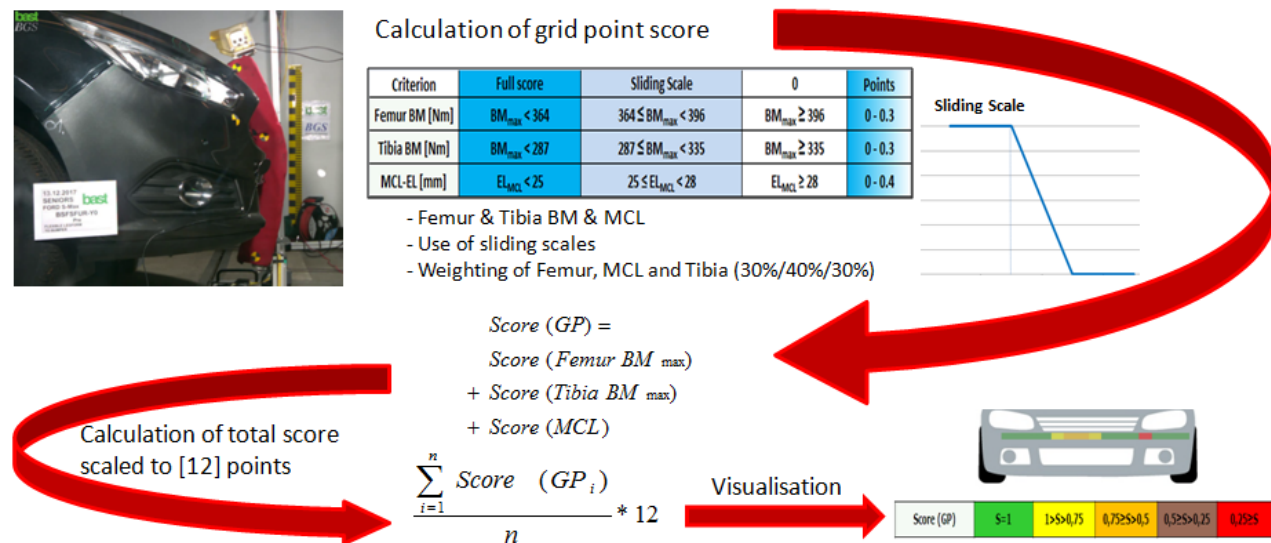


Fig. 8. FlexPLI-UBM assessment procedure.

Experimental Tests

Tests against the SAE Buck and actual representatives of the Sedan, SUV and Van/MPV categories, as summarised in TABLE II, confirmed the significantly improved kinematics of the FlexPLI with UBM as seen in the simulations. In particular during the AI of the impact, the kinematics correlated well between THUMSv4 and FlexPLI-UBM while the rebound of the FlexPLI Baseline was already initiated, as illustrated in Fig. 9 for the Van/MPV impact at vehicle centreline:

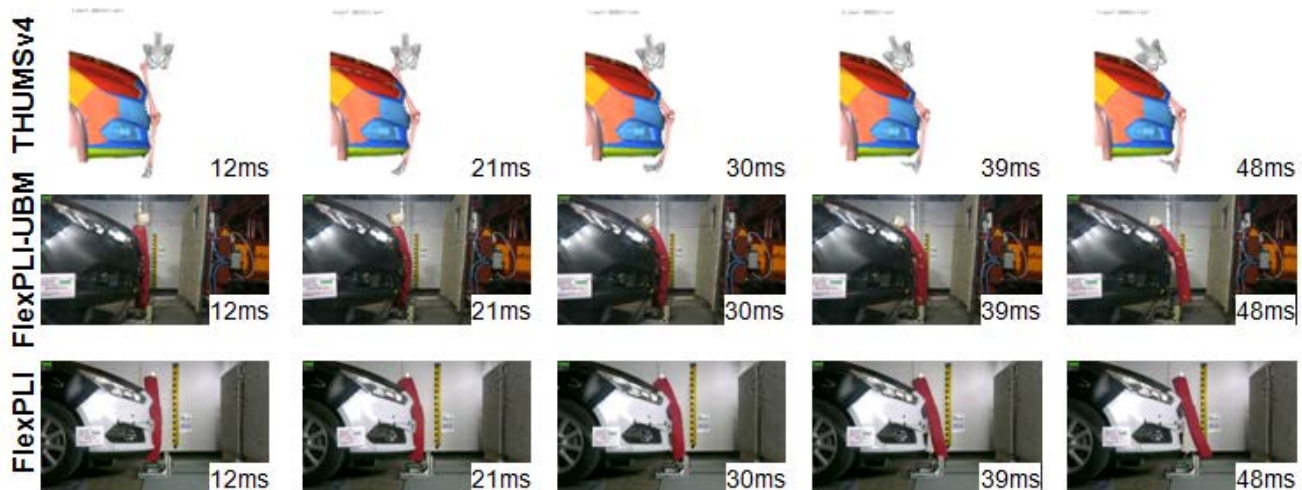


Fig. 9. THUMSv4, FlexPLI-UBM and FlexPLI impact at Van/MPV centreline.

The overall good kinematics correlation between THUMSv4 and FlexPLI-UBM was confirmed by the time histories, see Fig. 10. Duration of impact, timings of maximum loadings and shapes of the waveforms correlated well, while the FlexPLI Baseline showed a clear mismatch particularly for femur and knee. Furthermore, the repeatability of FlexPLI-UBM maximum results was good with coefficients of variation (CV) of less than 2% for femur, between 0.8% and 6% for tibia and 5.1% for MCL.

