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Validation of a biofidelic LS-DYNA model and comparison vs. a traditional ATD Finite Element model for assessing Knee-Thigh-Hip injuries

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Abstract: An enhanced Finite Element (FE) model of the human Knee-Thigh-Hip (KTH) is presented as a reliable mean for prediction of bone failure mechanism for a frontal knee impact. Improvements made to the original model are reported. The model is validated against a cadaver sled test. An identical finite element impact event was replicated with use of the LSTC dummy. The main terms of comparison were the axial femur force and the possibility to locate a potential bone fracture in the lower extremity. The KTH model gave more reliable results than the dummy model in terms of fracture prediction.

Keywords: KTH; knee-thigh-hip; FE; finite element; flesh modelling; impact crashworthiness; dummy modelling.

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1 Introduction

1.1 Real-world Knee-Thigh-Hip injuries in car crashes

Distribution of injuries in frontal car crashes has changed since the increased use of safety belts and airbags, which did offer better protection of the upper region of the body, but did not address prevention of injuries in the lower extremities. In the past years, studies have shown an increase in injury to the Knee-Thigh-Hip (KTH) region of the human body as consequence of car impacts, with common types of injuries including femur and pelvis fractures, and hip dislocations (Kuppa et al., 2001).

Several research projects have been conducted to better understand the distribution of lower extremities injuries during frontal impacts and they showed that these real-world crash injuries are a function of the geometry, loading, position and material properties of the KTH (Sochor et al., 2003; Yoganandan et al., 2001; Monma and Sugita, 2001).

Finite Element (FE) model is a potential tool to explore injury mechanisms to the KTH when subjected to frontal impact, similar to one that would be experienced in a frontal car crash. Understanding the mechanisms of these KTH failures could help in better and safer design of car interiors. The objective of this study was to enhance and validate an FE model of the KTH and to compare its results with respect to those obtained by use of a FE model of an Anthropomorphic Test Dummy (ATD) for the same impact event simulation.

1.2 Bone and flesh modelling

Several studies have been performed in past to accurately map bone thicknesses in lower extremities. The distribution of cortical bone in the proximal femur is believed to be critical in determining fracture resistance (Treece et al., 2009). Neto et al. (1999) examined 10 pairs of femurs and determined that cortical layer thickness progressively increased from the metaphysic to diaphysis. Thickness was determined for distinct sections. In a series of experiments on 16 cadaveric femurs, Treece et al. (2009) developed a technique that is capable of producing unbiased thickness measurements down to 0.3 mm in upper extremities of the femur. High resolution mapping techniques estimated accurate thickness in the femoral head and in the lesser trochanteric region. Using Computed Topography (CT), Bousson et al. (2000) examined 16 cadaveric male femurs in the age range of 70–99 years and determined an average cortical femur thickness of 4.474 ± 1.202 mm.

Existing FE models lack accurate material properties of soft tissues while representing lower extremities. During automotive impacts, the flesh and fat get compressed, absorb part of the impact energy, and transfer and distribute the rest of the energy to the skeleton (Untaroiu et al., 2005). Therefore, the accuracy of the viscoelastic mechanical properties of the FE flesh is of crucial importance in computing fracture criteria. Mechanical properties of flesh were modelled using Kelvin-Maxwell viscoelastic material model in LS-DYNA. Characteristic values used in this model were based on those reported by Chang et al. (2008).

1.3 Finite Element dummy modelling

In the years between 1950 and 1970, an ATD was developed by General Motors to study occupant response to vehicular collisions, and evaluate the effectiveness of vehicle restraint systems. The ATD was shown to accurately capture the response of an actual human to a crash event. Improvements in computer technology allowed engineers to use numerical simulations to predict occupant response to a collision. Previously, only costly and time consuming physical tests were used to accurately determine occupant response. Since the increase in popularity of numerical models to study motor vehicle safety, several models of the ATD have been developed (Teng et al., 2007).

Many of the numerical models of the ATD were developed for use with the LS-DYNA FE code. Teng et al. (2007) describes four ATD models, three of which were developed by Livermore Software Technology Corporation (LSTC) and one by First Technology Safety Systems (FTSS) and ARUP Associates. Table 1 shows basic information about different models.

