VALIDATION OF LOWER EXTREMITY MODEL IN THUMS

A Chawla, S Mukherjee, D Mohan and Parihar A
Department of Mechanical Engineering, Indian Institute of Technology, New Delhi 110016 INDIA.

ABSTRACT

Three sets of test results on cadaver knee have been reported in literature. The existing legforms, dummies and mathematical models do not validate under all these three test conditions. A computational model such as THUMS is expected to validate under multiple boundary conditions. Evaluation of the FE model of knee in THUMS against multiple test conditions has been conducted. The FE model in THUMS has been reoriented to obtain the configuration used to evaluate knee impacts to pedestrians. The model has then been validated against the test conditions used by Kajzer et al. Strain limits are used to evaluate the injury process of knee in the simulation. The injury patterns predicted in simulations confirm with autopsy results. Response of the knee FE model has also been analyzed for tests conducted by Kerrigan et al .

KEYWORDS: Knee, Finite Element Modeling, Validation

LARGE PROPORTIONS of AIS 2+ injuries that occur in vehicle-pedestrian collisions are to the lower extremities. Atkinson etal, 2000 mention that knee injuries alone, consistently account for more than 10% of the serious injuries every year in the US while the injuries to the lower extremity as a whole accounts for more than a quarter of the injuries sustained in road accidents. Lower extremity injuries account for 39% of non-fatal vehicle pedestrian accidents in Japan (Matsui 2001).

Atkinson et al (2000) reports that most knee injuries occur in age group 15 to 40. Serious ligament damage rated as AIS 2 occur at lower velocities, below 20 Kmph, while fractures are observed when impact occurs at higher velocities. The International harmonized research activities data shows that sources of injury to the lower extremities are 67.1% by front bumper, 12.1% by bonnet leading edge, and 7.6 % by front panel when injuries were caused by vehicle structures (Atkinson et al 2000). To reduce the incidences and extent of knee injury, it is essential to understand the mechanics of knee injury for lower limb impacts.

Kajzer et al (1997, 1999) performed tests on knee using cadavers at two impact velocities, and indicated that knee injury types differ with the shear and bending load on the knee joint. Kerrigan et al (2003) examined the failure of knee joint components in shear and bending. Finite element models have been developed to investigate knee injury criteria. (Nagasaka et al 2003, Maeno et al 2001, Takahashi et al 2003)

A model of human knee has been developed by the Honda group and has been validated against some of the results obtained by Kerrigan et al (2003) (Takahashi et al 2003). The THUMS model, developed by TCRDL has a knee model which has been validated against various experimental tests Nagasaka et al (2003).

Apart from these computational models, physical models such as POLAR II pedestrian dummy from Honda R&D and EEVC legform from TRL are also in use. Results from these models have been compared against bending and shearing tests reported by kajzer et al (1990, 1993, 1997, 1999). , Even though the current knee tolerance level has been prescribed on the basis of these tests, the validity / bio-fidelity of the experimental procedure is still a matter of debate (Bhalla et al 2003).

The Total Human Model for Safety (THUMS) developed by TCRDL group validates against test conditions reported by Kajzer et al (1999) [Nagasaka et al (2003)]. No validation has been reported for THUMS under test conditions used by Kerrigan et al (2003). The earlier validations have been reported by the Toyota group who developed the THUMS model. The work reported in this paper is an independent validation of the THUMS knee model against Kajzer's tests and Kerrigan's tests.

The mesh of THUMS model, which is available in the posture of the driver in a car, has been modified to straighten the knee. Dynamic test conditions reported in Kajzer et al (1997,1999), Kerrigan et al (2003) were recreated in PAMCRASH^{TM} and simulations conducted. The FE simulation results were compared against experimental results. The model was also used to study the injury processes in the knee region.

The THUMS model corresponds to a 50 percentile American male. Cadavers on which the tests were conducted were not all 50 percentile. As needed, the FE model was scaled to match specific cadavers while conducting simulations. The response of THUMS knee model to the Kerrigan et al (2003) test conditions has been analyzed.

FE MODEL

The base finite element model (AM50, height 175 Cms, weight 77 Kg) used for this research work has been developed by Toyota Central Research and Development Labs, Japan in collaboration with Wayne state university (Wattanabe etal, 2001). The lower limb model includes cortical and spongy parts of femur, tibia and their condyles, patella (cortical and cartilaginous), meniscus and ligaments. Cortical part of bones has been modeled with shell elements while the spongy part has been represented by solid elements. Apart from these, crural tissues (muscles) and skin have also been modeled with solid elements and membrane elements respectively. Four ligaments in the knee, anterior cruciate ligaments (ACL), posterior cruciate ligaments (PCL), medial collateral ligament (MCL), and lateral collateral ligament (LCL) are of significance and are modeled in THUMS using membrane elements.

In the current work, the lower extremity mesh of THUMS has been repositioned according to the different test conditions. In order to simulate the Kajzer tests, a fully extended knee matching the test conditions has been obtained by straightening the leg by pushing it out through simulations. The lower extremity was then segregated from the remaining model for the simulations. The Kerrigan tests have been simulated by segregating the knee model from the rest of the body and applying boundary conditions. Other than repositioning, re-meshing of some zones near the knee and additional contact interfaces were needed for consistent simulations. The default material properties in THUMS have, not been modified in this study.

To access failure of tissues in the simulations using THUMS, bone fracture is said to occur if the tensile strain goes beyond 0.0082 for the femur and 0.0096 for the tibia. Similarly ligament failure is assumed to take place if the ligament strain goes beyond 0.09, 0.5 and 0.5 for the MCL, ACL and PCL, even though other sources of ligament characteristics report different limits [Bermond etal, 1994, for instance]. (Please see note under the "Discussions" section)

For comparison of forces between the experimental and simulations, experimental impact force data from the load cell attached to the impactor is compared with the contact force experienced by the impactor in the simulation. As reported in the experiments, the bending moment was measured a weighted average of the loads at the 6 axis load cells at femur and tibia ends. This procedure may not estimate the bending moment accurately in dynamic situations. In simulation, the bending moment is computed across the horizontal section through the knee.