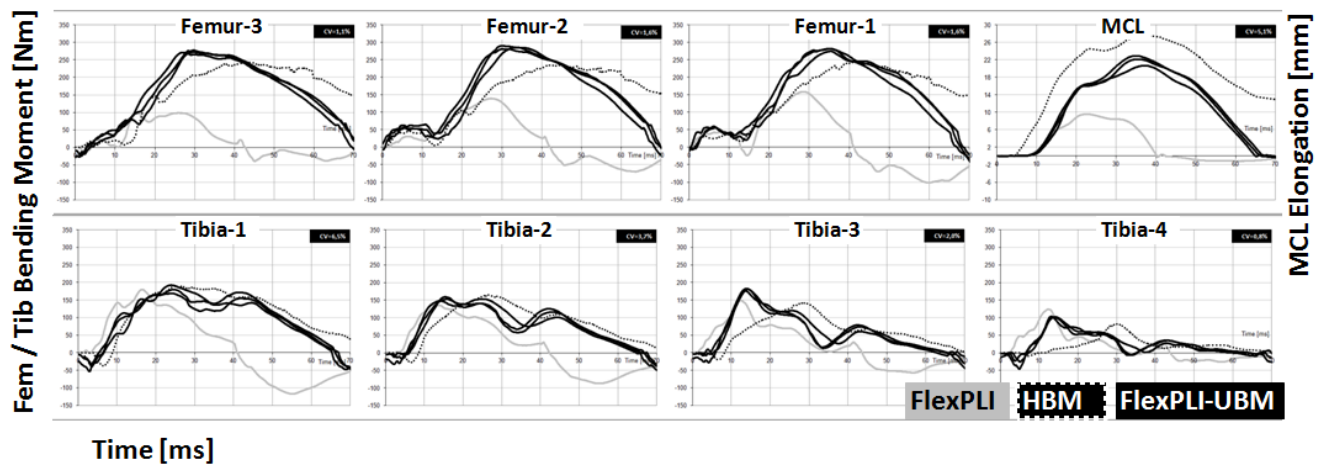


Fig. 10. Qualitative correlation for FlexPLI Baseline vs. HBM and FlexPLI-UBM vs. HBM – Van/MPV centreline.

A comparison of the kinematics during the centreline impact against the SUV representative is depicted in Fig. E-1 (Appendix E). The FlexPLI-UBM well replicated the behaviour of the HBM lower extremities during the impact while the FlexPLI Baseline was already in the rebound phase when the HBM had not reached maximum loadings, compare Fig. E-2 in Appendix E. Like for the Van/MPV, the FlexPLI-UBM showed good repeatability of test results on the SUV with CVs between 0.5% and 2.8% for femur and tibia and 9% for MCL.

Besides centreline impacts, the performance of the FlexPLI was studied during impacts against angled front surfaces at and beyond EoB. Fig. E-3 in Appendix E illustrates the superior kinematics of the FlexPLI-UBM with much less rotation and more humanlike behaviour at EoB with no rebound during the AI. The time histories confirm the good qualitative correlations (Fig. E-4, Appendix E) with waveforms very similar between THUMSv4 and the FlexPLI-UBM regarding shape, peak loadings and timings and the FlexPLI Baseline sometimes significantly differing, especially in the uppermost femur and lowermost tibia segments.

Another focus of the present study was set on the possible further benefit when providing a setup for perpendicular impacts which was realised by rotating the vehicle by 30 degrees around its z-axis. Additionally, applicability of the FlexPLI-UBM on further outboard areas with steeper impact angles was investigated. Confirming the simulation results, Fig. E-5 in Appendix E demonstrates rotating the vehicle limiting the rotation of the impactor during the impact phase on the one hand, while maintaining the extent of rotation further outboard the EoB on the other hand.

These observations are confirmed by a comparison of the corresponding time history curves, see Fig. E-6 (Appendix E). Here, especially the waveforms for femur and knee segments did not show any significant differences. This indicates, in principal, applicability of the vehicle rotation. However, at approx. 45ms after the impact the rotated vehicle generated a second peak in the femur values which was not reflected by the waveforms of the HBM.

IV. DISCUSSION

An extensive series of simulations with THUMSv4 and the FlexPLI with and without UBM has been carried out against a generic buck and actual vehicles according to draft test procedures, taking into account human anthropometry and the geometry of the lower legform impactor FlexPLI. While both rigid and flexible attachment of an UBM were considered; the latter was found to be superior in terms of kinematics, providing time lag and hip joint rotation as seen during HBM simulations. Correlation studies were performed and transition equations for THUMSv4 vs. FlexPLI-UBM and FlexPLI Baseline vs. FlexPLI-UBM were defined, giving

satisfactory coefficients of determination in order to draft assessment procedures for future inclusion in consumer test programmes.

Experimental tests against generic as well as actual vehicle frontends showed the potential benefit when introducing the UBM into the FlexPLI for lower extremity testing. Assessing femur injuries became possible for all different vehicle categories taken into account in this study, including the Sedan, SUV and Van/MPV. The tests furthermore showed good repeatability and the possibility of combining the upper legform to WAD775 and lower (or upper) legform to bumper test within one unique test. The application of the UBM also contributes to a significant limitation of impactor rotation during tests against angled surfaces, comparable to the kinematics of human lower extremities. It must however be taken into account that the symmetrical behaviour of the FlexPLI is not in line with asymmetrical output values of THUMSv4 during impact against angled surfaces at both EoB. Vehicle rotation contributed to a further reduction of impactor rotation but, based on the tests so far, did not improve biofidelity.

