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Using Human Body Models to Assess Knee Ligament Injury in Knee Hyperextension

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ABSTRACT – Shared autonomous vehicles open possibilities for novel seating configurations, enabling greater interior spaciousness by making the front row seats rear-facing or removing one row of seats altogether. Frontal crash simulations with a forward-facing Hybrid III mid-size male FEM demonstrated that the unrestrained legs can swing up freely until they stop at the end of the range of knee extension. High tibia moments and tibia indices result. Similar crash simulations with the GHBMC M50-O demonstrated knee ligament separation, while those with the more advanced GHBMC F05-O did not. To better understand the knee responses, the mass, C.G. and moments of inertia of the GHBMC M50 legs were applied to the GHBMC F05 with its more detailed representation of the knee. The peak knee ligament loads are compared to published failure load data.

INTRODUCTION

Initial FEM simulations were conducted with a Hybrid III mid-size male on an automotive bucket seat and restrained by a 3-point belt. The pulse represented a 35-mph rigid barrier impact. Both legs swung up to their stops resulting in high upper tibia bending moments and tibia indices above the injury assessment reference value. The knee rotation stop arm and the shoulder bolt that prevent further rotation are both rigid. Metal-to-metal impact is prevented by a thin sleeve of natural rubber surrounding the shoulder bolt. The upper tibia moments spiked upon reaching the rotation stop, which is different than the injury mechanism of combined compression and bending upon which the tibia index was based.

NASS-CDS data from calendar years 2006 through 2015 were interrogated for frontal crashes involving belt-restrained right front and rear passengers with any knee injury. All 27 occupants meeting the search criteria were viewed. As expected, none of the knee injuries were attributed to a non-contact mechanism involving the leg swinging up and hyperextending the knee. A follow-up search for knee injuries in rear occupants of limousines was also conducted, but no cases were found. The conclusion from viewing the individual NASS-CDS cases is that occupant compartment intrusion or leg contact to the glove compartment door, knee bolster or knee airbag prevented the legs of front passengers from swinging freely to the point of knee hyperextension. For rear seat occupants, it is the back of the front seat that restricts leg swinging. Existing field data are insufficient to assess the relevance of existing injury assessment reference values in hyperextension.

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Crowell et al. (2017) reported two military cadets who injured their PCL while removing their combat boots during military survival swim training. Both are assumed to be non-contact injuries. In these medical case studies, neither the kinematics nor the failure loads are known.

Kerrigan et al. (2003) conducted preliminary ligament tension tests on bone-ligament-bone specimen of the medial collateral ligament (MCL) and lateral collateral ligament (LCL), oriented to represent 0° knee flexion. After extensive preconditioning and four sub-injurious quasi-static step loads, the specimens were ramped to failure. Van Dommelen et al. (2005) continued the tests with MCL and LCL, as well as separated fiber bundles of the anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL). Pre-conditioning, sub-failure ramp-and-hold tests, and sub-failure distractions at three different distraction rates preceded the final load to failure test of each specimen.

This paper describes simulations with the Global Human Body Model Consortium's 50th percentile Male and 5th percentile Female Occupant models (GHBMC M50-O and GHBMC F05-O) to investigate the possibility of knee injury in non-contact hyperextension. Ligament loads from the simulations are compared to published failure loads.

METHODS

The simulation represents a belt-restrained occupant in a bucket seat on a sled. The 3-point belt had a buckle pre-tensioner and a 4-kN load limiter. The forward-facing 2nd row seat has the seat cushion angle of 7° and seat back angle of 23°. The acceleration pulse represented a 35-mph rigid barrier impact.

The detailed GHBMC models M50-O_v4-5 and F05-O_v3-1 were used, initially with failure criteria for ligaments and cortical bone turned on and later with the failure criteria turned off. The more refined knee of the newer F05 yielded results that were contradictory to the less detailed M50. To investigate whether the differing results could be attributed to inertial differences of the F05 and M50 legs, the leg mass, moments of inertia and C.G. of the M50 were applied to the F05 legs. The dimensions of the lower extremities and the failure criteria were not changed from the F05. This iteration will be identified as “F05 legs up-weighted.”

RESULTS

Table 1 gives the peak values from the lap belt, femurs and tibias from the Hybrid III and GHBMC M50 simulations. The kinematics were similar. The Hybrid III knee reached the rotation stop at 77 ms. GHBMC M50 reached peak extension at 84 ms and 98 ms with the ligament and cortical failure criteria turned on and off, respectively. With ligament and cortical bone failure criteria turned on, the GHBMC M50 exhibited distraction of both knees with failures of the MCL, ACL, and PCL. There was no failure of the LCL. See Figure 1.

Table 1. Peak measurements from Hybrid III and GHBMC M50 with failure criteria turned off.