SIMULATIONS FOR THE 20KMPH KAJZER TESTS

The FE model of knee of THUMS was simulated under the test conditions used by Kajzer et al (1997, 1999) (reproduced below in Figure 1 to Figure 4).

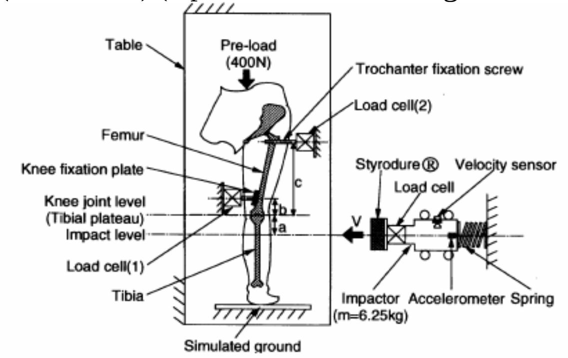


Figure 1: Shearing test set up (Kajzer et al 1999)

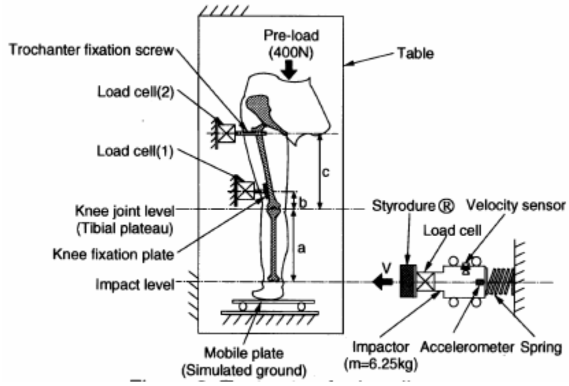


Figure 2 : Bending test set up (Kajzer et al 1999)

These experiments, conducted for knee impacts in shear and bending, were intended to recreate the impact condition in case of pedestrian-vehicle front collisions. The foot of the cadaver was supported on a table, and a load of 400 N was applied to the torso to simulate the effect of the upper body load. A foam covered 6.25 Kgs mass was impacted at speeds of 20 kmph (Kajzer et al 1999) (Figure 1 and Figure 2) and 40 kmph (Kajzer et al 1997) (Figure 3 and Figure 4).

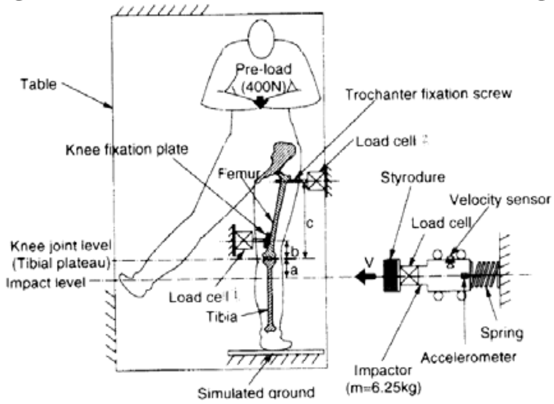


Figure 3: Test set up for shearing at 40 kmph (Kajzer et.al. 1997)

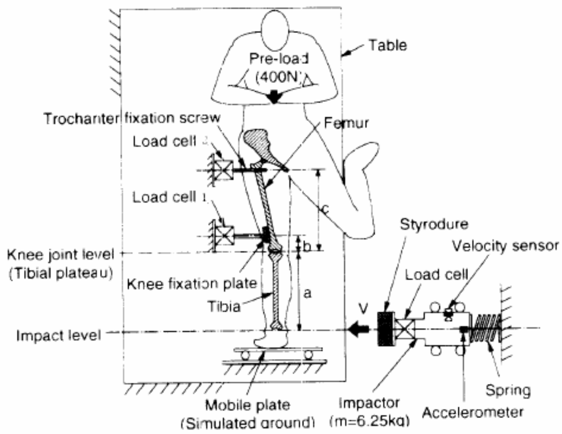


Figure 4: Bending test set up used by Kajzer et.al. (1997)

To replicate the test conditions, the nodes on the femur bone at the upper and lower locations indicated in the figure were restrained to move only in the vertical direction. A vertical concentrated load of 400N was applied at the top of the femur. A rigid support plate was modeled under the foot and a nominal value of 0.3 was used for coefficient of friction between foot and lower supports for shearing test simulations. For bending tests, a roller support was provided at the base. Coefficient of friction between foot and support plate is taken as 0.01.

The experimental setup consists of the full body. In the experiment the upper part, other than mass and inertia contributions would additionally interact through the long fibres in human body. In simulation, the nodes at the fixation screw are fixed to mimic the trochanter fixation screw and no long fibres are currently modeled in THUMS in that region. Therefore using the complete THUMS model is expected to increase computation times without changing the result. This was verified by conducting initial simulations with and without the upper body and no significant difference was observed. Hence, simulations reported here are only with the leg model.

The test results used are for TEST NO. 24S (shear) & TEST NO. 23B (bending) for Kajzer et al 1999 and TEST NO. 16S (shear) for Kajzer et al 1997. These tests were selected because the cadavers used for these were similar to the THUMS AM50 (Height = 175 cm, Weight 77 Kg). This is in line with simulations developed by Maeno etal, 2001. Figure 2 compares the upper and lower tibial displacements and impact forces for TEST NO. 24S, simulation results reported by Maeno et al 2001 and those obtained in our simulations. While the peak forces are predicted well by the simulations, the displacements diverge somewhat after the initial 20 ms. The first peak observed in the contact force after the initial peak is due to pressing of impactor against the fibula and the second peak is due to the bounce off as the impactor comes to rest.

