Improvements and Validation of an Existing LS-DYNA Model of the Knee-Thigh-Hip of a 50th Percentile Male Including Muscles and Ligaments

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Summary:

A detailed review of an existing LSDYNA finite element (FE) model of the Knee-Thigh-Hip (KTH) of a 50th percentile male was accomplished. The main scope was to refine some aspects of the model for obtaining a more appropriate and biofidelic tool for injury mechanics investigation of the KTH in frontal car crashes. Detailed reviews of this model were performed with regards to material properties of the bone models used for representation of the pelvis, femur and patella. To investigate bone fracture mechanisms due to impact, the erosion material failure method was abandoned in favor of the adoption of a more realistic detection of failure locations using stress contour plots. Qualitative validations of the pelvis and femur bones of the new model were performed against cadaveric specimen tests conducted at University of Michigan Transportation Research Center. In addition, quantitative validations were performed with use of the Roadside Safety Verification and Validation Program (RSVVP), developed to validate numerical models in roadside safety. The approach for these validations was also different. Earlier work had compared the finite element results to the physical test corridors whereas this work used a direct comparison of each finite element validation simulation to a specific corresponding test. Validation of the bone models were based on comparison of the impact forces from contact between the dashboard and knee region of the KTH model. For each case, force simulation results were in good agreement with experiment outcomes, and FE fracture locations matched failure modes from cadaveric tests. Quantitative results indicate that the test and FE time histories can be considered to be the same, and they therefore represent the same impact event. A new validated dynamic representation of ligaments was adopted for prediction of avulsion ligament injuries in high speed frontal automotive collisions when lower extremities are subjected to high strain rates. FE results from ligament avulsion agreed with test data and injury criteria recommended from literature. A different model of the knee patellar tendon was implemented with use of material SEATBELT and the introduction of slip-rings to constrain the patellar tendon to the biomechanically correct line of action. This refined LSDYNA finite element model of the KTH resulted in a more biofidelic representation of the human KTH and represents a suitable and reliable tool for exploration of KTH fracture mechanisms resulting from frontal vehicle crashes.

Keywords: KTH, Finite Element Modeling, Bone Validation, Ligament Validation, Patellar Tendon.

1 Introduction

Recent studies have shown that the distribution and severity of injuries in frontal car crashes has significantly changed in recent years [1, 2, 3, 4]. Kuppa et al. [1] found that about half of the Abbreviated Injury Scale (AIS) level two or greater lower extremity injuries occurred in the knee-thigh-hip (KTH) region in the 1993 through 1999 NASS CSS data. A better understanding of the mechanisms of KTH failures would help in the process of designing vehicle safety systems and interiors which would minimize lower extremity injuries.

A validated LSDYNA FE model of the KTH was developed by Silvestri and Ray [5] to parametrically explore failure mechanisms of the KTH according to variations in seating posture and loading conditions in frontal car crashes. This research aimed to improve certain modeling aspects of the KTH FE model from Silvestri and Ray [5].

Erosion material method was abandoned to detect fracture in the pelvis, femur and patella bones. Instead, bone stress contour plots were used to define failure mechanisms due to impact.

Formerly, only a qualitative bone validation procedure was used for the original KTH model where the response was qualitatively assessed with respect to its position in the physical experimental corridors [5]. In this research, a quantitative validation approach was used as suggested by Ray et al. [6]. The three main revised model components (i.e., condyle, femur and pelvis) were qualitatively validated by comparing the fracture locations in the FE simulations with those obtained from the physical tests. In addition, a quantitative validation assessment was performed by considering five different comparison metrics that assess the agreement between the force time histories of the FE results and the physical tests [7].

An FE ligament representation including dynamic failure properties was also included in the original FE KTH model from Silvestri and Ray [5]. In fact, the previous ligament model did not account for strain sensitivity properties which could be very important in lower extremity front crash scenarios which normal occur at moderate strain rates. Peck et al. [8] developed a dynamic failure model of ligaments for use in FE simulations of high speed automotive crashes. This model was incorporated in the FE model by Silvestri et al. [9].

