EVALUATION OF THE RESPONSE OF MECHANICAL PEDESTRIAN KNEE JOINT IMPACTORS IN BENDING AND SHEAR LOADING

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ABSTRACT

The knee joint is especially susceptible to injury in the pedestrian impact loading environment. However, the mechanical response, injury mechanisms and injury thresholds for lateral impact loading of the knee joint remain poorly understood. This paper reviews real world crash data and PMHS tests and identifies knee joint injuries commonly seen in pedestrian crashes. This is compared with results from knee joint shearing and bending tests reported in the biomechanics literature. It is shown that lateral knee joint shearing is unlikely to occur in real world pedestrian crashes.

Next, the ability of mechanical knee joint impactors, commonly used in vehicle countermeasure development, to replicate PMHS tests is studied by performing quasi-static and dynamic lateral bending and shearing tests on the TRL legform and the POLAR II knee joint. The test boundary conditions chosen are similar to those used in PMHS knee joint tests performed at the University of Virginia. Results show that both mechanical impactors are stiffer than PMHS knees in bending, although the relative difference is smaller with the POLAR II knee. In shear loading, the PMHS knee is capable of much higher shear displacement than that permitted by the TRL impactor. While the POLAR II knee permits larger shear displacement, the loads required to produce these displacements are much higher.

INTRODUCTION

Despite the prevalence of knee joint related injuries in pedestrian impacts, only a small number of experimental investigations have been performed to quantify the response of the knee joint to high rate lateral loading. An understanding of the level of forces that lead to injury is important for the development of impactors for use in vehicle countermeasure development. In addition, recent advances in computational techniques have made it possible to develop structurally detailed models of the knee joint (Takahashi et al. 2000, Schuster et al., 2000) using sophisticated material representations. Unfortunately, in the absence of experimental studies, validating these models has been difficult and finite element models of the knee joint have so far found limited applications in the design and development of vehicles for pedestrian safety.

Most current computational models and knee joint test devices have been validated by comparing with knee shear and bending tests performed by Kajzer et al. (1990,1993,1997,1999). The knee joint tolerance levels reported have been used to propose acceptance levels for sub-system impactor tests (EEVC, 1998) for vehicles sold in Europe. Similar testing is also being considered in other parts of the world, such as in Japan and Australia, even though the validity of these test methods remains controversial. For instance, authors have argued that legform impactors should not ignore the effect of upper body mass on knee bending (Takahashi et al., 2001) and that rigid segments in impactors do not account for bone bending effects (Konosu et al., 2001), which can have a significant effect on measured knee bending and shear. In addition, University of Virginia (UVA) recently reported data from a pilot study (Kerrigan et al., 2003) of knee lateral loading that questioned the biomechanical validity of the results found in the literature, which form the basis for these test methods and the design of most physical knee joint models.

In this paper, a more thorough analysis of the Post Mortem Human Subject (PMHS) data reported by Kerrigan et al. (2003) is presented and comparisons are made with literature. New data is presented for the response of two of the most commonly used mechanical legforms, the EEVC legform from TRL and the knee joint of the POLAR II pedestrian dummy from Honda R&D. These impactors are subject to dynamic and quasi-static, lateral, 4-point-bending and shearing loads using a test set up similar to that used in the PMHS knee experiments by Kerrigan et al. (2003). Stiffness of the test devices and the proposed injury thresholds are compared with the knee joint response observed in the PMHS knee experiments.

REVIEW OF REAL WORLD PEDESTRIAN KNEE INJURIES

The knee joint is a complex mechanical structure and its response is sensitive to the loading environment applied. Thus, it is important that in experimental studies of knee injuries, the joint is subject to the same boundary conditions that exist in a pedestrian-vehicle crash. Therefore, the ability to replicate real world pedestrian injuries is the most important measure of the validity of experimental testing.

Ashton et al. (1981) performed detailed accident reconstruction of 44 pedestrian crashes with knee injuries. They found tibial plateau and femoral condyle fractures (25% cases) and ligamentous injuries (43% cases). In 39% of the cases the knee on the non-struck side (side of pedestrian not struck by vehicle) sustained injuries. Bunketorp et al. (1981) performed a retrospective study of 34 injured pedestrians and found knee injuries in 13 cases (7 intra-articular fractures, 6 isolated ligament injuries.

Analysis of AIS 3+ lower limb injuries (PCDS, see Takahashi et al., 2000) from data collected from pedestrian impacts with late model vehicles showed that leg injuries occurred in 29.8%, femur injuries in 12.4%, and knee joint injuries in 18.67% (8.4% tibial plateau, 4.4% dislocation, 4.0% ligament 0.9% patella and 0.9% femoral condyles) of the cases. Teresinski et al. (2001) performed a detailed analysis of 357 pedestrian fatalities, 214 (60%) of which had knee injuries. Among pedestrians struck from the side (165 cases), knee injuries were found in 94% of the cases. Bone bruises on the tibial plateau due to compression were found in a large proportion of the cases, although, bruises of the femur were rare. Isolated injuries to the ACL were also rare (in 2 cases out of 165) in lateral impacts. Furthermore MCL injuries were found in 35% of the cases and damage to bone near the insertion site in more than 13% of the cases (Teresinski, 2003). From an analysis of injury mechanisms, the authors hypothesized that in lateral impacts 73% of all injuries resulted from valgus loading.