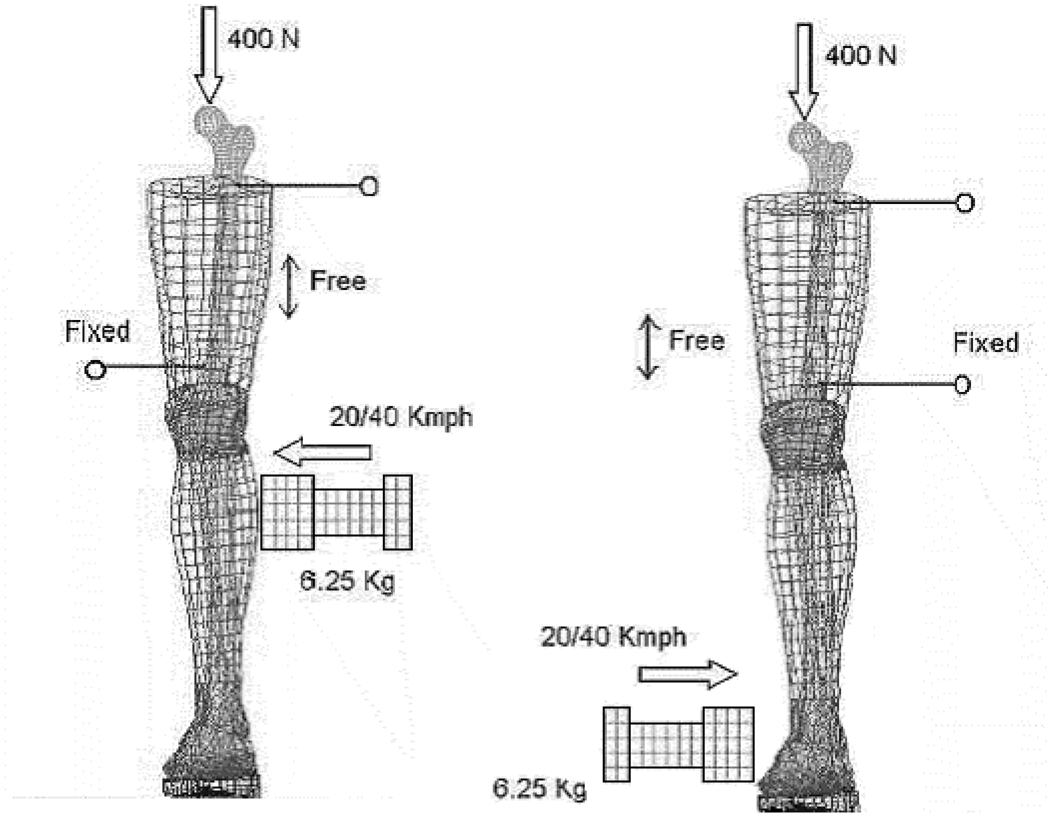
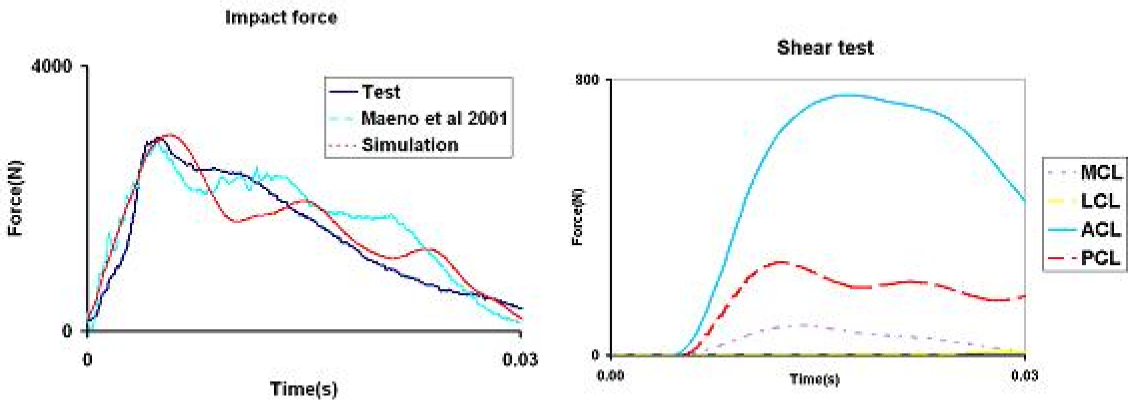


Figure 5: Knee shear and bending test set up (Kajzer et al 1997, 1999) modeled for simulations.



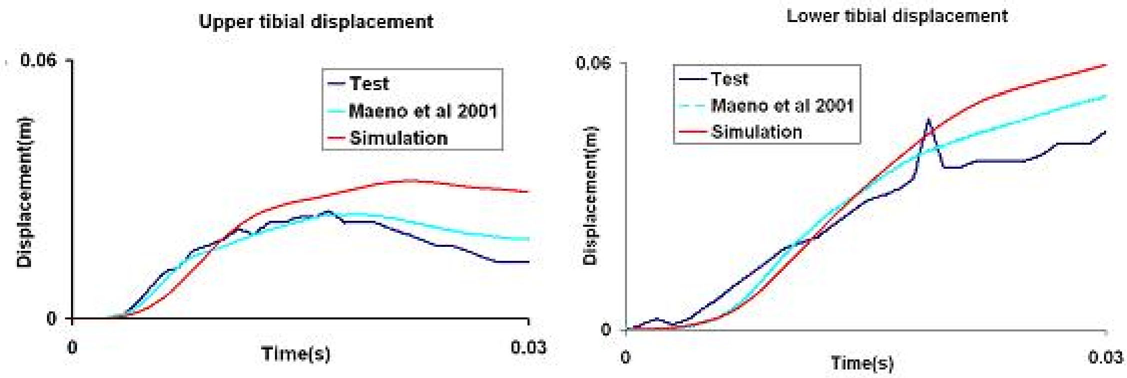


Figure 6: Ligament forces and comparison of upper, lower tibial displacement and impact force for 20 kmph tests in shear.

The injury patterns observed in these simulations have been compared with the ones actually observed in autopsies after the experiment. ACL damage was the most common injury reported in all shearing tests while no damage has been reported to the MCL in these tests. Fracture of tibia or femur epiphysis / diaphysis was reported for only one of the five tests conducted at low speeds (Kajzer et al 1999). In the simulations, the strain in the condyle regions goes upto 0.026 and in the cortical region upto 0.01. Both these are because of large localization of stresses at the support and in the contact region.