Table 1 Finite Element dummy models

Model	*Number of nodes*	*Number of elements*	*Number of parts*	*Number of materials*
LSTC rigid dummy	6437	3963	113	109
LSTC deformable dummy	5731	5825	109	99
VPG deformable dummy	8512	5694	101	94
FT-ARUP deformable dummy	24,230	24,243	267	159

Source: Teng et al. (2007)

In Teng's et al. (2007), the four models were used to simulate the Quality Control (QC) tests used in the manufacturing of actual ATDs. In all of the QC simulations, all of the ATD models performed within the QC limits. The FT-ARUP model gave results that best agreed with experimental results in all tests. In all tests, the LSTC Rigid Dummy gave results that 'over-predicted' the ATD response (gave larger forces, deflections, etc.).

In addition to the QC test simulations, Teng et al. (2007) simulated each of the ATD models undergoing a sled test, and compared the accelerations given by the simulation to measured accelerations from a physical test. As with the QC tests, all of the ATD models showed good agreement with experimental results. The FT-ARUP model agreed most closely with the experiment, and the LSTC Rigid Dummy predicted the largest accelerations of all the ATD models.

Although the FT-ARUP model gave results that closely agreed with experimental results, it was not used in this study because it is much less computationally efficient than the other models, and it is only commercially available. The LSTC and VPG Deformable Dummies were also not used because they were either unavailable or lacked sufficient documentation. As a result, the LSTC Rigid Dummy model was used.

The LSTC Rigid Dummy comes with documentation explaining pre and post-processing methods (Guha et al., 2008). This document explains how to position the dummy (putting it in the correct place in space, rotating limbs, etc.) using the dummy positioning feature in LS-PrePost (LSTC, 2007a). It also explains how to post-process results, such as femur and tibia axial forces, tibia moments, etc.

There are, however, limitations on limb rotations when using the dummy (usually about ±20 degrees). Also, documentation at the start of the dummy input deck suggests further limitations to prevent numerical problems, e.g. the upper legs cannot be positively or negatively rotated in flexion more than six degrees to prevent initial penetrations with the pelvis (Guha et al., 2008).

1.4 Injury criteria

Researchers have developed injury criteria to determine risk of injury at a certain part of the body for an occupant. These criteria typically relate some measurable quantities (e.g. axial force, bending moment, displacement, etc.) to a risk of a certain level of AIS injury. Abbreviated Injury Scale (AIS) injury levels range from one to six, with a level of one corresponding to a minor injury, and a level of six corresponding to an injury the victim could not survive. For lower extremity injuries, one of the most widely used criteria is the femur axial force. The threshold limit for femur axial force to prevent injury, according to FMVSS 208 is 10 kN for the 50th percentile male (Sochor et al., 2003). This corresponds to a 35% chance of sustaining Abbreviated Injury Severity (AIS) 2+ injury (Kuppa et al., 2001). The 10 kN femur axial force is usually measured on an ATD with a load cell located in the femur.

Some researchers have suggested that other factors affect occupants risk to injury. Mertz (1993) proposed an injury criterion of 9070 N for the 50th percentile male, which is based on a model that considers both maximum femur axial force and the loading duration. Kuppa et al. (2001) studied the results reported by Morgan et al. (1990), and found that femur axial force alone is sufficient in predicting injury to the femur. Kuppa's analysis involved using logistic regression to analyse the results of 26 impact tests conducted by Morgan. Kuppa et al. (2001) was able to write equations for injury probability of KTH complex as a function of the recorded femur axial force.

2 Finite Element KTH modelling

The previous FE KTH model from Silvestri and Ray (2009) was considered. A few changes were made to this model and are reported in the next sections. A new model of the femur was developed with better node connectivity for force transmission, flesh representation was added in order to make the model more biofidelic and account for friction force with the seat, and torso modelling was considered for a realistic representation of mass of the upper body and to add seatbelt constraints to the model. Explanation regarding muscle and ligament representation included in this model can be found in Silvestri and Ray (2009).