While an analysis of real world crash data is useful for identifying the most common injury mechanisms, the large amount of uncertainty associated with pedestrian impacts makes it difficult to analyze the loading environment which leads to these injuries. Further insight in this regard can be gained from detailed reconstruction of pedestrian crashes using simulations. For instance, Rooij et al. (2003) have reconstructed a case from the PCDS database that involves a 35 year old subject of close to 50^{th} % male anthropometry. Reconstruction shows that a vehicle moving at 65 km/h struck the pedestrian, who was in a walking stance. Bumper contact is below the knee. Injuries reported included leg fracture and ACL avulsion on the struck side knee; ACL + PCL rupture and fracture of the medial tibia plateau on the non-struck side knee.

Full-body PMHS tests are another important source for understanding the response of pedestrian impact loading. While such tests permit careful measurement of forces and motion, the role of active muscle response in pedestrian impacts is not well understood. Brun-Cassan et al. (1984) performed four PMHS (aged 70-84 years) tests using a small production car moving at 40km/h. They reported comminuted fractures of the tibia-fibula complex in all tests. In one case, this was accompanied by partial rupture of the cruciate ligaments. Kallieris et al. (1988) reported results from 11 PMHS tests at speeds ranging from 23 to 41 km/h. Leg fractures were found in almost every case but no ligamentous injuries were produced. Cesari et al. (1989) reported knee joint injuries occurring in 12 staged cadaver collisions at speeds ranging from 20 km/h to 39 km/h. The main injuries were either bone fracture, which occurred more commonly at higher speeds, or ligament rupture, but both never occurred simultaneously. Bunketorp et al. (1981) tested 19 lower limb specimens by impacting with simulated vehicle fronts that included an adjustable bumper and hood leading edge. Tests were performed at speeds of 20-28km/h using varying car front geometries. Leg fractures outside the joint were reported in only 4 of 19 tests. Ligamentous damage included 10 cases of MCL injury, 10 cases of PCL injury, 3 cases of ACL injury, and 3 cases LCL damage. A high incidence of PCL injuries have not been reported in other studies.

It is important to understand the role of leg fractures in relation to knee joint related injuries. Contact failure of the long bones in pedestrian impacts results in subsequent lowered forces in the knee joint and thus fewer knee injuries. There are two factors that can contribute to a loading environment which lead to leg fractures. First, at higher loading rates knee joint ligaments are stiffer and stronger (see, for instance, Lee et al., 2002,). Thus, as confirmed from the PMHS tests discussed above, higher impact speeds increase the likelihood of leg fractures. Second, a stiff bumper can lead to high impact forces resulting in stress concentrations at the impact site and thus failure at the impact point. Most PMHS tests performed in the 1980s used older vehicles. In the last few decades, however, bumper designs have evolved from the narrow and rigid designs in the 1970s to present day bumpers that are wider (i.e. distributed leg loading) and bumpers with an outer polymer shell (i.e. more compliant). Furthermore, although bumper standards, such as

FMVSS 581 and FMVSS 215, require bumpers to be strengthened to protect damage to the vehicle exterior in low speed crashes, increasingly vehicles are being designed for pedestrian safety, which require more compliant bumpers. Thus, it is hypothesized that as bumpers continue to become softer and wider, leg fractures will be less common and the focus of lower limb safety research will shift to preventing ligamentous damage to the knee joint.

In the studies reviewed here, the knee structures most commonly injured are

- MCL (valgus loading),
- ACL (medial shear) in the struck side knee, although these are rarely isolated injuries.
- PCL (lateral shear) in the non-struck side knee
- Tibial plateau fractures (varus or valgus loading leading to compression of tibial plateau by the femoral condyles)

Thus, experimental tests designed to study knee joint injuries should focus on designing boundary conditions that reproduce these failure mechanisms.

REVIEW OF KNEE JOINT TESTING

A vast number of experimental studies of knee joint loading have been reported in the orthopedic biomechanics literature. However, these studies involve loading the knee joint at low rates to sub-failure levels. Since several of the load bearing structures of the knee joint, such as the ligaments and articular cartilage have a non-linear rate dependent response, extrapolation of results from these studies to higher strain and higher strain rates is not appropriate. Thus, experiments that involve loading rates equivalent to those produced in pedestrian crashes need to be performed.

Viano et al. (1978) were among the first to study the impact response of the human knee joint. Since their experiments were designed to study the effect of knee bolsters on vehicle occupants, the tests included anterior-posterior loading (56 kg impactor at 6 m/s) of the tibia of a flexed knee joint. The injuries induced were either multiple fractures of the impacted bone or ligament failure (PCL and LCL avulsion). Since the knee was not loaded laterally, these tests only provide limited insight for the pedestrian loading environment.

Ramet et al. (1995) reported results from 20 quasi-static loading of PMHS knee joints in shear and bending. In the bending tests that produced injuries, MCL damage was observed in every case. Interestingly, MCL injuries dominated the shearing tests as well, with ACL injuries occurring in only 2 cases.

Kajzer et al. (1990, 1993) reported results from low speed impact loading of the knee joint in shear (9 tests at 16 km/h and 10 tests at 20 km/h) and bending (7 tests at 16 km/h and 10 tests at 20 km/h). Full lower limbs were tested by rigidly mounting the femur end and impacting the leg at the ankle level in the bending tests and with a twin pronged impactor that loaded the knee at the ankle and just below the knee joint for the shear tests. An axial pre-load of 400 N was applied to the specimen to simulate half body weight. The tests were performed on an older populations (mean age 77 years).

In the shear tests, injuries obtained were tibial spine fracture (10/19 cases), which have not been reported in real world crashes, and ACL failure (14/19 cases). Injury to the fibula head on impact and damage to the LCL, which is attached to the fibula head, was reported in 16/19 cases.