In the experiments, the ACL is one of the first ligaments to be damaged. ACL avulses and subsequently MCL damage occurs if shearing continues. In the tests, ACL damage was seen in 3 of the 5 cases, but MCL damage was not seen in any. ACL strain of 0.2 was observed in the simulations, which is below the yield limit of 0.5 being used in THUMS. The MCL strain was seen to be 0.1, which is higher than the yielding limit being used. Thus one would predict larger likelihood of MCL failure as compared to ACL failure. Hence, it is felt that even though the FE model represented the knee dynamics in shear at 20 kmph reasonably well, but it does not correctly predict the ligament injuries under these low velocity impact conditions. This could be for a variety of reasons, including, variations in the ligament properties, material properties of ligaments and their constitutive laws not being well established, ligament rupture not being included in the models. These factors are detailed in the “discussions” section of the paper.

The upper and lower tibial displacements and the impact force in bending have been compared for the 20kmph bending tests. Figure 7 shows the comparison of the results for TEST NO. 23B, simulation results of Maeno et al 2001 and our simulation results. For the first 10 ms, impact force curve for bending simulation conforms to the measured data, but subsequently drops to near zero values. In simulation there is a break in contact between impactor and foot due to bounceoff. The contact force rises again when the contact is regained.

MCL injury (2 out of 5 cases) is the primary mode of injury observed in low velocity bending tests (Kajzer et al 1999, 1993). Bone injuries are scarce. MCL is the first ligament to take up the load in bending followed by ACL and PCL. The damage pattern in simulation shows MCL stretch, with a strain of 0.24, which is well above the yielding limit for MCL. It is also observed that after the initial impact, the leg moves away from the impactor, due to which the contact force reduces to almost zero and then rises. It is observed that primary resistance to bending is due to the MCL, which once avulsed, would lead to failure of the ACL and the PCL. Figure 7 shows the forces in the ligaments in these simulations. An examination of the material models used in THUMS indicates that the stiffness of the ACL/PCL is about eight times that of MCL/LCL. So the ACL/PCL takes up most of the load and the MCL forces are consistently low in the simulations. This would suggest that the material law used for the ligament in THUMS needs to be reexamined.

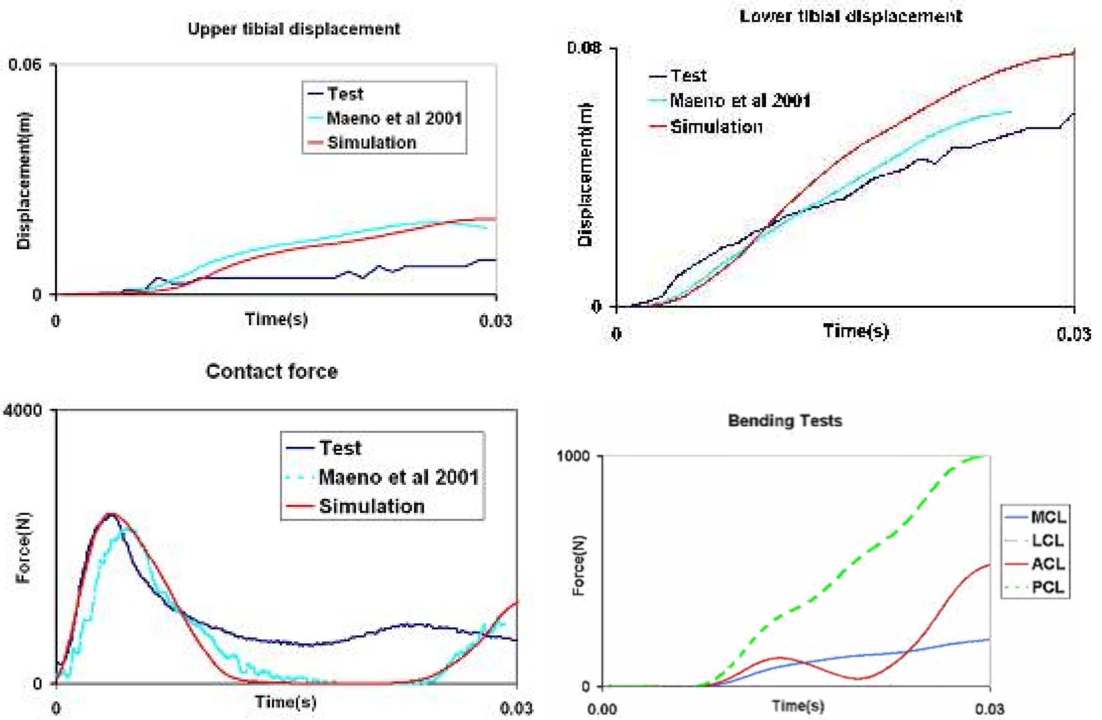


Figure 7: Ligament force and comparison of impact force, upper tibial displacement and lower tibial displacement with test results reported by Kajzer et al (1999) and validation results by Maeno et al (2001)

Real world accidents do not lead to pure shearing or bending loading. Teresinski and Madro (2001) reported from vehicle collision data that 8.3% of the impacts were from the front, 30.3% from the rear and 57.5% from the side of pedestrians. The loading to lower extremities can differ in these impacts. Seen from the front, the lateral bending can be classified as inward or outward. The influence of inward / outward bending was examined using the FE model. The validation reported above has been reported for the outward bending of knee (for the left leg). The results of the inward bending are illustrated in Figure 8.

It is observed that the impact force time histories are similar for inward and outward bending. While in outward bending, MCL stretch occurs, it is the LCL that is stretched in inward bending. Therefore, the similarity in response for both the cases can be attributed to the similarity of the failure mechanism.

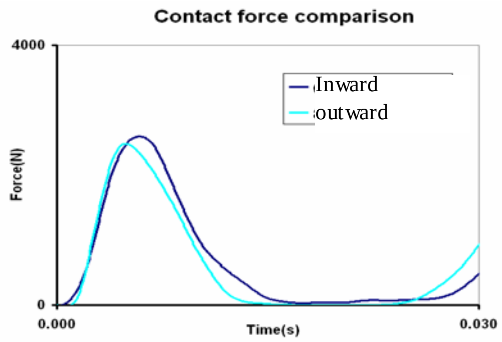


Figure 8: Comparison of impact force for inward and outward bending of knee joint.