The test programme was limited to a total number of 51 tests. A round-robin test series including more vehicle shapes and test laboratories, investigating the robustness of the test procedures and tools as well as reproducibility of test results would be beneficial before final inclusion. The reduced costs for tests and vehicle spare parts will offset the adaptations to test equipment needed in some laboratories.

V. CONCLUSIONS

The present study proposes for the first time a combination of pedestrian upper leg, knee and lower leg injury assessment based on one combined component test using the FlexPLI with UBM, offering improved biofidelity and injury assessment ability compared to the FlexPLI. Test and assessment procedures for implementation in consumer rating are developed, replacing those for the upper legform to WAD775 and the upper or lower legform to bumper, addressing the vehicle categories Sedan, SUV and Van/MPV and significantly contributing to the robustness of results at the end of the test area. Since the established femur transfer function does not necessarily cover the entire vehicle fleet, the development of a range with consistent FlexPLI-UBM injury risk prediction should be considered during further research. Besides, further modification of the impact height independently from the injury causing vehicle part should be taken into account. A real-world impact assessment is recommended to be done after sufficient market penetration of vehicle design countermeasures meeting the modified requirements.

VI. ACKNOWLEDGEMENT



The research leading to the results of this work has received funding from the European Union's Horizon 2020 Research and Innovation Programme under Grant Agreement No 636136.

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VIII. APPENDICES

APPENDIX A: COLLISION DATA

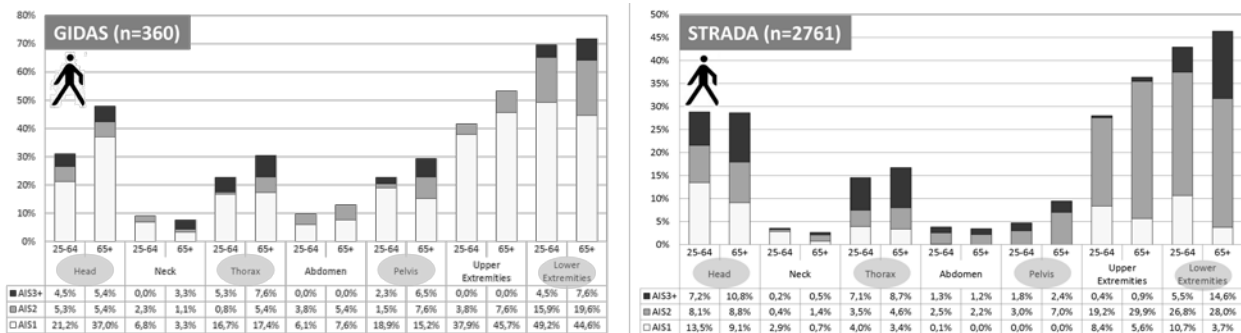


Fig. A-1. Percentages of AIS1, AIS2 and AIS3+ injury severities for the different pedestrian body regions within GIDAS and STRADA. Each column adds up to 100% by adding all percentages from AIS0 to AIS9 [3].