2.1 Femur cortical bone

One of the first issues that arose when considering the previous KTH model was the mesh of the femur cortical bone. In fact, there was no node connectivity between knee femoral condyle and bone femoral shaft. The problem was faced by considering a tied contact between these parts, to allow for transmission of forces through nodes at the interface. A new and better representation of the femoral cortical mesh would guarantee, however, more accurate transmission of forces from the knee through the femur bone after impact occurs. Thus, a new cortical femur model was developed, in order to assure node-to-node connectivity at these interfaces. Comparison between the previous and the new femur mesh model is shown in Figure 1.

Moreover, femur bone thickness change was taken into account during the process: three layers of eight-node solid brick elements were used to model the femur (Figure 2) and their thickness was accordingly modified to replicate the total thickness of different femur sections (Figure 3). Total cortical thickness varied from 2.03 mm to 4.08 mm through the femur bone. The material model was unchanged: card MAT_59 (*MAT_COMPOSITE_FAILURE_SOLID_MODEL) from LS-DYNA code was considered for replicating the orthotropic properties of the cortical femur bone, with a density of 1900 kg/m^3 and a Young's Modulus of 11,500, 17,000, and 11,500 Mpa for the longitudinal, transverse and normal directions respectively (Silvestri and Ray, 2009).

Figure 1 (a) Comparison of previous femur mesh and (b) new cortical bone model with node-to-node connectivity

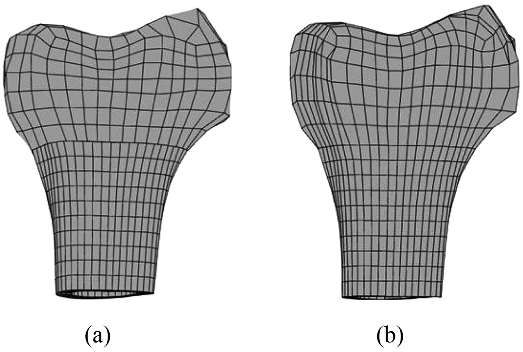


Figure 2 Three layer solid modelling of the FE cortical femur

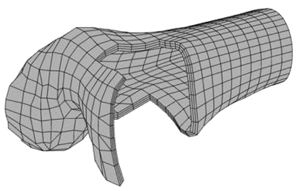
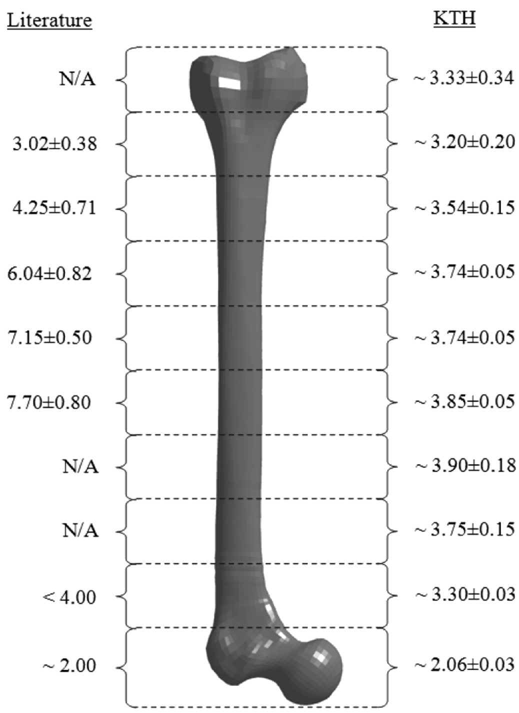


Figure 3 Cortical thickness (expressed in mm) modelling for the FE cortical femur



Source: Neto et al. (1999) and Treece et al. (2009)

The new KTH model now accounts for 11,708 elements and 15,568 nodes.

2.2 Thigh, buttocks and torso flesh

Previous FE model for the KTH did not include a three dimensional representation of the human flesh, but accounted only for its inertia by considering lumped masses added to the external nodes of the bone geometry. This resulted in a stiffer response of the axial femur force when compared to the one obtained from cadaver sled test. The flesh of the thigh was modelled with solid elements and was represented with LS-DYNA *MAT_KELVIN-MAXWELL_VISCOELASTIC material model. Flesh elements were connected to the bones with a 'tied node' contact card. This would allow for transmission of femur force into the flesh and simulate more realistic results.