SIMULATIONS FOR THE 40KMPH KAJZER TESTS

An arrangement similar to the one used for 20 kmph tests was used for 40 kmph tests. The difference being in the velocity of impactor. The foam-covered impactor, weighing 6.25 kg was incident on the lower extremity at 40 kmph to simulate the tests reported by Kajzer et al 1997. Even though the full cadaver was used for these tests, the other parts would not play a significant role because of the kind of constraints applied at femur and were not included in the simulation.. Other conditions were the same as for the 20 kmph tests.

The time histories of impactor force, bending moment and displacement from the simulation of TEST no. 16S are shown in Figure 9. The progression of upper tibial displacement in simulation is in good agreement with Test 16S and 17S results. The shear force and bending moment is higher than the test results after the initial 20 ms. This phenomenon could be attributed to the absence of a failure model for ligaments in the current model [Please see discussion section]. The ligaments hence continue to stretch and provide resistance, even after the point of avulsion.

Bone injuries including femur diaphysis fracture, tibial fracture, osteochondral fractures were the most frequent injuries reported by Kajzer et al 1997. Apart from bone injuries, ACL damage was also observed. In simulation, the peak strain in femur at lower support was 0.027 which is higher than the allowable ultimate strain 0.008. This would suggest femur damage. The strain values (0.09) observed in tibial condyles were also found to exceed ultimate strain (0.0097). ACL damage was also observed in the tests (6 out of 10 cases). In simulation, peak ACL strain is observed to be 0.25 as against the yielding limit of 0.5. Figure 9 shows the ligament forces observed in these simulations.

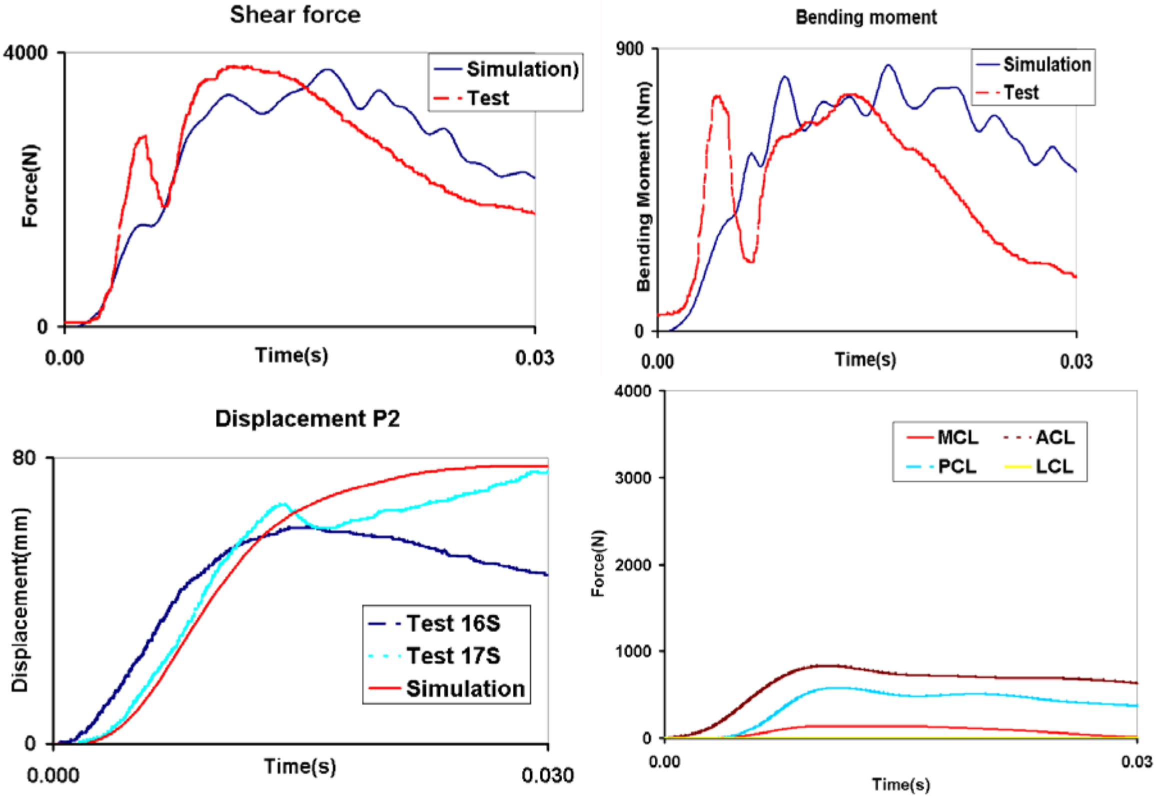


Figure 9: Ligament force and comparison of shear force, bending moment and upper tibial displacement. Test data is from Kajzer et al (1997).

The THUMS lower extremity was also analyzed under test conditions used by Kajzer et al 1997 for conducting bending tests on knee at 40 kmph. Figure 10 compares the results obtained in the simulations with those obtained in the experiments. The time and force histories predicted by the simulation differs from the test results and damage patterns are also different. Though the shear force history is somewhat similar to the test results, the predicted bending moment history is at variance. Autopsy indicated that apart from MCL damage, femur injuries are expected in such loading condition. But the simulation predicts tibial condyle injuries, with the peak strain (even though localized) in tibial condyle being as high as 0.12 against an ultimate limit of 0.0096. In addition, MCL strain of 0.4 in simulation suggests likelihood of MCL failure, even though MCL failure was observed only in 3 out of 10 cases in the experiment. Therefore from the above criterion, it seems that THUMS lower extremity does not respond very well under bending test conditions used by Kajzer et al 1997.

The response of the THUMS knee model validates well in shear and bending against 20 kmph tests and to a certain extent in the shear test at 40 kmph. It does not, however, seem to validate in bending at 40 kmph. However it was documented that the response of a cadaver is unpredictable at high velocity impact conditions such as the one used by Kajzer et al (1997). The results obtained in bending have differed significantly from test to test (Nagasaka et al 2003). The repeatability of these tests therefore needs to be examined and a response corridor established in order to use these tests for FE model validation. In the tests reported in Kajzer et al (1997) none of the cadavers used for bending tests had configuration similar to that of AM 50. The simulated ground support could be a source of uncertainty, as the exact behavior of roller supported plate under dynamic conditions might deviate from the ideal system modelled.