Kajzer et al. (1997) reported results from 10 tests performed in shear and 10 tests performed in bend loading by impacting PMHS (age 51 (SD 15) years) limbs near the knee joint (shear) and at the level of the ankle (bend tests) at 40km/h. The test set up was similar to that shown in figure 1. As before, an axial compressive load of 400N was applied on the specimens. In 6/10 tests, comminuted fractures outside the knee joint were observed. This was accompanied by ACL damage in 5 cases and MCL damage in 2 cases. Only in one case was the ACL injury the only injury induced. As has been discussed earlier, it is questionable to use damage outside the knee joint to characterize the response of the joint, especially since the bone fractures occur early in the test. Similarly, in 7/10 bending tests the only damage was outside the knee joint. All of the remaining three cases had MCL damage accompanied by ACL damage (1 case), PCL damage (1 case), or both (1 case).

In the bending tests (Kajzer et al. 1993) MCL rupture or avulsion in 12/17 cases includes fracture of the medial condyle. In 3 cases, the MCL damage was accompanied by ACL damage. A tibial condyle fracture was obtained in one case. These injuries have been reported in real world crashes and result from valgus knee joint loading.

Kajzer et al. (1999) reported results from 5 tests performed in shear and 5 bending using PMHS (age 63 (SD 16) years) limbs loaded at 20 km/h with a test apparatus similar to that used by Kajzer et al. (1997, see figure 1). In shear loading, injuries were induced in only 3/5 cases, all of which had ACL damage. One of these cases included a fracture outside the knee joint. In bending tests, 2 tests produced only MCL injuries, 1 only an extra-articular bone injury and no injuries were produced in 2 tests. Thus, at slower impact rates a higher incidence of ligamentous damage within the knee joint was produced. This is in line with observations made earlier based on cadaver tests, where at higher speeds, lower limb injuries were dominated by fractures. However, in order to study knee joint injuries tests need to be developed that are capable of producing damage within the joint.

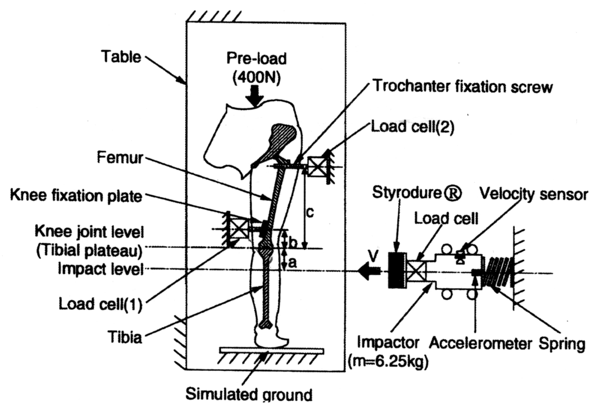


Figure 1. Schematic of the test set up used in the shear tests performed by Kajzer et al. (1999). In the bending tests, the impactor was lowered to load the limb at the ankle.

Kerrigan et al. (2003) reported results from a pilot study of knee joints performed at the University of Virginia (UVA) that suggested significant differences from past studies. The ability of these bending and shear tests to replicate the pedestrian knee joint loading environment is evaluated. The tests data is appropriately scaled in order to compare with the response of a 50th adult male. The scaled stiffness curves are analyzed and the injury threshold and injury mechanism are compared with earlier tests.

UVA KNEE JOINT TESTS

Figures 2 and 3 shows a schematic of the bend and shearing test set-up used. Details of the testing, such as specimen information, specimen preparation techniques, test fixtures and methodologies are available in the earlier publication (Kerrigan et al. 2003).

Rather than test whole limbs, as has been done in earlier studies, knee joints were isolated by sectioning the femur and the tibia-fibula complex. This permitted isolating the knee joint by two six-axis load cells and directly recording the load environment experienced by the knee joint. Since multiple knee structures are often damaged in knee joint studies, it is important to identify the timing of failure events in the tests. Analysis of high speed video (used in earlier studies) is not reliable because the load bearing structures cannot be seen in the intact joint. Thus, Kerrigan et al. (2003) used acoustic sensors mounted on the femur and tibia. These sensors record acoustic emission that accompanies failure events.

In the knee joint bending tests, the rotation of the supports was directly measured using angular velocity sensors, which provide an accurate measurement of knee bending angle. In the shear tests, applied shear displacement was measured using displacement transducers. However the results reported for actuator displacement are expected to be higher than the applied knee shear displacement because there was bending in the fixtures of the test set up (Kerrigan et al., 2003). Since the tests were imaged using high-speed digital cameras, the motion of the cups was obtained from video analysis. Thus, shear displacement results after compensating for bending are shown in this paper.

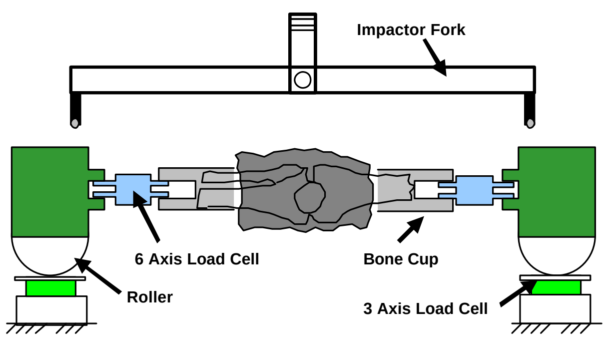


Figure 2. Schematic of the four-point knee bend test set up used in the UVA knee tests (Kerrigan et al. 2003).