APPENDIX B: WORKFLOW

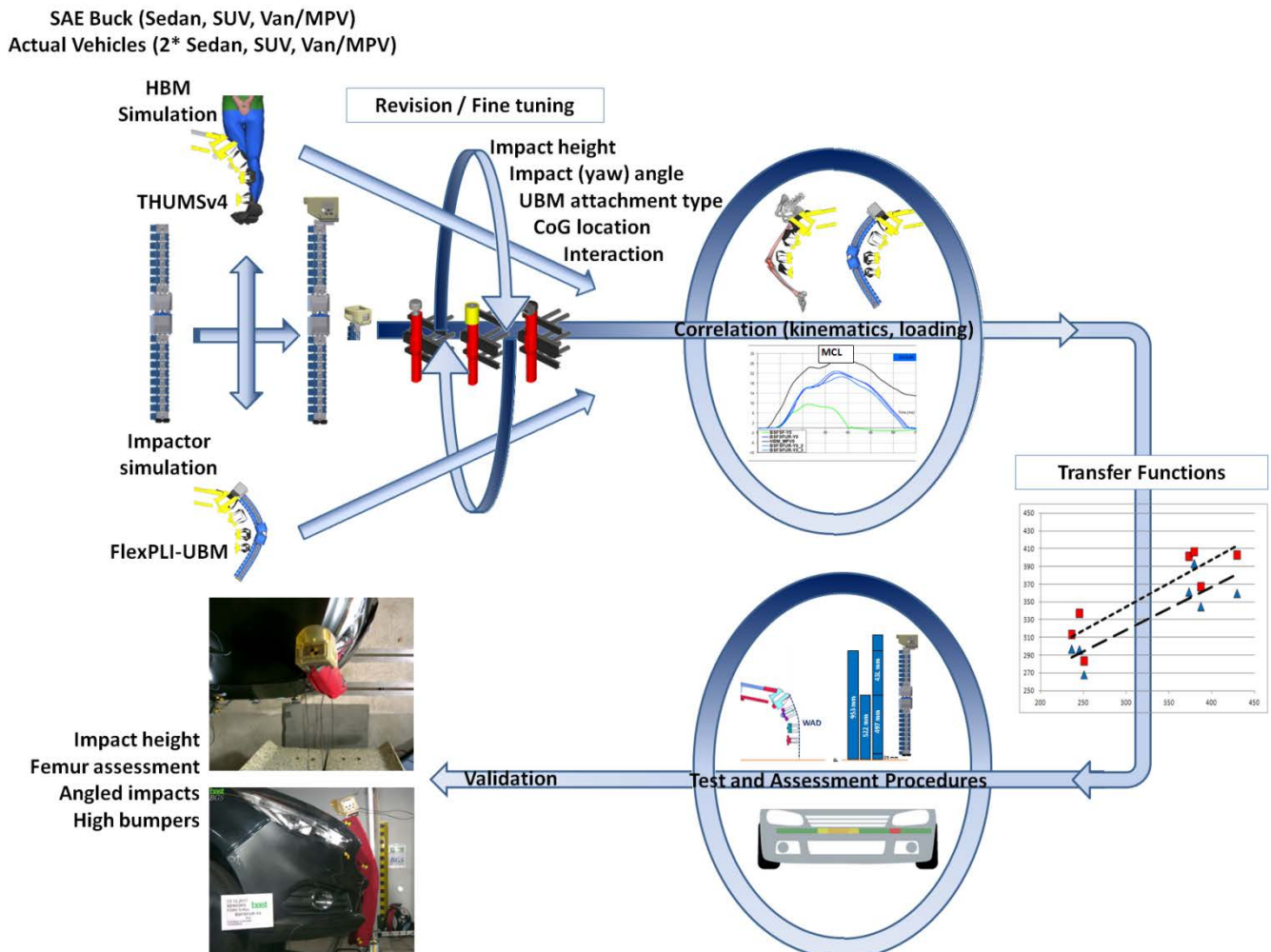


Fig. B-1. Workflow describing the development of the FlexPLI-UBM and its test and assessment procedures.

APPENDIX C: SIMULATIONS

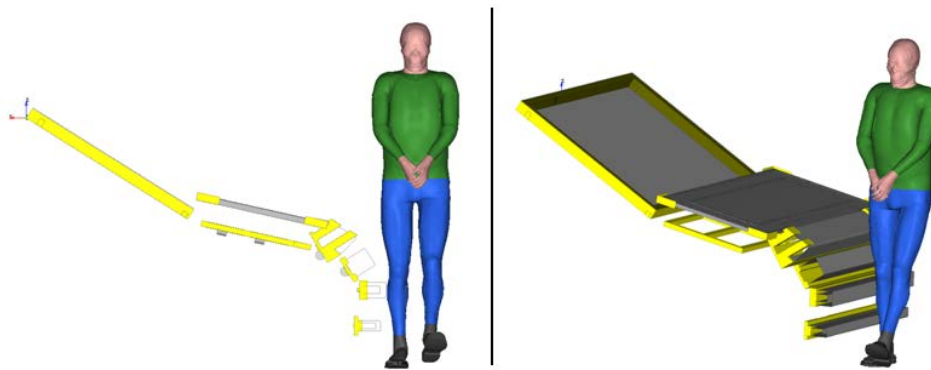


Fig. C-1. THUMSv4 positioning in SAE Buck and vehicle simulations. Example SAE Buck (Sedan).

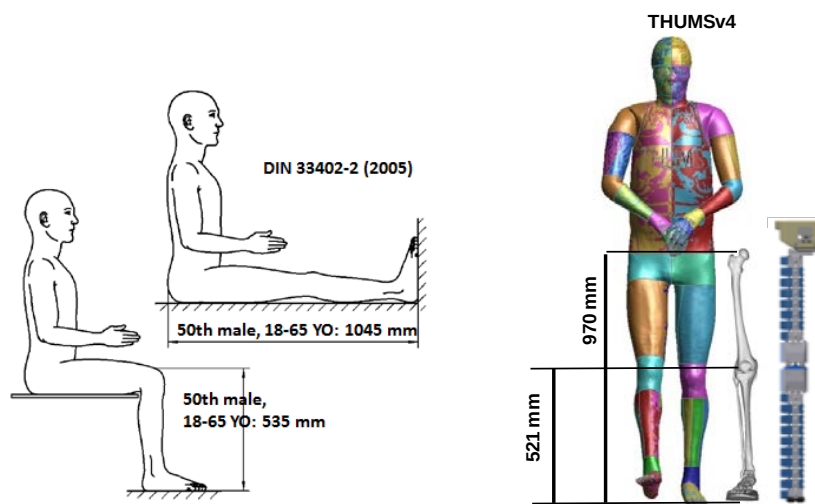


Fig. C-2. Anthropometric data: human and THUMSv4 (50th).

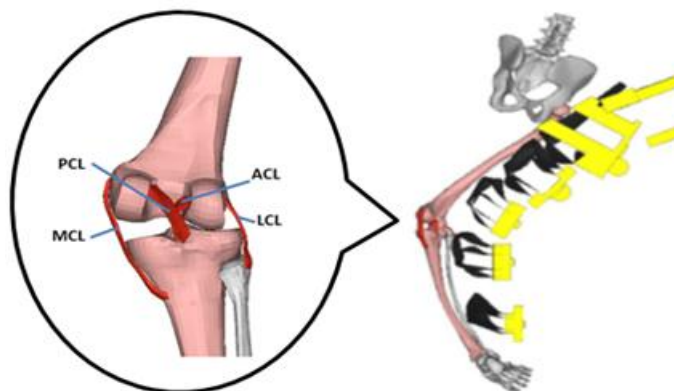


Fig. C-3. THUMSv4 knee geometry and ligament elongations during impact.