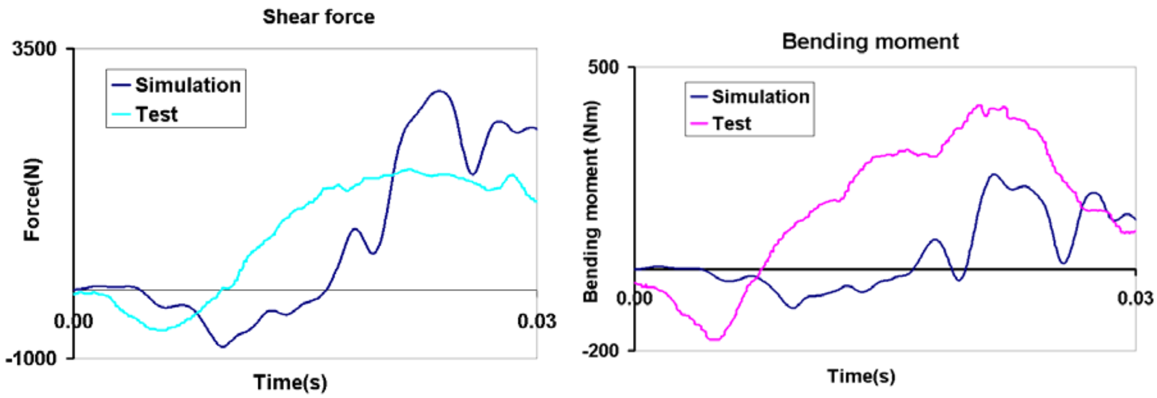


Figure 10: Shearing and bending moment comparison for bending tests at 40 kmph.

SIMULATION OF THE KERRIGAN'S TESTS

Kerrigan et al (2003) reported four point bending and shear tests for human knee joint. They hypothesized that pure shear conditions never occur in practice and such damage patterns have not been observed in real life crashes. Hence, they recommended pure bending tests or combination of shear and bending tests of the type reported by Bose et al (2004). The four point bending test reported by Kerrigan et al (2003) have been modeled and compared with the experiments in Figure 11.

The bending moment and the shear stresses limits reported by Kerrigan were at variance with the tests against which earlier standards were being set [Kerrigan et al, 2003]. To establish the suitability of the THUMS model under different configurations, modelling for the test conditions used by Kerrigan assumes importance. Our simulations indicate that the FE model of knee joint which validated against the Kajzer test results, reproduce the kinematics reported by Kerrigan et al 2003 as shown in Figure 12. This indicates that the disagreement between Kerrigan's and earlier results could be due to the different boundary conditions in the test. *A single model can predict the kinematic response for both tests.*

Though the bending moment plot obtained from simulation of four point bending test is qualitatively similar to that of test result, the magnitudes do not match. (Test (1st peak 100Nm), simulation (65 Nm)). One source of error could be the simple massless model of the links and measuring instruments that were attached to femur and tibia used in the simulation, mostly ignoring their inertial property. The THUMS knee model may also need to be enhanced to reflect the geometry and properties of the soft tissues better.

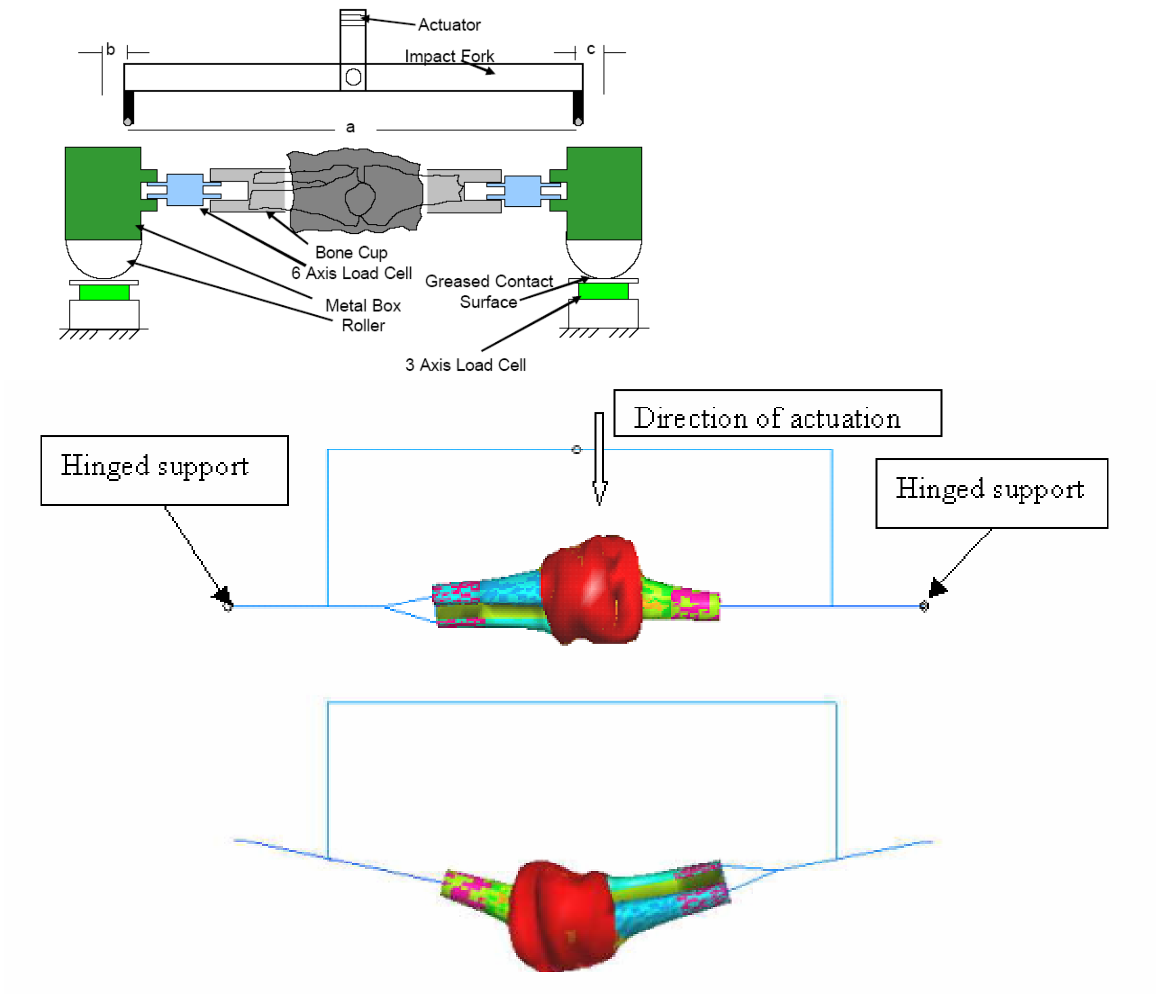


Figure 11: Experimental set up (Kerrigan et al 2003) and the set up used for simulation.