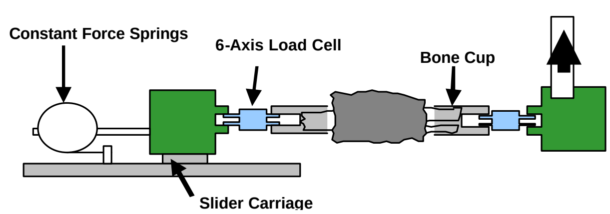


Figure 3. Schematic of the shear test set up used in the UVA knee tests (Kerrigan et al. 2003). Constant force springs are used to apply a compressive axial force on the knee joint.

As suggested by Irwin et al. (2002), results should be appropriately scaled in order to account for the varying anthropometry of the subjects tested. Using a methodology of dimensional analysis similar to the one proposed by Irwin et al. (2002), scaling factors for force, λ_{force}, momentum, λ_{moment}, and displacement λ_{disp} are easily related to an equivalent length scaling factor λ_{Lequiv}:

- $\lambda_{force} = (\lambda_{Lequiv})^2$
- $\lambda_{moment} = (\lambda_{Lequiv})^3$
- $\lambda_{disp} = \lambda_{Lequiv}$

An equivalent length scaling factor, λ_{Lequiv}, can be derived by accounting for both mass and height of the subject as $\lambda_{Lequiv} = (\lambda_{mass}.\lambda_L)^{1/4}$ by recognizing that $\lambda_{mass} \sim \lambda_L^3$. Scaling factors were derived by using the weight, 164.1lb, and height, 69.29", of the H-model (Finite Element Human Model, Takahashi et al., 2000) as a reference. It should be noted that these numbers are close to that for a 50th % Male (169.8lb, 69.8", based on Cheng et al., 1994). Table 1 shows the derived scaling factors and the scaled results are shown in Figures 4 and 5.

Table 1: Factors derived for scaling test results

Spec	Test	Age (yrs)	Sex (M/F)	Height (in)	l_L	l_{mass}	l_{Lequiv}
124R	Bend	58	F	5'9"	1.00	1.31	1.07
167L	Bend	66	M	5'10"	1.01	0.81	0.95
135L	Bend	63	M	5'8"	0.98	0.93	0.98
121L	Shear	40	M	5'	0.87	0.82	0.92
167R	Shear	66	M	5'10"	0.87	0.82	0.92
169R	Shear	62	M	5'7"	1.01	0.81	0.95

Notes: λ_x = (x in test subject) / (x in target anthropometry)

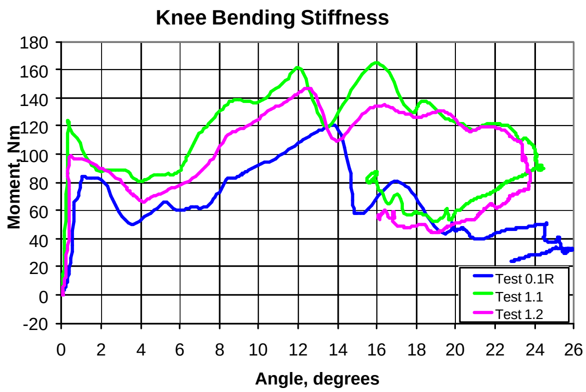


Figure 4. Bending stiffness of the knee joint. Results have been scaled to match an adult male of 164.1lb, mass, and 69.29", height.

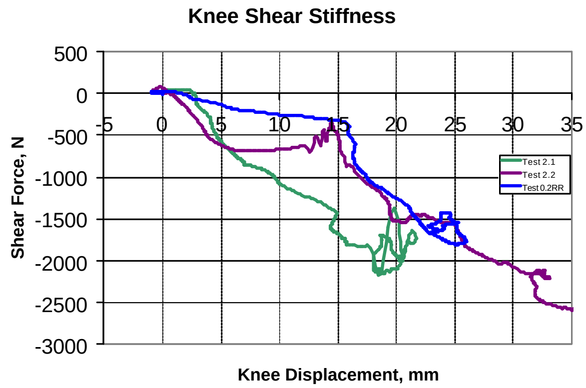
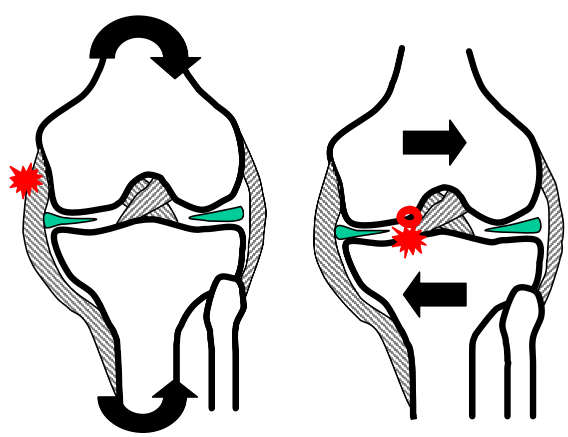


Figure 5. Shear stiffness of the knee joint. Results have been scaled to match an adult male of 164.1lb, mass, and 69.29", height. Results from Test 0.2RR should not be used for quantitative analysis, since the specimen used had been subject to prior testing.