Another modification of the original FE KTH model concerned the representation of the patellar tendon which is now modeled with use of material SEATBELT and sliprings to replicate the physiological line of action of the tendon. The revised KTH model is shown in Figure 1.

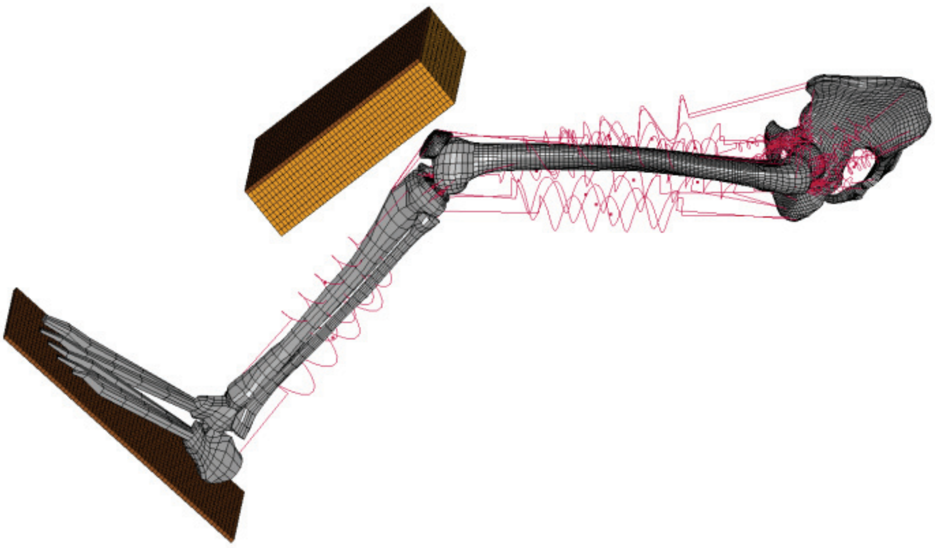

Figure 1: Revised Knee-Thigh-Hip model.

2 Methods and Results

2.1 Bone Modeling

In the original KTH model from Silvestri and Ray [5] both cortical and trabecular parts of the femur were represented with an LSDYNA composite material model and failure was detected using element erosion. As found in literature, however, cancellous bone stiffness and strength are two orders of magnitude lower than in cortical bone [10]. As a consequence, in this improved KTH representation, trabecular bone was not used in the model since it contributes little to the overall stiffness or strength. Others have used this approach in biomechanical crash models [10].

A density of 1.22e-6 Kg/mm^2 and modulus of elasticity of 17,000 Mpa was used for modeling the pelvis bone. The patella was modeled with the same material properties as the cortical pelvis in the revised model. The original model assumed the patella was rigid.

Element erosion is one common way of representing failure in LSDYNA. When the failure condition is reached, the element stiffness is set to zero such that the element can no longer support load. While there are advantages to this method there are also some disadvantages, especially with respect to biological materials. The erosion process removes mass which is non-physical although the error is small if the element size is small. More important in biological materials, element erosion can suggest fracture patterns that are not necessarily correct. When a biological material fails, the fracture propagation can be quite complicated since the materials are orthotropic and non-homogeneous. In biomechanical applications, identifying the location of the failure is generally the most important aspect and this can be done more reliably by observing the location of the maximum stress contours. The original model of the KTH used element erosion to represent failure but the revised version identified fail locations simply by using the stress contour of the failure stress. Ultimate stress values considered for detecting fractures in the bone materials are reported in Table 1.

Table 1: Yield stress values considered for fracture of bone material.

Bone	Description	Yield Stress (Mpa)
Femur	In plane shear	60
	Transverse shear	60
	Transverse shear	60
	Longitudinal compressive	130
	Transverse compressive	190
	Normal compressive	130
	Longitudinal tensile	50
	Transverse tensile	50
	Normal tensile	50
Pelvis		157

The three FE component validations (i.e., pelvis, femur and knee condyle) were re-run with the revised and improved KTH model. The use of force corridors from the experimental test was no longer used to validate the FE models. Instead, the FE results were compared to one specific component test. This was done because the experimental tests for the same component that make up the corridors included substantial differences with respect to the experiment setup, ram applied displacement, impactor shape and material, as well as gender, age and health conditions of the cadaver from which the specimens were taken. The new approach involved choosing one representative experiment for each component validation and replicating that specific test setup and input conditions as exactly as possible for comparisons between the physical experiment and the FE results. The impact force at the contact between the patella bone and the impactor foam was quantity used to compare the knee force from experiments and the FE results.