Measurement	Hybrid III	GHBMC M50
Lap belt tension (kN)	8.1	7.4
Femur tension, left/right (kN)	6.0, 6.0	3.7, 4.1
Tibia tension, left/right (kN)	3.9, 3.9	1.5, 1.8
Upper tibia moment, left/right (Nm)	1170, 834	85, 87

The simulation was repeated with the GHBMC F05 with failure criteria turned off. The lighter 5th percentile female had peak lap belt load of 4.9 kN and peak femur and tibia loads of 2.88 kN and 1.05 kN, respectively. The knees did not reach the same peak extension as GHBMC M50.

Figure 1. GHBMC M50-O knee distraction and ligament failures at 84 ms.

The final simulations were conducted with the GHBMC M50 and the F05 with legs up-weighted to those of M50. The peak femur loads of the two models were nearly identical. In both models, ligament and cortical bone failure criteria were turned off to obtain the peak tension of the knee ligaments which are given in Table 2. Within the first 100 ms of the simulation, peak elongation rates of the knee ligaments were between 200 and 400 mm/s.

Table 2. Peak ligament loads from GHBMC M50 and F05 up-weighted with failure criteria turned off.

Measurement	GHBMC M50	GHBMC F05, legs up-weighted
MCL tension, left/right (N)	117, 99	43, 29
LCL tension, left/right (N)	378, 408	216, 287
ACL tension, left/right (N)	609, 630	299, 274
PCL tension, left/right (N)	616, 777	686, 708

DISCUSSION

The peak ligament loads given in Table 2 are compared to the mean failure loads from the combined studies of Kerrigan et al. (2003) and van Dommelen et al. (2005), summarized in Table 3. Note that van Dommelen et al. did not test intact cruciate ligaments. The ACL was separated into the antero-medial (aACL) and postero-lateral (pACL) fiber bundles. Similarly, the PCL was separated into the antero-lateral (aPCL) and postero-medial (pPCL) fiber bundles. Arnoux et al. (2005) observed that the fiber bundles of intact cruciate ligaments fail at different times due to their different lengths and orientations. Therefore, the failure load of the intact PCL cannot be approximated by the sum of the aPCL and pPCL. The peak PCL tensions of the GHBMC F05 with up-weighted legs are higher than the mean failure load of the pPCL.

It is also important to note that van Dommelen et al. (2005) tested the MCL and LCL at 0.016, 1.6 and 1600 mm/s, while the separated fiber bundles of the ACL and PCL were tested only at 1600 mm/s. At the lower elongation rates, the ligaments were loaded from an initial pre-tensed state. For tests at 1600 mm/s, the ligament was initially slack in order to get the actuator up to speed before loading the specimen.

Table 3. Elongation rate, sample size and mean and standard deviation of failure loads from knee ligaments (van Dommelen et al., 2005).

Ligament	Elongation rate (mm/s)	Sample Size	Failure load (N)	
			μ	σ
MCL	1.6	5	1400	220
LCL	1.6	6	440	78
aACL	1600	4	990	560
pACL	1600	3	1000	190
aPCL	1600	2	650	-
pPCL	1600	3	290	94

All knee ligaments in the GHBMC M50-O and F05-O are modelled with the same constitutive material model. Failure criteria are defined for the M50, but not for the F05. Both models have a knee capsule. However, the knee capsule of M50 is defined as null shell elements for contact purposes only, while the F05 is an elastic material. Muscle insertions on the femur and tibia are different between the two models. The F05-O lower extremity model has the anterior thigh muscles connected to the distal femur and the posterior muscles connected to the proximal tibia and fibula. For the M50-O, the muscle group and the flesh are combined into one material that is not tied to either the distal femur or proximal tibia and fibula.

This study has two significant limitations. First, the GHBMC M50-O and GHBMC F05-O have not been validated for inertially-induced leg rotation from flexion to extension or hyperextension. Second, the knee ligament failure data from Kerrigan et al. (2003) and van Dommelen et al. (2005) are based on extremely small sample sizes of excised bone-ligament-bone specimens that were subjected to extensive pre-conditioning and numerous sub-failure tests prior to the final test to failure. Separation of the fiber bundles of the cruciate ligaments and replication of the boundary conditions of an intact knee are additional concerns. The distraction rates of the test to failure were different than the extension rates in the simulations.

CONCLUSION

Initial simulations with Hybrid III mid-size male FEM demonstrated high tibia indices resulting from the leg swinging up to the knee rotation stop. Simulations with GHBMC M50-O demonstrated ruptures of the MCL, ACL and PCL, while those with the more advanced knee of the GHBMC F05-O did not. To investigate how the advanced knee model responds to the heavier legs of a 50th percentile male, an additional simulation was conducted with the leg mass, CG and moments of inertia of GHBMC F05-O replaced by those of the GHBMC M50-O. Peak PCL tension exceeded the mean failure loads of the individual fiber bundles of PCL from a very limited study.

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