In each of the bending tests (figure 4), an initial peak is observed which lasts for the first few milliseconds. Since the load cells that isolate the knee joint rotate with the specimen, this initial spike is a result of inertial loading of the specimen. However, these accelerations occur only at the beginning of the test (less than 4^o bending, figure 4) and do not affect the failure data recorded. This was confirmed by measurements made with an actuator mounted accelerometer. After the initial peak, the moment continues to rise steadily until a rapid fall in load representing failure, which was accompanied by acoustic emission in each case recorded by acoustic sensors placed on the femur and tibia. Figure 6(a) shows the injury mechanism (MCL tear) common to all three knee bending tests performed. In one case (Test 0.1), this was accompanied by fracture of the medial femoral condyle and in another case, Test 1.1, by a partial tear of the ACL and damage to the lateral meniscus. Thus, the peak in bending moment between 12^o and 14^o, which is accompanied by acoustic signal suggesting a failure event, is hypothesized to be due to failure of the MCL.

Three shear tests were performed, one (Test 0.2RR) of which was subject to repeated testing and thus the force-displacement data should not be used for quantitative analysis (figure 5). An initial inertial spike is not seen in the shear tests data shown in figure 6 because the loads reported are from the femoral side knee load cell which moves little during the test. Figure 6(b) shows the most common injury mechanism (partial ACL tear and osteo-chondral fracture). In one of the two tests, Test 2.2, where the applied shear displacement was larger, this injury mechanism was accompanied by damage to both menisci and the ACL tear was complete. The osteo-chondral fracture observed in all the shear tests results from the tibial spine gouging/plowing into the femoral condyles. Thus, it is clear that the applied shearing force is resisted by both, bone-bone contact within the knee joint and the ACL. The importance of bone-bone contact in resisting shear displacements in pedestrian impact is unclear since real world pedestrian crash data currently reported in the biomechanics literature does not suggest that such an injury mechanism is common. It is likely that osteo-chondral fractures were observed in these tests because the shearing displacement was applied in the presence of axial force corresponding to full body weight. Future testing should investigate the role of this compressive joint load in order to mimic real world crash injuries.

In comparison with bending tests, the relative timing of knee damage in shear tests is difficult to evaluate. The knee shear forces are seen to have a steadily increasing trend with shear displacement. Since tibial-spine gouging/plowing is likely an ongoing process, a drop in forces is likely due to ACL damage. Thus, it is hypothesized that the early peak in shear forces (at 12.7 mm of shear displacement, 693N shear force) in Test 2.2 is due to ACL failure. Similarly, ACL failure in Test 2.1 occurs at a shear force of 1839N and a shear displacement of 17.8 mm.



(a) (b)

Figure 6. Damage mechanism in knee bending, (a), and knee shear, (b), loading.

Table 2 compares the results of these knee tests with the test results from Kajzer et al. As pointed out earlier, most of the tests performed by Kajzer et al. caused extra-articular damage rather than knee joint damage. Thus, in Table 2 only load levels from tests that caused ligamentous damage are included. While the results from Kajzer et al. (1993) are comparable with those reported in the UVA tests, results from Kajzer et al. (1997,1999) are much higher. Similarly, for the shear tests (Table 3), the tests performed by Kajzer et al. which led to knee joint damage are compared. The mean shear forces ranging from 2400N to 3200N reported are much higher than those reported in the UVA tests, 1839N and 693N in the two shear tests performed.

Table 2: Reported bending tolerance of the knee

	Speed	**Structure damaged**	**Moment Nm(±SD)**	**Angle °(±SD)**
Kajzer et al. 1993	7 @ 16km/h	MCL in 5 tests	101 (±21)	9 (±2)
	10 @ 20km/h	MCL in 7 tests	123 (±35)	11 (±3)
Kajzer et al. 1997	10 @ 40km/h	MCL in 3 tests	284 (±18)	14.6 (±0.2)
Kajzer et al. 1999	5 @ 20km/h	MCL in 2 tests	358 (±167)	12 (±3)
Kerrigan et al. 2003**	3 dynamic, nonimpact	MCL in all tests	**143 (±20)**	**12.7 (±0.9)**

* Results from Kajzer et al. (1993, 1997, 1999) shown are only for the tests in which knee ligament damage was induced
** Peak bending moments at time of MCL failure shown.

These differences in results can be largely attributed to differences in test methodologies. As pointed out earlier, if the loading environments are different, the injuries produced are different and thus the reported tolerances will be different. In the bending tests performed at UVA, a twin pronged impactor was used. The resulting four-point bend environment ensures that the knee joint is subject to minimal shear forces. A servo-hydraulic test machine was used to apply displacement controlled (constant velocity) boundary conditions in both the bending and shear tests. Such a methodology minimizes inertial effects by ensuring that the joint is loaded at constant velocity with minimal acceleration of the impactor during most of the impact phase. It should be noted that even in such a test environment, some acceleration is unavoidable as the impactor starts from rest and accelerates to a constant velocity. However, this usually lasts only for the first few milliseconds, well before the onset of injury. Using a servo-hydraulic test machine is preferred over using a free flying impactor because controlled acceleration of the specimen makes it possible to produce knee joint injuries without failure of bone outside the knee joint.