*CONSTRAINED_NODE_SET option was used to achieve realistic behaviour of the flesh at the time of impact and avoid any numerical errors such as element negative volumes. Using this card, every node is constrained with every other node in the set; hence forcing the entire set to move equivalently in global *x*, *y* and *z* axes. Representation of flesh provides a larger contact area at the interface between the thigh and the seat. Hence, it captured frictional force between the lower extremity and the car seat, allowing for a larger surface contact with respect to the one obtained with bone modelling only. It also takes into account mass and inertia distribution around the femur bone. Representation of the FE thigh flesh and material properties used for the model are reported in Figure 4 and Tables 2(a) and 2(b).

Figure 4 FE thigh (left) and buttocks-torso (right) flesh representation

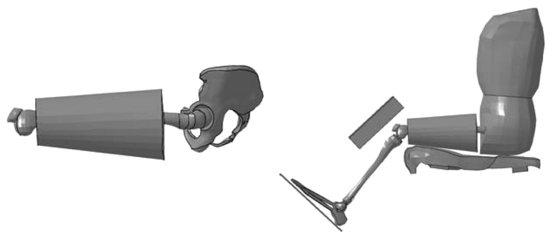


Table 2(a) Material properties for FE buttocks-torso model

	Torso	*Buttock*
Mass density	2.7650E-08	8.7000E-09
Elastic modulus	2.0000E+05	2.0000E+05
Poisson's ratio	0.495	0.495

Table 2(b) Material properties for FE thigh model

	Thigh
Mass density	2.8892E-09
Bulk modulus (Elastic)	0.25
Short time shear modulus	0.069
Long time shear modulus	0.155
Maxwell's decay constant	1000

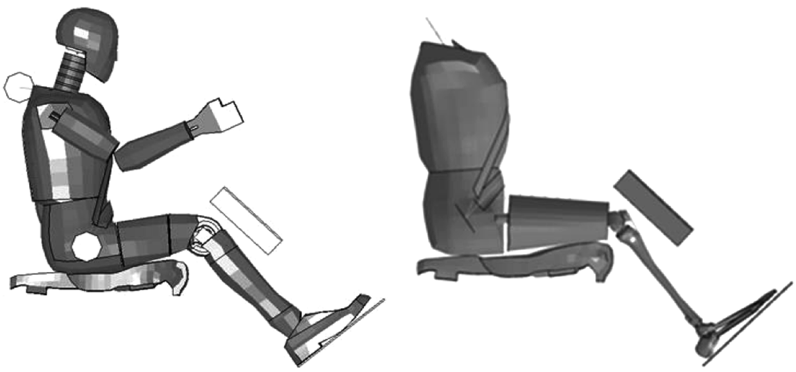
Geometry for the FE buttocks and torso parts was obtained from the LS-DYNA 50th percentile male LSTC dummy in the form of .iges curves and surfaces. External surface of the buttocks was modelled with solid elements and served to numerically capture the friction between the lower extremity and the seat model. The buttocks flesh was connected and constrained to the pelvic movements and merged to the representation of a simplified human torso (Figure 4). Flesh for the torso and buttocks was modelled with simple elastic material properties (Tables 2(a) and (b)) since the purpose of this research was not to analyse its properties but to ensure realistic behaviour of the KTH while considering upper body inertia and the possibility of adding seatbelt restraints.

3 Method

The FE KTH and LSTC dummy models were both simulated under similar conditions to a physical sled test on a cadaveric specimen (Rupp, 2002). In both simulations, the KTH and the dummy were positioned on a seat by applying the global damping feature, which globally defines mass weighted nodal damping to nodes of deformable bodies and mass centre of rigid bodies (LSTC, 2007b). This would allow for a more realistic initial surface contact and friction level between the KTH (or dummy) and the interior of the car (seat) prior to impact.

Both the KTH and the dummy models were positioned with initial zero-degree angles of thigh adduction and flexion to replicate the initial position of the cadaver reported for the sled test. Lateral views of the setup for the FE simulations are reported and compared in Figure 5.