Two approaches were considered for validation of the component models. First, a qualitative approach where the curves obtained from the FE simulations were visually compared to those observed in the tests (Figure 2). Second, a quantitative validation approach was used to compare the curves where comparison metrics such as the Sprague and Geers (S&G) MPC metrics [7] and the analysis of variance (ANOVA) metrics were calculated [11].

Differences in magnitude and phase between the experimental and simulation curves were evaluated respectively by the M and P components of the Sprauge and Geers metrics. A combined score including affects of both magnitude and phase was then calculated as the C component of the Sprauge and Geers for an overall evaluation of the curve agreement. For each of the MPC metrics, zero indicates a perfect match (i.e., the two curves are identical).

The ANOVA metrics are standard statistical tests by which is possible to assess whether the variance of the residuals between two curves can be attributed to random error only. In other words, any differences between two curves representing the same event must be attributable only to random experimental or numerical errors which cause the mean residual error (and the corresponding standard deviation) not to be exactly equal to zero. The smaller are these values, the better is the match between the experimental and numerical curves. In this work, the ANOVA metrics were evaluated on the residuals normalized to the peak value of the experimental curve.

A program called the Roadside Safety Verification and Validation Program (RSVVP) was developed to validate numerical models in roadside safety [6]. RSVVP was used in this research to compute the comparison metrics for a quantitative validation of the KTH FE model. For each validation case, the curves were filtered and synchronized by minimizing the absolute area of the residuals before the computation of the metrics. Table 2 shows the values of the metrics from the comparison of the test and numerical curves for each of the three validation cases. Generally, MPC metric values under 10 are considered quite good. An average standard error and standard deviation of the error in the ANOVA metrics less than 10 is likewise considered very good.

Figure 2: Qualitative validation results for the pelvis (a), femoral head (b), and condyle (c) components.

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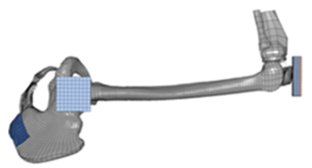
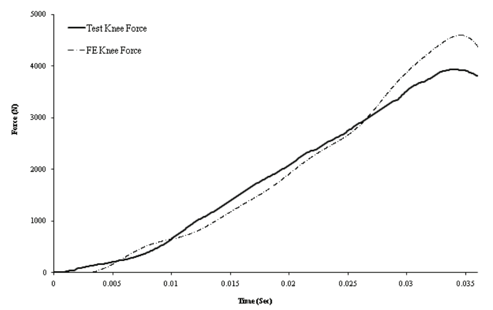
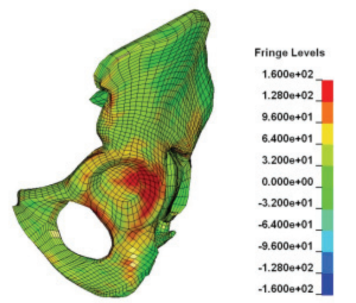
</td><td>
</td><td>
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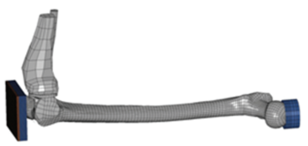
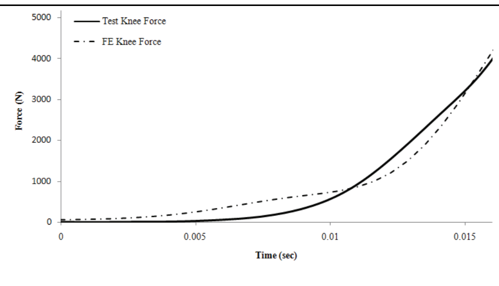
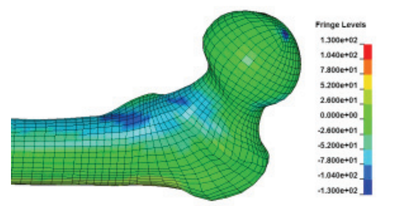
</td><td>
</td><td></td></tr>
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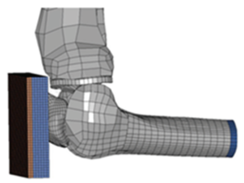
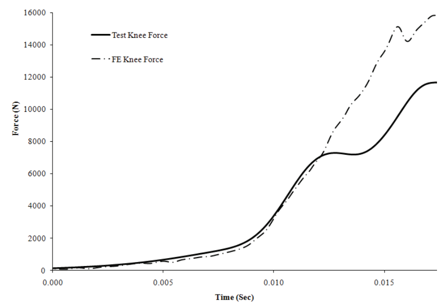
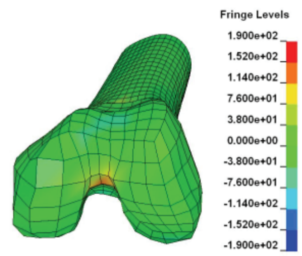
</td><td>
</td><td>
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Table 2: S&G MPC and ANOVA metrics values [12].