As discussed earlier, the presence of rate-dependent load bearing structures in the knee (such as the ligaments) makes both the stiffness and failure levels of the entire knee joint sensitive to the applied loading rate. Thus, in order to reproduce the loading environment existing in a 40 km/h (11.11 m/s) vehicle-pedestrian impact loading environment, appropriate boundary conditions should be chosen in order to match the knee joint loading rate (ideally, by matching the knee joint bending rate and knee shear displacement rate). Since in a vehicle-pedestrian impact, the lower limb is free to move at both the hip joint and at the ankle joint, these motions are restricted in knee joint tests, lower impact velocities need to be chosen. Kajzer et al. (1997) mention an alternative approach where they state that the effective mass of the free flying impactor used was calculated from the bumper force and leg accelerations developed in former full-scale car pedestrian impactors. In the UVA knee bending tests, the mean knee bending rate is 0.952°/ms, which is similar to those (approximately 0.941°/ms over first 17 ms) produced by Kajzer et al. (1997) in their tests at 40 km/h. In the UVA knee shear tests, the mean knee shearing rate is 0.898 mm/ms, which is much lower than those (approximately 4.3 mm/ms over the first 6 ms) produced by Kajzer et al (1997).

Table 3: Reported shearing tolerance of the knee.

	Speed	Structure damaged	Force, N (±SD)
Kajzer et al. 1990*	9 @ 15km/h	ACL in 7 tests	2570 (±370)
	10 @ 20km/h	ACL in 7 tests	3220 (±460)
Kajzer et al. 1997**	10 @ 40km/h	Epiphysis, ACL	3200 (±1000)
Kajzer et al. 1999***	5 @ 20km/h	ACL in 2 tests	2400 (±200)
Kerrigan et al. 2003****	3 dynamic non-impact	ACL in all tests	**1266 (±810)**

* In all but 1 of these cases, the ACL injury was accompanied by damage outside the knee joint (fibula head crush). Results shown are for all tests.
** Results shown are for the cases with no diaphyses or metaphysic fracture (ie damage outside the joint).
*** Results are for the two cases with only ACL damage.
**** Test 0.2RR was repeatedly tested and is not analyzed.

KNEE JOINT MECHANICAL IMPACTORS

There are several instrumented mechanical legs that have been developed for use in vehicle design and pedestrian safety countermeasure development. Most of these model the knee joint by a deformable element. Examples of these include the Rotationally Symmetric Pedestrian Dummy (Cesari et al., 1991); the JARI-I impactor (Matsui et al. 1999) and the TRL legform impactor. The TRL legform is the device recommended for EEVC WG 17 sub-system impactor tests and is, thus, the most widely used device. However, the biofidelity of the device has been questioned by several researchers (Takahashi et al., 2001, Konosu et al., 2001). In addition, Matsui et al. (1999) performed tests on the JARI-1 and TRL legforms to replicate the bending and shear tests performed by Kajzer et al. (1990, 1993, 1997 and 1999) and found that these legforms did not fall in the proposed biofidelity corridors, especially for shear displacement.

Recognizing that a geometrically accurate model of the knee joint would be necessary in order to build a knee joint model that could be validated for use in both bending and shear, Artis et al. (2000) and Wittek et al. (2001) discuss the development of a new legform impactor based on the knee joint of the POLAR II dummy. The development of the original version of the POLAR pedestrian impact dummy (Honda R&D) was reported by Akiyama et al. (2001) and Huang et al. (1999). The design was based on the advanced frontal crash test dummy, THOR (Rangarajan et al., 1998). In subsequent work, POLAR II was developed by adding more geometric details especially to the knee joint. In this knee joint, reusable compressions springs are used to represent the cruciate and collateral ligaments, ellipsoids are used to represent the femoral condyles and an elastomeric pad is used to represent the tibial meniscus. The resulting force bearing structures, springs, rubber tubes, polymer menisci have a highly nonlinear response. Wittek et al. (2001) and Takahashi et al. (2001) evaluated the response of this impactor to knee joint studies performed by Kajzer et al. (1997, 1999) by subjecting it to similar impacts. They found that the response was close to those reported by Kajzer et al. for both shear and bending experiments.

It should be noted that none of the existing legform impactors incorporate the effect of active musculature. However, this is justifiable since past studies (Takahashi et al. 2000) have found that the lateral bending response of the knee joint is not significantly affected by the muscles and tendons surrounding the knee.

In the present study, the response of the POLAR II knee joint and the TRL legform were compared with those reported by Kerrigan et al. (2003) from PMHS tests. This is done by applying boundary conditions, similar to those in the PMHS tests to the impactors.

TEST METHODOLOGY

Figure 7 shows the bending test set up for the POLAR II knee. As in the PMHS tests, the knee joint was attached to rectangular cross-section tubes that were mounted on rollers. Load was applied using a hinged twin-pronged fork. The test configuration results in pure bending with minimal shear forces transmitted to the knee joint. Load was applied using a servo-hydraulic test machine in displacement control. Instrumentation to measure the history of displacement, angles, and loads was the same as that used in PMHS tests to permit direct comparison. Thus, it should be noted that the results reported here are not from the instrumentation available standard with these impactors.

The POLAR II knee shear test set up is shown in figure 8. The femur end was mounted through supports on a linear bearing and a compressive axial force of 800 N was applied using four constant force springs. The tibia end was attached directly to the actuator of the servo-hydraulic test machine.

In both test configurations, the knee loading environment (including knee lateral bending moment) was measured directly using a 6-axis load cell mounted between the impactor and the femur attachment. The bending angle of the knee joint was measured with angular rate sensors mounted directly on the femur and tibia. As discussed earlier in the context of the PMHS tests, this shear set up results in bending of the fixtures. Thus, the shear displacement was computed from analysis of the high speed video recordings of the tests.

Since the PMHS test was designed for knee joint specimens isolated by sectioning mid-femur and mid-tibia, the methodology had to be modified to account for the TRL impactor, which is a full legform and thus much larger. For the TRL bending tests (figure 9), simply supported boundary conditions were used. The twin pronged fork was used to push directly on the surface of the impactor to force it into valgus loading. Idealized four-point beam bending was assumed and the measured support loads were used to calculate the bending moments reported here.