Figure 5 Lateral view: (a) FE LSTC dummy and (b) FE KTH model



(a) (b)

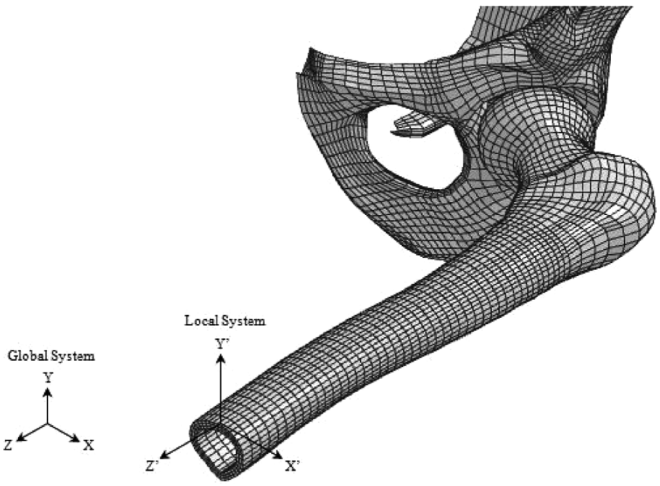
In the FE simulations, the upper bodies of the KTH model and the LSTC dummy were restrained with application of seatbelts. The initial distance between the lower extremity and the knee-bolster was set to be 38 mm, after carefully analysing pictures from the report of the cadaver sled test. After the application of global damping, initial distance from dashboard and flexion angles were slightly different. It is believed that these minor changes would not affect any end results.

In the FE simulations, an initial velocity of 13.41 m/s (50 km/h) was applied to all nodes. Deceleration curve obtained from physical test was imposed on the sled parts in the FE simulation to initialise the impact sequence.

As axial femur force was the key parameter analysed by NHTSA to detect possible KTH fractures during frontal car crash, it was also recorded from KTH and Dummy simulations and compared to that one obtained from the cadaver sled test (Kuppa et al., 2001). This evaluation formed the basis of KTH fracture prediction and a means to compare reliability of KTH vs. LSTC Dummy.

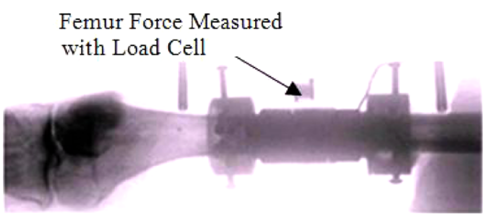
In the KTH simulation, axial femur force was obtained by use of the *DATABASE_CROSS_SECTION_SET card, which allows for defining a cross section for resultant forces written to ASCII file SECFORCE. A local coordinate system was necessary to obtain local forces from the cross-section in the mid-shaft of the FE model (Figure 6).

Figure 6 Local coordinate system used to detect femur local axial force from a cross-section of the cortical femur bone



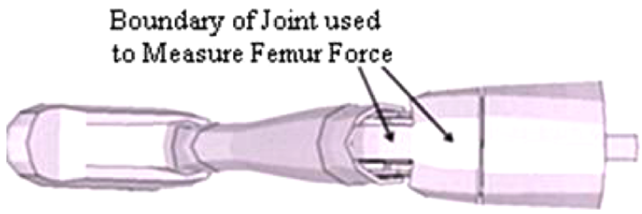
In the simulation with the LSTC dummy, axial force was measured using the *CONSTRAINED_JOINT_LOCKING keyword as recommended in the FE LSTC dummy model's documentation (Figure 7) (Rupp, 2002). This keyword allows users to define a joint between two rigid bodies, and in this case allows users to measure forces in rigid bodies at this interface.

Figure 7 (a) Femoral axial force locations considered for the cadaver test; (b) FE LSTC and (c) FE KTH (see online version for colours)



(a)

Figure 7 (a) Femoral axial force locations considered for the cadaver test; (b) FE LSTC and (c) FE KTH (see online version for colours) (continued)



(b)

(c)

Source: Rupp (2002)

4 Results

The LSDYNA solver, version 971 was used for all simulations (LSTC, 2007b). The computers used for the simulations are "Sunfire X2200M series" with two dual core AMD Opteron 2220 2.8 GHZ CPUs and 12 GB of RAM. Integration time-step set for KTH and LSTC Dummy simulations was 4.50E-07 and 2.35E-07 s respectively. Total simulation time varied depending on these factors.