MPC Metrics	**Pelvis**	**Femur**	**Condyle**
Sprague-Geers Magnitude	9.2	2.2	-4.7
Sprague-Geers Phase	2.5	5.6	9.9
Sprague-Geers Comprehensive	9.6	6	10.9
ANOVA metrics			
Average	0.03	0.03	-0.07
Std	0.07	0.05	0.13

2.1.1 Pelvis Validation

In the pelvis FE simulation (i.e., Figure 2 (a)), at time 0.034 sec (i.e., time at which the peak force occurred in the test), contour von Mises stresses were plotted for the cortical pelvis bone. The results show that at the time when the peak knee contact force occurred in the test, the Von Mises stresses in the acetabular cup region of the pelvis reach the critical ultimate failure stress of 160 MPa. This indicates that failure of the pelvis would start in the acetabular cup region of the bone at the time of maximal impact force applied. Since an acetabular fracture of the pelvis was also obtained in the experimental test, these results show that the FE pelvis model replicates the failure mode observed in the test.

The quantitative metrics for the comparison of the knee-bolster time histories in the pelvis impact experiment are presented in the first column of Table 2. The magnitude and phase components of the Sprauge-Geers metrics were both under 10 percent which is considered very good. The combined Sprauge-Geers metric was just over 10 but is also considered a good comparison. The average error between the signals as measured in the ANOVA metrics was less than 10 percent and the standard deviation of the error was 13 percent which are both good. Since the FE knee contact force time history results in good quantitative comparison scores using both the Sprauge-Geers MPC and the ANOVA metrics and the location of the ultimate stress coincides with the experimentally observed acetabular fracture, the FE pelvis model can be considered validated.

2.1.2 Femur Validation

In the femur FE simulation (i.e., Figure 2 (b)), at time 0.01719 sec (i.e., the time at which the peak knee bolster force occurred in the test), contours of the longitudinal compressive Z-stresses were plotted for the cortical femur. The results show that at the time when the peak knee contact force occurred in the test, the ultimate Z-direction stress of 130 Mpa is reached in the intertrochanteric region of the femur. The FE results indicate that failure of the femur would start in the intertrochanteric part of the head at the time of maximum impact force at the knee bolster. Since intertrochanteric fracture of the femur was observed in the experimental test, the FE femur model accurately replicates the failure mode of the test. The knee bolster contact force is also plotted for both the test and the FE result in Figure 2b. Since the shape of experimental and FE knee bolster contact force time histories are qualitatively similar, the FE femur model can be considered validated.

The quantitative metrics for the comparison of the knee-bolster time histories for the femur impact are presented in the second column of Table 2. The magnitude, phase and combined components of the Sprauge-Geers metrics are all well under 10 percent which is considered very good. The average error between the signals as measured in the ANOVA metrics was three percent and the standard deviation of the error was five percent which is exceptionally good. Since the FE knee contact force time history for the femur impact results in good quantitative comparison scores using both the Sprauge-Geers MPC and the ANOVA metrics and the location of the ultimate stress coincides with the experimentally observed intertrochanteric fracture, the FE femur model can be considered validated.