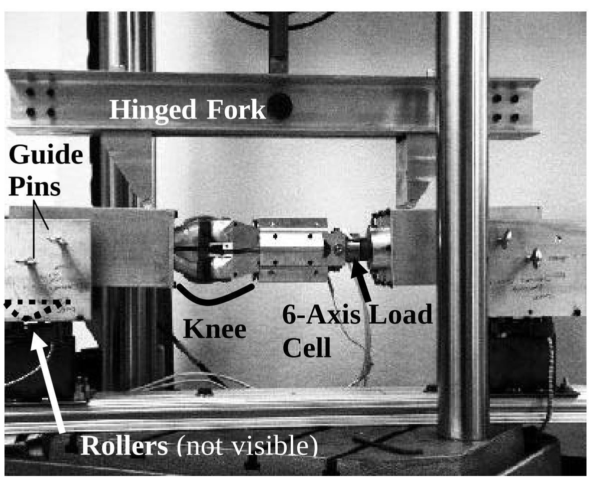


Figure 7. POLAR II knee 4-point bend test set up.

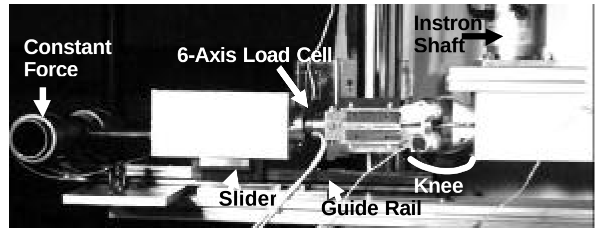


Figure 8. POLAR II knee shear test set up.

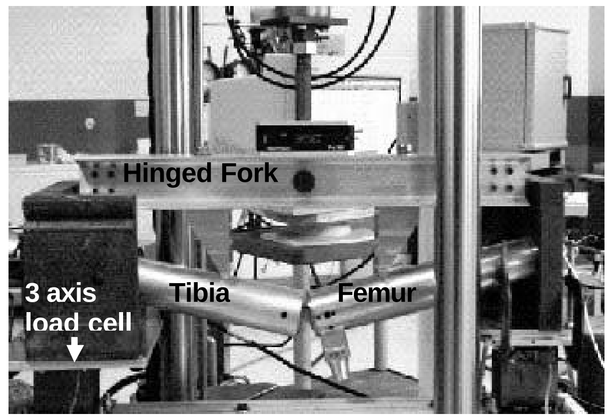


Figure 9. TRL impactor 4-point bend test set up.

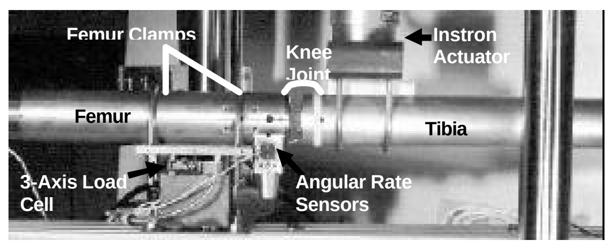


Figure 10. TRL impactor shear test set up.

For the TRL shear tests (figure 10), since the knee joint does not have an articulating surface, it was judged that the application of a compressive axial load would not affect the knee shear response. Thus these tests were performed without the axial load. As shown in figure 10, shear was produced in this impactor by holding the femur end fixed and attaching the tibia end to the actuator. Shear forces reported here were measured at the fixed support. Shearing displacement was computed from video analysis by tracking the displacement of the tibia relative to the femur.

Table 4 shows the impactor test matrix. Tests were performed in both quasi-static and dynamic loading at 1 m/s. As previously stated, for knee bending tests, this impactor velocity results in knee bending rates similar to those reported by Kajzer et al. (1997) in impacts at 40 km/h. Multiple tests were performed in each configuration in order to establish repeatability of test procedures. A fresh pair of ligaments was used in each test with the TRL impactor. The POLAR II knee consists of reusable parts and thus none of the components were replaced between tests. Post test inspection of the knee joint showed that permanent damage had not been produced in any of the components.

Table 4: Impactor bending and shear test matrix

	Quasi-static	Dynamic
TRL Legform Bend	1 tests	2 tests
TRL Legform Shear	-	1 test
POLAR II Knee Bend	2 tests	2 tests
POLAR II Knee Shear	-	1 test

RESULTS

Figure 11 shows the results of the dynamic and quasi-static bend tests performed on the TRL legform plotted along with the three PMHS tests (data shown is scaled to 50th % male) performed by UVA. The two dynamic TRL bending tests show identical results. This establishes repeatability of test procedures. Since the bending deformation is resisted by the two deformable knee elements, the trend of the bending results show an initial steep slope elastic portion followed by reduced stiffness corresponding to plastic, permanent deformation of the knee elements. The dynamic tests result in slightly higher forces, which could be due to slight rate dependence in the deformable metal elements. In addition, it is possible that during the bending tests, some shear deformation is produced which engages the shear damper, leading to a loading rate dependent response

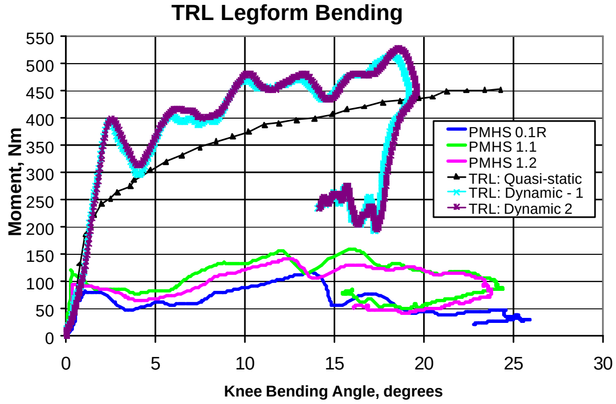


Figure 11. Comparison of the TRL legform lateral bending stiffness with that measured in the PMHS knee tests.