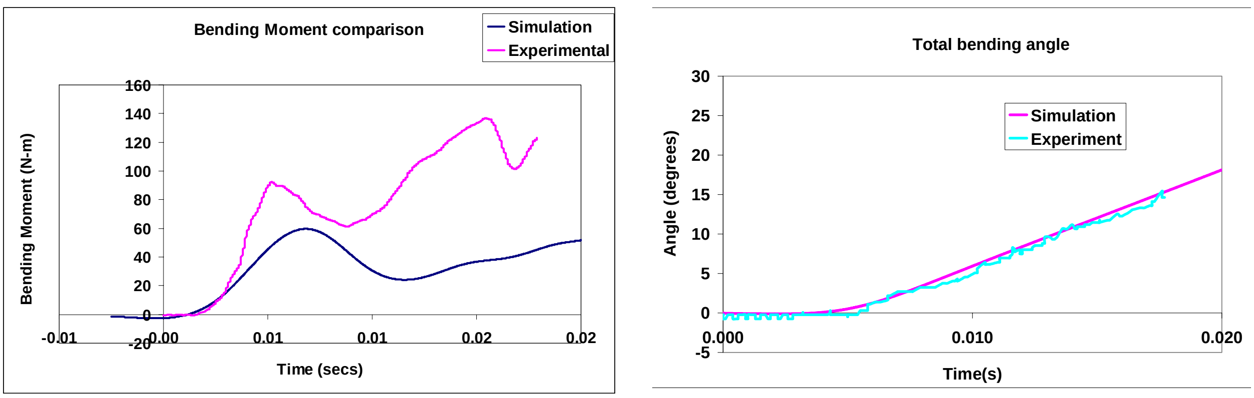


Figure 12: Plots for bending moment and total bending angle (Test results are from Kerrigan et al 2003)

DISCUSSIONS AND CONCLUSIONS

SUMMARY OF SIMULATION RESULTS: The following conclusions can be drawn from the work discussed above:

1) The predicted response from simulation of tests reported by Kajzer et al (1999) compare satisfactorily with the test results. The injury pattern is also similar to the pattern observed in shear and

bending tests. Hence it can be concluded that the FE model of human knee validates against the said results and can be used to predict injury pattern of human knee under low speed impact.

2) The time histories predicted by simulation of shear tests at 40 kmph conform to the test results for the first 20 msecs. The injury pattern is also similar. Subsequently, there is a variation which can be attributed to the absence of failure mechanisms in the Thums model.

3) The injury pattern of FE model of human knee does not match that of tests in bending at 40 kmph (Kajzer et al 1997). Hence more work needs to be done to be able to use the model to predict the impact response in bending at these speeds.

4) The simulation predicts the kinematics of Kerrigan's tests.

ISOLATED KNEE vs FULL CADAVER TESTS: The experiments reported by Kajzer et al (1990, 1993, 1997, 1999) were the only one being used for all validation work for physical models and computational models. The tolerance levels for knee reported in these tests were close to that of femur and tibia. Kerrigan et al 2003 reported tests on knee region in more controlled test conditions, where the knee region was specifically isolated and tested in shear and bending with different set of boundary conditions. It was reported that the knee tolerance values were much lower than the ones reported previously in literature. The variation is predicted from the same FE model.

The results reported by Kerrigan et al (2003) were found to disagree with previous results available in literature. Therefore, the simulation for the test conditions used by Kerrigan was significant. Four point bending test reported by Kerrigan et al (2003) was simulated, where the links were modeled using bar elements and were assumed to be rigid. It was found that the FE model of knee joint, which validated against the Kajzers test results, resulted in bending moment qualitatively close to that reported by Kerrigan et al 2003. The plots for bending moment and total angular displacement are given in Figure 11. This suggests that the disagreement of Kerrigan's results with previous results could be due to the different boundary conditions involved.

As mentioned in Kerrigan et al (2003), axial loading might have its distinctive effect on knee tolerance under impact. Simulation with the FE model which was validated for Kajzer's experiments returns lower bending moment when simulated for Kerrigan's test without axial loading. The effect of axial loading could be investigated by varying it in simulations using validated models. Segregation of knee joint from the rest of the leg also might have its effect on knee tolerance values observed in simulation.

INJURY PATTERNS: Response in shear and bending tests at 40 kmph (Kajzer et al 1997) has been found to vary from test to test. The shear force and bending moment were matched against Test 16S. It can be seen that the bending moment and shear force values are higher after the peak in simulation than the test results. This could be attributed to the fact that the material model defined for knee ligaments does not allow elimination of elements to obtain the effect of avulsion of ligaments. While the validation was reasonable for the first 20msecs of the shearing test, the same model did not respond satisfactorily subsequently and also in bending at 40 kmph. The maximum shearing force in simulation was found to be higher than the test result, while the maximum bending moment was found to be lower. The local strains in femur region were not high enough to predict a fracture in that region, which has been consistently observed in bending tests. However, MCL, PCL and ACL injuries were predicted as reported in autopsy results.

LIMITATIONS OF THE STUDY: This study has been limited somewhat by the non-availability of complete data of the experiments and deviations in the geometry of the cadavers from the AM 50. Variations in geometry as well as properties from cadaver to cadaver, repeatability of the experiments and establishing appropriate corridors in these experiments are other issues, which need to be addressed. The simulations use idealized boundary conditions which may not have been reproduced in the experiments as the motivation of the tests was to establish injury corridors and not FE validation.