Behaviour of the FE LSTC dummy and the KTH during the time of impact were compared to the cadaver sled test outcomes. Comparisons were made also with the same event simulation, but in case of the FE KTH without flesh modelling. As presented in the next sections, comparison was made with respect to the axial femur force output and the bone fracture result, when possible.

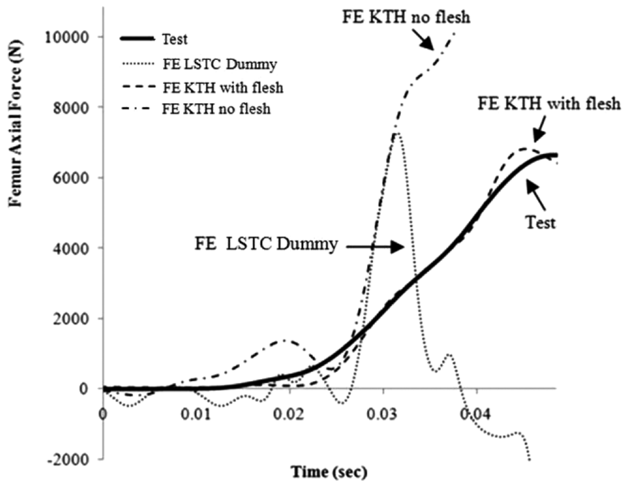
4.1 Femur axial force comparison

Axial femur forces from the KTH and the LSTC dummy simulations were compared to those obtained from physical experiment (Figure 8). Note that positive value, in this case, indicates the compressive force.

The curve plot was trimmed at the time of maximum impact force. In fact, bone fracture in the femur cadaver and the FE KTH model already occurred at this time.

The maximum femur force recorded in the experiment was 6648 N. The simulated maximum force obtained from the left femur of the LSTC dummy was 7810 N, while the force resulted from the KTH model with flesh was 6818 N and without flesh was 10287 N.

Figure 8 Comparison of the femoral axial force from the test and the FE simulations



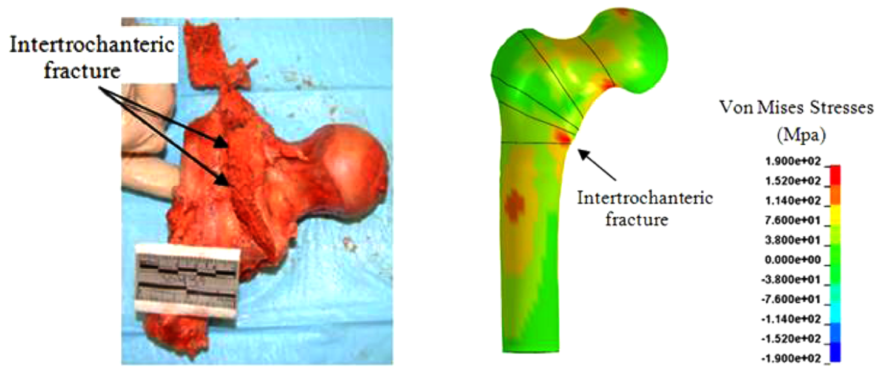
Source: Rupp (2002)

4.2 Bone fracture comparison

Outcomes of the simulations with the sled test results in term of bone fracture mechanisms consequent to the impact. Von Mises stresses were plotted during the impact event and bone fracture was considered to occur when the value of those stresses reached ultimate strength of the cortical bone (Silvestri and Ray, 2009).

As a result of the impact, the left femur of the cadaver in the physical experiment reported an intertrochanteric fracture. The same type of fracture was obtained in the simulation with the FE KTH model. Figure 9 compare the left femur fracture of the cadaver with a plot of the Von Mises stresses of the left femur of the KTH model. Von Mises stresses reach the fracture value of 190 Mpa for longitudinal compression direction in the intertrochanteric region of the femoral head, matching the results obtained from the sled test. Fracture of the femur in the FE KTH model occurred before the time of the maximum impact.

Figure 9 Comparison of bone fracture mechanism obtained in the physical test (a) and in the FE KTH (b) simulation (see online version for colours)



(a) (b)

For the FE LSTC dummy simulation, Von Mises stresses could not be obtained from the elements of the femur, since the bone was modelled as a rigid material. Thus, there was no possibility to visually locate any bone fracture for the lower extremity of the FE dummy. From literature, the only option to predict a fracture in the lower extremity of the LSTC dummy is to evaluate the peak femur force, which resulted in 7810 N as reported in Section 4.1, and compare it to the femur force threshold defined by NHTSA. Even in this case, however, it would not be possible to locate the fracture along the bone geometry.