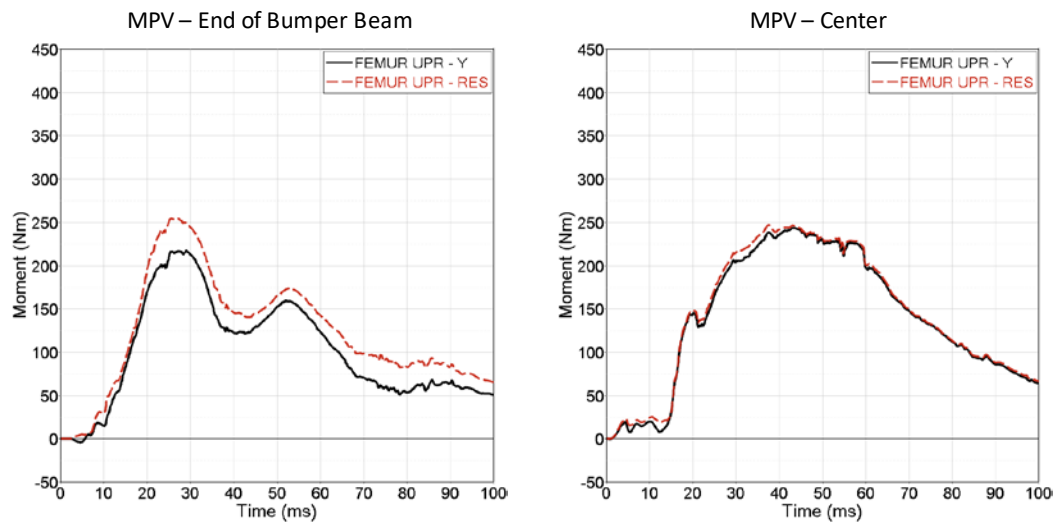


Fig. C-4. HBM Femur-1 bending moment M_y vs. M_{res} at EoB (left) and at vehicle centreline (right) – Example Van/MPV.

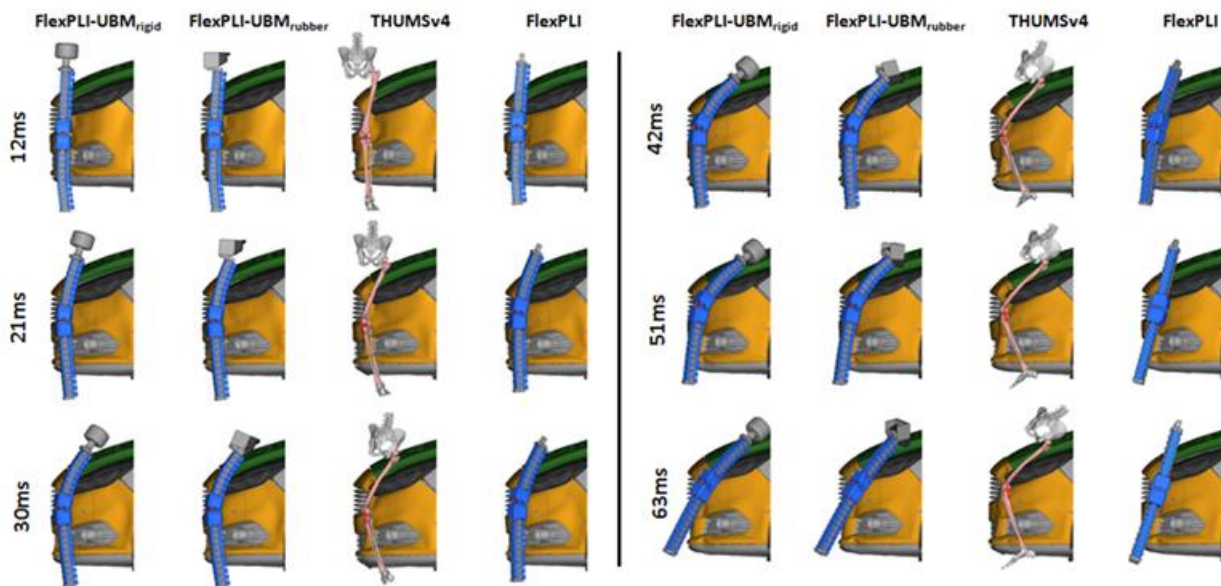


Fig. C-5. Comparison of impactor and HBM rotation during impact against actual MPV at EoB.

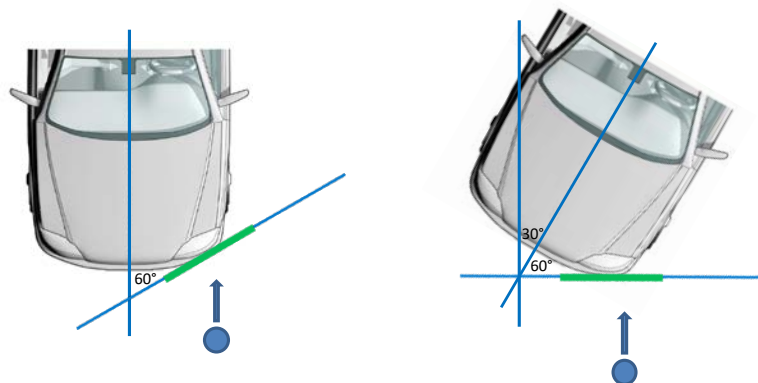


Fig. C-6. Vehicle rotation to compensate for FlexPLI rotation at angled surfaces.