Another point of note is that in simulations boundary conditions at the support often give high localized stresses / strains which have to be ignored during interpretation.

These simulations have given us a good insight into the THUMS knee model and also requirements needed from human body FE model. Other than comparing with the experiments, the numerous simulation runs give us a basis to highlight aspects of these models that need further attention for fidelity under diverse loading conditions.

DESIRABLE IMPROVEMENTS IN MODEL: The THUMS knee needs to be modeled in greater detail. There are parts of knee joint such as the joint capsule that have not been included in the current model. These parts hold the joint together. In addition the ligaments are not modeled in detail. The current model does not allow the ligaments and tendons to be pre-tensioned. This leads to the tendons slacking on knee flexion, affecting the mechanics.

Failure models are not defined knee components in the model of THUMS we worked with, though inclusion of failure model for ligaments has been reported in later versions of the THUMS model [Nagasaka et al (2003)]. In the model we used, elements continue to stretch indefinitely under load, without failure / rupture. So the associated force does not drop to zero. Incorporating an appropriate rupture model for the bones as well as for the ligaments is suggested for improving the impact response of the knee.

The simulation results suggest that the properties of ligaments in knee joint need to be verified. This is particularly important as ligament injuries are of considerable interest in most situations. The peak force in MCL is observed to be much lower than the force in ACL, even though MCL is being loaded more severely. The contact interfaces have been defined for limited regions in knee in THUMS. This results in incorrect model performance under impact conditions, as the regions of contact change in dynamic conditions. The contact response in FE simulations is also affected by the resolution of the local mesh. Contact stresses between bones coming in contact are of particular concern as they cause condyle fractures in tests, but are not being predicted reliably in the current model. We feel that a better contact algorithm will help human body simulations considerably, and are working towards one.

To summarize, in this paper we have compared the response of the THUMS knee model against 20kmph shear and bending tests, 40kmph shear and bending tests and 4 point bending tests. The 4 point bending tests by Kerrigan predict different loading limits from the earlier standards. Simulations suggest that the difference is because of the varying test conditions and may not represent different limits physically. The current model validates reasonably well in most cases except the 40kmph bending tests. The possible reasons for the deviation in 40 kmph bending tests have been discussed and possible directions for improvement have been suggested. There is scope to update material models for soft tissues, include failure / rupture models, improve contact interfaces and inclusion of structures currently omitted from the knee. We are currently investigating most of these issues and would have more suggestions in these areas in the months to come.

References

1) Atkinson T., Atkinson P, *'Knee injuries in motor vehicle collisions: a study of the National Accident Sampling System database for the years 1979–1995'* , Accident analysis and prevention, 32 (2000) pp 779–786
2) Bermond, F., Ramet, M., Bouquet, R., Cesari, D., 1994, "*A finite element model of the pedestrian leg in lateral impact",14th ESV Conference*, pp 199-209.
3) Bhalla K., Bose D., Madeley N. J., Kerrigan J., Crandall J., Longhitano D., Takahashi Y., '*Evaluation of the response of mechanical pedestrian knee joint impactors in bending and shear loading'*,19th ESV conference, Japan, 2003.
4) Bose D., Bhalla K., Rooij L., Millington S., Studley A., Crandall J., *Response of the knee joint to the pedestrian impact loading environment*, SAE paper no. 2004-01-1608.

5) Kajzer, J., Cavallero, S., Ghanouchi, S., Bonnoit, J.,Aghorbel, *Response of the knee joint in lateral impact: Effect of shearing loads*, IRCOBI 1990, pp. 293-304.
6) Kajzer, J., Cavallero, S., Bonnoit, J., Morjane, A., Ghanouchi, S., *Response of the knee joint in lateral impact: Effect of bending moment*, IRCOBI 1993.
7) Kajzer, J., Schroeder, G., Ishikawa, H., Matsui, Y., Bosch, U., "*Shearing and Bending Effects at the Knee Joint at High Speed Lateral Loading,*" Society of Automotive Engineers, SAE Paper 973326, 1997.
8) Kajzer, J., Ishikawa H., Matsui Y., Schroeder G., Bosch U., "*Shearing and Bending Effects at the Knee Joint at Low Speed Lateral Loading*," Society of Automotive Engineers, SAE Paper 1999-01-0712, 1999.
9) Kerrigan J., Bhalla K., Madeley N., Funk J., Bose D., Crandall J., '*Experiments for establishing pedestrian impact lower injury criteria'*, SAE Paper 2003-01-0895, 2003.
10) Konosu A, Tanahashi M, '*Development of a biofidelic pedestrian legform impactor'*, 18h international ESV conference, Paper No. 378, 2003.
11) Maeno Tomoyuki, Hasegawa Junji, '*Development of a finite element model of the total human model for safety (*THUMS*) and application to car-pedestrian impacts'*, 17th international ESV conference, Paper No. 494, 2001.
12) Matsui Y., *'Biofidelity of TRl legform impactor and injury tolerance of human leg in lateral impact'*, Stapp Car Crash Journal, Vol 45 (2001)
13) Nagasaka K., Mizuno k., Tanaka E., Yamamoto S., Iwamoto M., Miki Kazuo., Kajzer J., *'Finite element analysis of knee injury in car-to-pedestrian impacts'*, Traffic injury prevention, 4:345-354,2003.
14) Takahashi Y., Kikuchi Y., Mori F., Konosu A., *'Advanced FE lower limb model for pedestrians'*, 18th International ESV conference, Paper no. 218, 2003
15) Tamura A., Furusu K., Iwamoto M., Kato C., Miki K., Hasegawa J., *' Development of a finite element model of the human lower extremity for assessing automotive crash injury potential'*, Springer convention, 2000.
16) Teresinski, G., Madro, R., 2001, "*Knee joint injuries as a reconstructive factor in car-to-pedestrian accidents*", Forensic Science International, 124 (2001), pp. 74-82
17) Watanabe I., et al.; '*Development of practical and simplified human whole body FEM model'*, JSAE spring convention, 2000 (In Japanese)

Back