2.1.3 Condyle Validation

The longitudinal compressive Z-direction stress contour was plotted for various time steps and the ultimate stress of 190 Mpa was observed at time 0.0122 sec, approximately the same time as the maximum knee-bolster contact force." The FE results indicate that failure would start in the intercondylar region of the bone at the time of maximal impact force applied. Since intercondylar knee fracture was obtained in the experimental test, these results show that the FE knee condyle model replicates adequately the failure mode of the test.

The quantitative metrics for the comparison of the knee-bolster time histories for the condyle impact are presented in the third column of Table 2. The magnitude, phase and combined components of the Sprauge-Geers metrics are all under 10 percent which is considered very good. The average error between the signals as measured in the ANOVA metrics was three percent and the standard deviation of the error was seven percent which is exceptionally good. Since the FE knee contact force time history for the condyle impact results in good quantitative comparison scores using both the Sprauge-Geers MPC and the ANOVA metrics and the location of the ultimate stress coincides with the experimentally observed fracture, the FE condyle model can be considered validated.

The comprehensive Sprauge-Geers metric was below 15 percent for all the validation cases and, in particular, the magnitude and phase metrics always were below 10%. Both the average and the standard deviation of the residuals in the ANOVA metrics were always below 10 percent of the peak value of the respective experimental curve. These quantitative results indicate that the experimental and FE time histories can be considered to be the same and they therefore represent the same impact event.

2.2 Ligament Modeling

In the original KTH model developed by Silvestri and Ray [5], the ligaments were modeled as discrete elements with nonlinear material properties. This model, however, did not consider the effects of strain rate on material behavior and failure representative of the strain rates experienced during car crashes. Peck [8] developed a dynamic failure prediction model for human ligaments for use in automotive collisions. In Peck's model the failure load of a ligament is related to its geometry and the strain rate at which it is loaded as shown in Equation 1.

$$P_F = 3.0194 \cdot \varepsilon_{rate} + 1{,}091.6 \cdot \frac{A_0}{L_0} \tag{1}$$

where P_F represents the failure load, ε_{rate} the strain rate applied, and A_0 and L_0 are the ligament initial cross sectional area and initial length, respectively.

Based on this dynamic failure representation, an FE model of the knee human ligaments was developed by Silvestri et al [9]: ligaments are represented as non-linear elastic spring material type in LSDYNA and strain rate effects are considered through a velocity dependent scale factor. This FE dynamic failure representation of ligaments was validated against experiments conducted by Viano [13]. A reduced FE model of the human lower extremity was defined with only inclusion of femoral and tibial bones to conform to the physical tests. Anterior, lateral, medial and posterior cruciate ligaments (i.e., ACL, LCL, MCL and PCL) were also modeled [9]. The bottom mid-tibial shaft was impacted in the axial direction of the femur with a 1.8 m/sec constant velocity.

Both numerical and experimental results showed that an initial failure of the PCL occurs at a relative tibial-femoral displacement of about 14 mm while an ultimate collapse of the PCL occurred at a relative tibial-femoral displacement of approximately 22 mm. The simulation results fit nicely with both the test data from Viano [13] and the injury criteria recommended by Mertz [14]. According to Anderson, the maximum strain a ligament can tolerate before failure is between 9 and 18 percent [15]. In the FE simulation, the two discrete elements failed at strains of 14.5 and 15.5 percent respectively, which are within the range reported by both Viano and Mertz.

This dynamic ligament model was applied to the ligaments of the knee (i.e., Anterior Cruciate Ligament (ACL), Posterior Cruciate Ligament (PCL), Lateral Collateral Ligament (LCL) and Medial Collateral Ligament (MCL)) and to those of the pelvic joint (i.e., Iliofemoral, Ischiofemoral and Pubofemoral).

2.3 Patellar Tendon Modeling

In the original KTH model, the patellar tendon was modeled as shell elements that passed over the patella [5]. This arrangement allowed the patella too much freedom of movement and allowed the patella to “lock” the knee joint at times. A method had to be found to constrain the patella and patellar tendon to a more realistic range of motion. The patellar tendon was modeled with spring *MAT_SEATBELT.