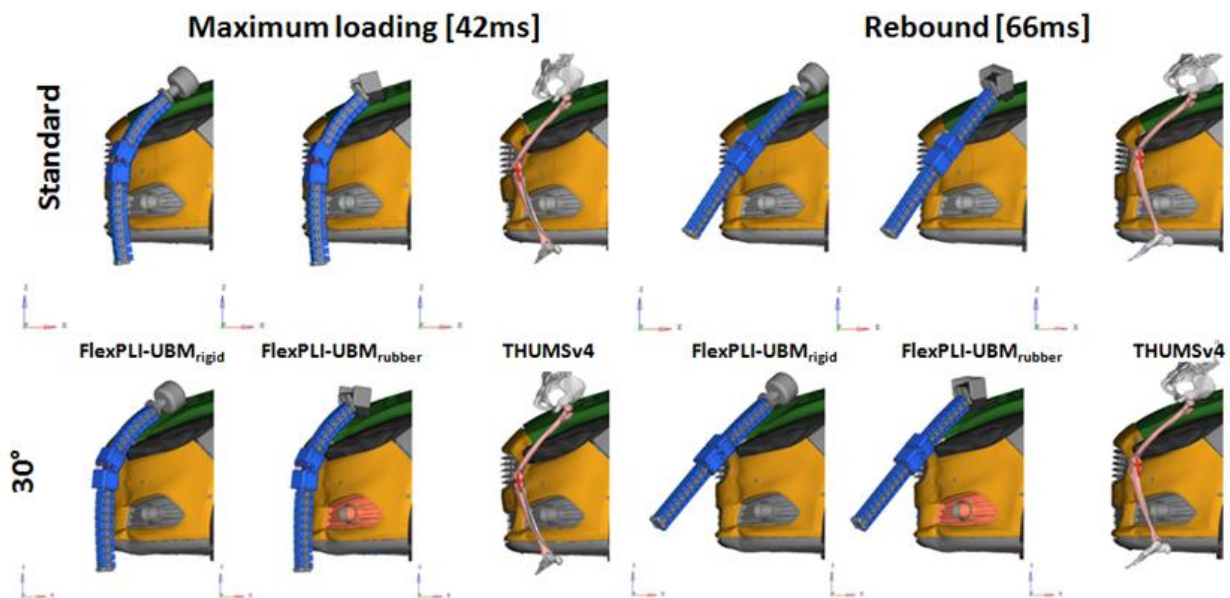


Fig. C-7. Impactor and HBM kinematics during impact against -EoB of non-rotated (upper row) and rotated (lower row) vehicle at time of maximum loading (left) and during rebound (right).

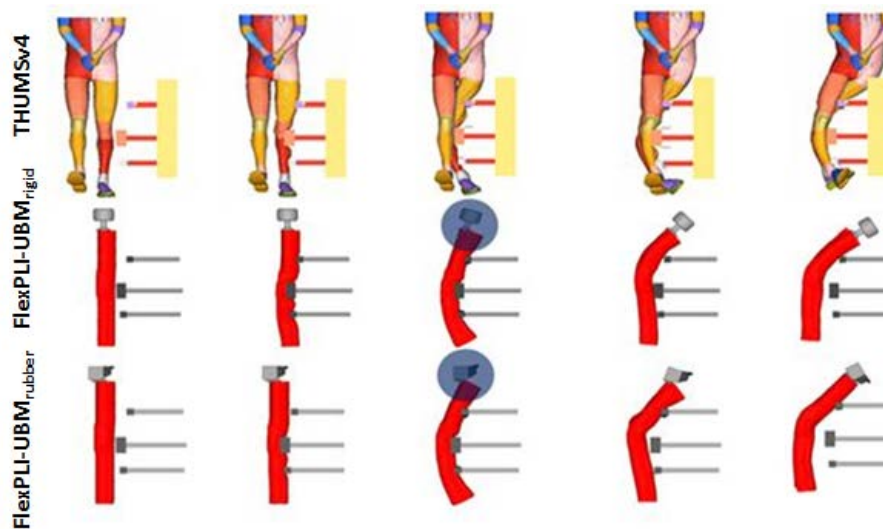


Fig. C-8. Impact kinematics during simulations against a generic test rig [15].

APPENDIX D: ASSESSMENT

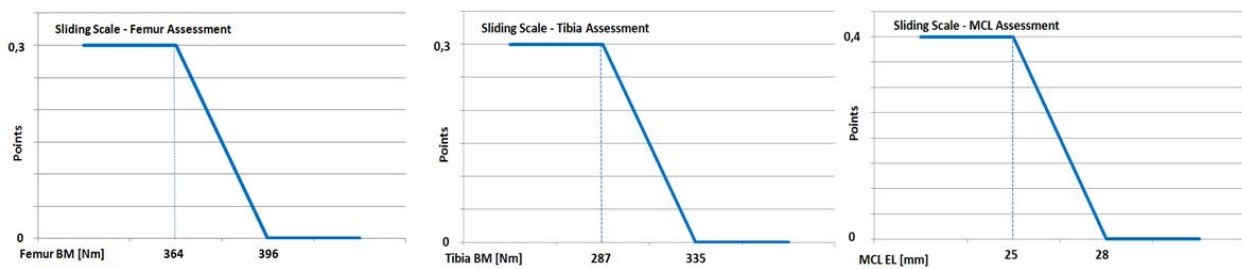


Fig. D-1. Sliding Scale for the femur, tibia and MCL assessment of the FlexPLI-UBM_{rubber}.

APPENDIX E: ADDITIONAL TEST RESULTS



Fig. E-1. THUMSv4, FlexPLI-UBM and FlexPLI impact at SUV centreline.