5 Discussion

From the axial femur force result (Figure 8), it is evident that the FE KTH model (with flesh modelling) is capable to replicate the curve obtained with the cadaver sled test, in the condition of neutral position prior to impact. The two curves from the test and the FE KTH simulation have a very similar slope, except for a small difference at the beginning of the impact, where inertial effects might have played a considerable role. The slopes of the curves from the FE LSTC and the FE KTH without flesh modelling appear to be very stiff with respect to the sled test one. This is believed to be related to the material modelling used for the dummy, which is a rigid material type, and to the fact that a damping factor caused by the flesh contact with the seat was missing in the KTH model without flesh modelling. Previous FE representation for the KTH did not include the human flesh model, but accounted for its inertia by considering lumped masses added to the external nodes of the bone geometry. This resulted in a stiffer response of the axial femur force as compared to the one obtained from cadaver sled test. With this study, the flesh was modelled with 3D elements and viscoelastic material properties. Since the flesh was connected to the femur with tied-node contact definition with a perfect node-to-node connectivity, it damps the forces and transfers them, or rather diffuses them, to constrained nodes in the flesh representation. Moreover, the maximum femur force of the FE KTH with consideration of flesh modelling is very close to the maximum value from the physical test, meaning that the two dynamics are very similar for the all duration of the event. The peak femur force from the FE LSTC simulation, on the contrary, is much higher than the test one and even the unloading region of the curve is very different from the one recorded in the cadaver experiment, leading to the conclusion that the dummy was not able to replicate the event in an acceptable way.

In both the experiment and the FE KTH simulation, same results about bone failure mechanisms were obtained, with the result of an intertrochanteric fracture in the femoral head. In the sled test, this fracture is considered to have occurred at the maximum femur force, since no special means were use to acquire the timing of any possible bone failure and correlate it to the femur axial force recorded. Similar consideration needs to be done for the FE LSTC dummy simulation, since in this case there cannot be any evidence of a material bone failure, being the material rigid. For this case, a comparison between the FE femur axial force value recorded and an injury criteria (such as the threshold proposed by NHTSA) is the only approach for defining bone fracture.

On the other side, with the FE KTH simulation it was possible to obtain the exact time of bone fracture, studying the values of the Von Mises stresses. This result can be used to better investigate human lower extremities injury criteria and for further development of a more correct injury threshold.

6 Conclusion and further work

This research aimed at enhancing a previous FE model of the human KTH available for studying fracture mechanisms consequent to frontal knee impacts. The mesh of the cortical femur was changed in order to improve the impact force transmission along the bone and to account for cortical thickness gradient change in the condylar and head region. A thigh flesh model and a basic replication of the geometry for the buttock and the torso was also added to consider realistic friction contacts between the lower extremity and the seat and to account for flesh and fat correct mass distribution. Global damping was applied to the model for allowing relaxation of the lower extremities on the seat cushion. Moreover, seatbelts were modelled to realistically constrain the motion of the model during the impact process. This enhanced model was successfully validated against results from a cadaver sled test, considering femur axial force and bone fracture outcomes.

Another FE model (the LSTC dummy model) was considered for the same impact event, with the same position and restraint conditions. Axial femur force response from the FE LSTC dummy was too stiff and too high in peak with respect to the one recorded from the sled test. This was mainly attributed to the rigid material model used for the FE dummy definition. Also, possible dummy bone fracture was related to the femur force peak value which had to be compared to the injury threshold criteria reported in literature. Using this method to study injury does not provide a way to precisely locate the point of fracture in the KTH complex.

These comparisons lead to the conclusion that the FE KTH model is suggested as a validated reliable mean for predicting lower extremity bone failure mechanisms in frontal impacts, capable to identify the fracture location. Further research could be addressed to improve the flesh modelling to extend it to the knee and femoral head regions of the leg. Works have started to validate the current FE KTH model for non-frontal axial impacts, to further investigate its capability to predict hip dislocation phenomena.

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