Three lines of springs were chosen to increase the stability of the movement and avoid unphysical rotation of the patella during joint movements such as extension of the knee (see Figure 3 (a)). The patellar tendon originates at the rectus femoris muscle and is inserted in three different aligned nodes in the tibia. The spring element size was set to 6 mm. The offset between seatbelts and contact elements was selected as one mm. Each spring is forced to pass through a slipring on attached to the patella. The sliprings are positioned on nodes of the patella to permit the patellar tendon to slide on the bone during movements as it happens anatomically (Figure 3 (b)). The tendon, therefore, keeps a physiologically correct line of action while the muscles are contracting.

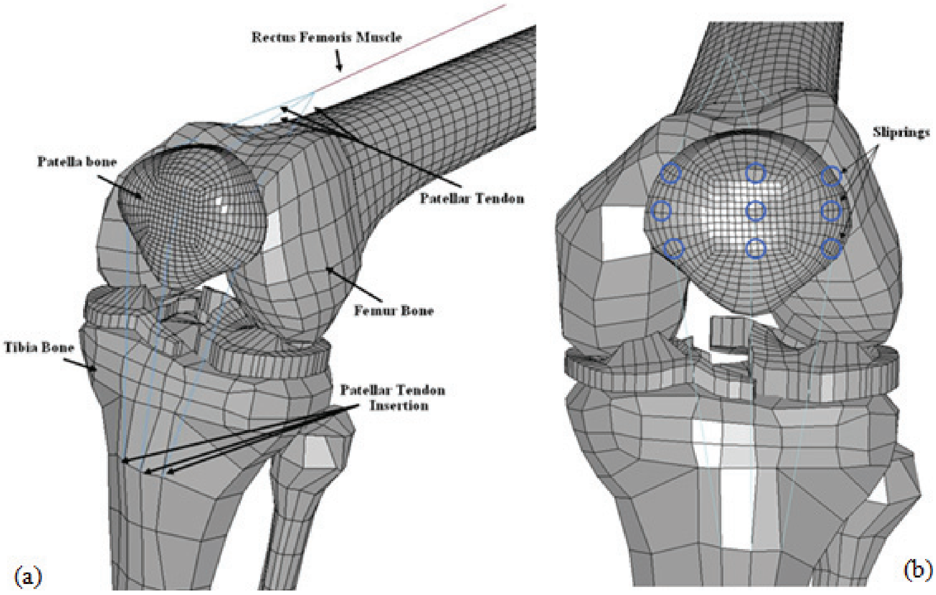

Figure 3: Patellar tendon modeled with spring seatbelt material proposed in LSDYNA (a) and position of sliprings for the three spring-lines on the patella bone (b).

The mass per unit length of the springs was defined to be 1.90e-03 Kg/mm^2. A minimum length of 0.05 mm is considered as input for controlling the shortest length allowed in any element and determining when an element passes through sliprings or is absorbed into the retractors. Normally, according to the LSDYNA manual, one tenth of a typical initial element length is usually a good choice [16]. A load curve (i.e., force vs. engineering strain) for loading is also inserted. To define the curve, the same method used to define curves for ligaments was followed. Based on physiology, an area of 163 mm^2 and an initial length of 156 mm for the patellar tendon were considered for definition of the failure load and strain values.

3 Conclusions

The paper has described improvements to the FE LSDYNA model of the KTH of a 50th percentile male originally developed by Silvestri and Ray [5]. A direct comparison of the knee impact force between the FE and the test outcomes was considered for each component validation. Results from simulations were always in good agreement with experiment findings, matching also fracture location in each case.

A validated dynamic failure prediction ligament model was inserted in the FE representation as a potential tool for detecting dynamic failure mechanisms of ligaments and avulsion ruptures. Future work should be directed to strengthen this FE model to account for mid-ligament failures.

The seatbelt material model was used to represent the patellar tendon and allowed for a more realistic description of the knee joint dynamics and the constraints between the patella bone and the patellar tendon. The improved LSDYNA model described in this paper provides an improved model for investigating fracture mechanisms in frontal car crashes.

4 Acknowledgments

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