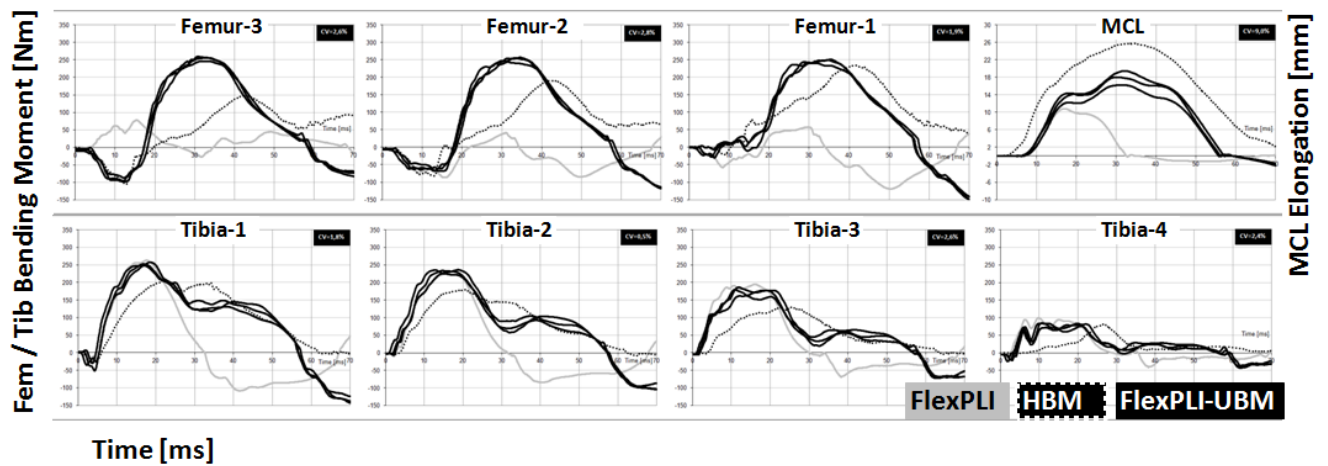


Fig. E-2. Qualitative correlation for FlexPLI Baseline vs. HBM and FlexPLI-UBM vs. HBM – SUV centreline.

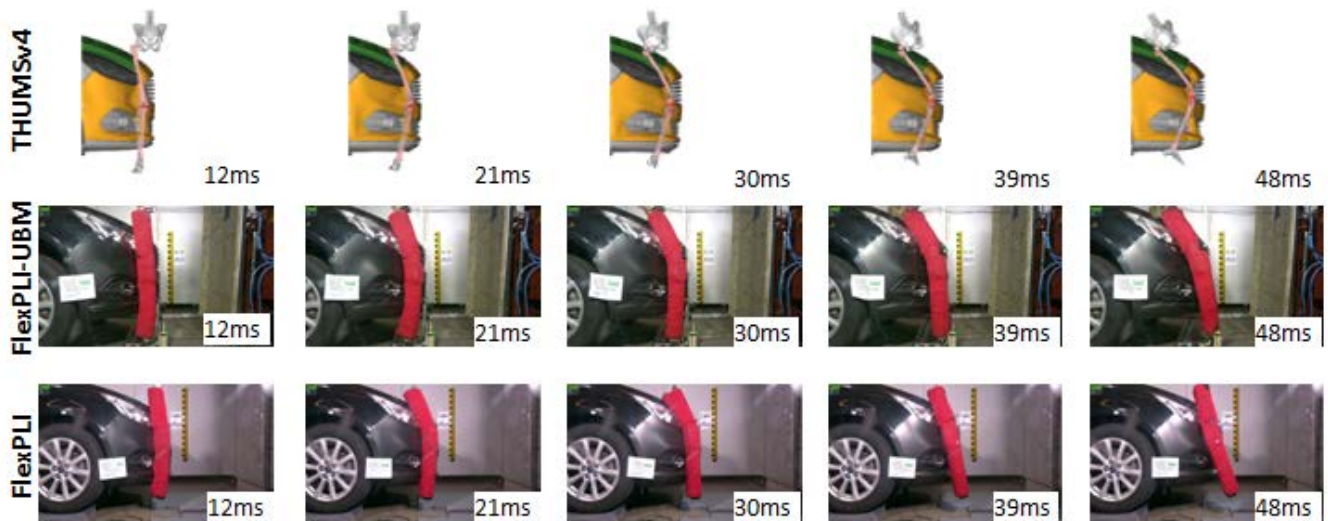


Fig. E-3. THUMSv4, FlexPLI-UBM and FlexPLI impact at Van/MPV +EoB (RHS).

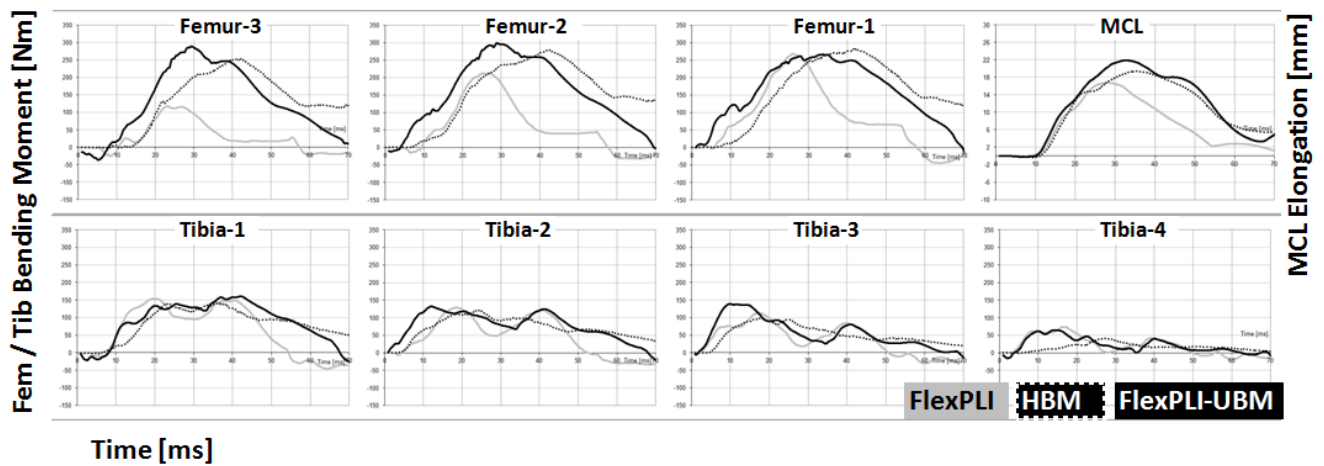


Fig. E-4. Qualitative correlation for FlexPLI Baseline vs. HBM and FlexPLI-UBM vs. HBM – Van/MPV +EoB (RHS).

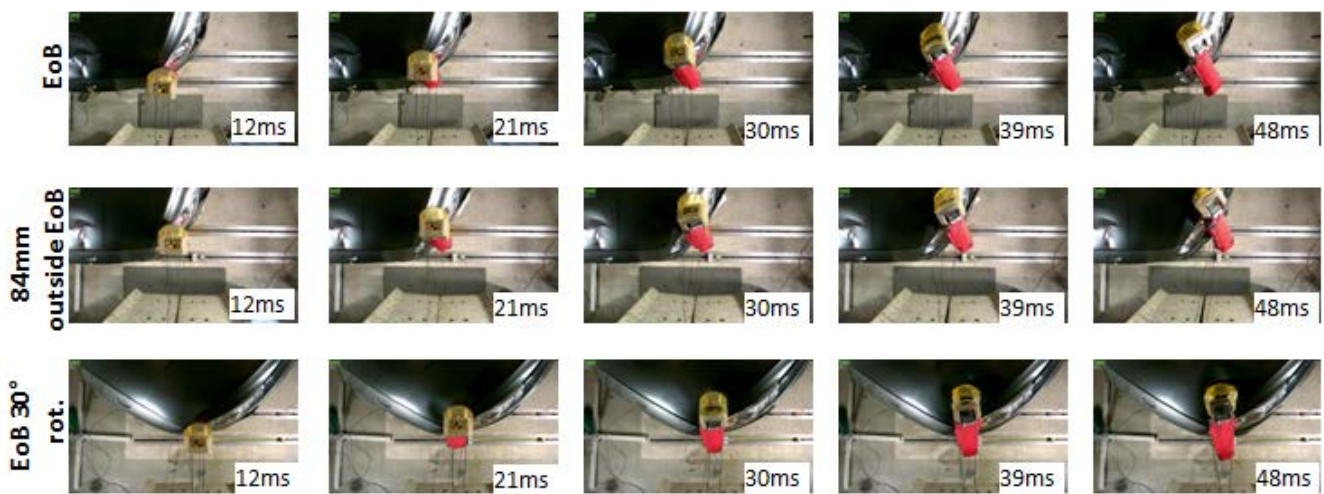


Fig. E-5. FlexPLI-UBM impact on Van/MPV -EoB (upper row), -EoB-84mm (middle row) and -EoB with 30° vehicle rotation (lower row); top view.

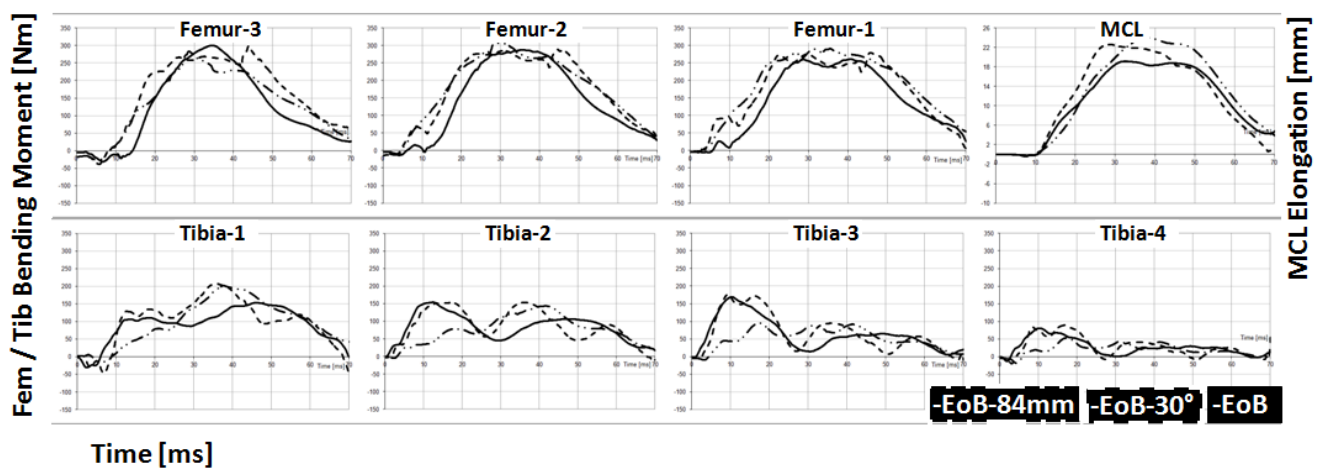


Fig. E-6. Qualitative correlation for FlexPLI-UBM – Van/MPV LHS with different setups.