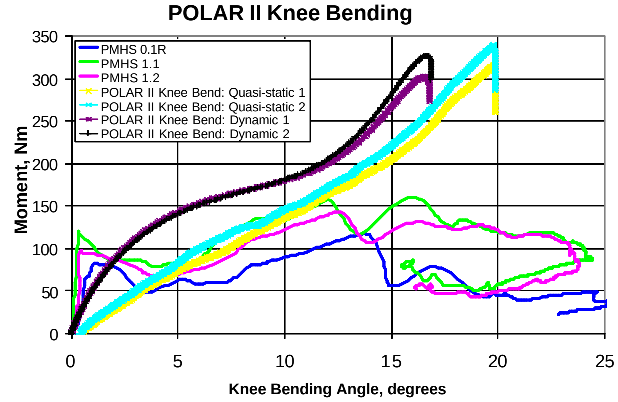


Figure 12. Comparison of the POLAR II knee lateral bending stiffness with that measured in the PMHS knee tests.

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In comparison with the UVA PMHS tests, figure 11 shows that the TRL knee joint is much stiffer. MCL failure in the three PMHS tests starts at a bending angle of 12.7^{o} (mean) , which corresponds to a mean bending moment of 143 Nm (mean at start of MCL failure). At the same bending angle, the bending moment produced in the dynamic TRL legform bend tests is 470.5 Nm, which is over 3 times larger than that in the PMHS tests.

Figure 12 shows the results of the two dynamic and two quasi-static bend tests performed on the POLAR II knee joint plotted along with the three PMHS tests (data shown is scaled to 50th % male) performed by UVA. The dynamic tests show higher forces, which can be attributed to the various rate dependent load bearing structures in the POLAR II knee. At 12.7^{o} of bending, which corresponds to MCL damage in the PMHS tests, the bending moment in the POLAR II knee had a mean value of 178.5Nm in the quasi-static tests and 218.3 Nm in dynamic loading. While these values are much higher than those reported in the PMHS tests (, the relative difference is smaller when compared with the results from using the TRL impactor.

Figure 13 shows the results of the dynamic shear tests performed on the TRL legform and POLAR II knee plotted along with the two PMHS shear tests (data shown is scaled to 50th % male) performed by UVA. The TRL impactor is designed for a maximum shear displacement of 8mm (injury is assumed to occur at 6 mm of displacement), thus the forces rapidly increase when it is loaded beyond this point. The PMHS tests performed at UVA do not clearly identify the timing of injury in the PMHS shear tests. Nevertheless, it is clear that the tolerance for shear displacement is at least 12.7mm (PMHS test 2.2) and possibly much higher, as discussed earlier. In contrast, the POLAR II knee permits higher shear displacements. However the forces generated are much higher than those in the PMHS tests.

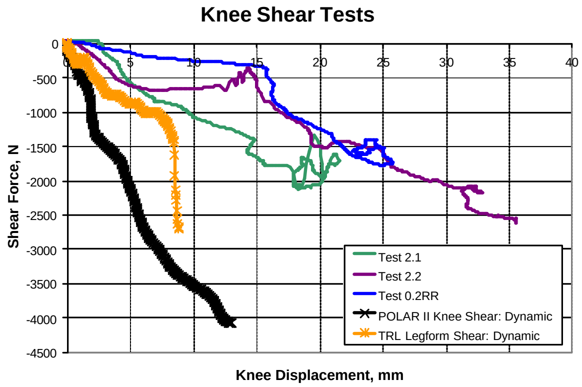

Figure 13. Comparison of the lateral shear stiffness of the POLAR II knee and the TRL legform with that measured in the PMHS knee tests.

It should be noted that there are some differences in the way shear displacements should be interpreted in the PMHS tests and the POLAR II knee joint. In the PMHS tests, the shear displacement is computed from the motion of the bone cups and any bending of the tibial and femoral end of the specimens was ignored. Such bending is not possible in the POLAR II knee joint, which has rigid ends for the femur and tibia. Bending of the long bones, which is considered important for biofidelity (Konosu et al.,2001) is incorporated by using a deformable tibia and the response of the entire legform has been validated by comparing with Kajzer et al. (Takahashi et al.,2001, Artis et al.,2001). However, in the current tests, only the knee joint was used. Thus, the deformations measured are under estimated for this loading condition.

DISCUSSION AND RECOMMENDATIONS FOR FUTURE WORK

Thus, it is seen that in comparison with the PMHS tests the TRL legform is much stiffer in lateral bending and is not capable of the large lateral shear displacements that can be induced in a human knee without injury. Similarly the POLAR II knee joint is stiffer than the PMHS knee in bending although the difference is smaller when compared with the TRL legform. In shear, the POLAR II knee permits application of larger displacements. However, the results suggest that the shear forces required may be much higher.

It is important to recognize that while these results suggest that the response of PMHS knees are not replicated by the knee joint impactors, the validity of some of the PMHS tests in reproducing pedestrian impact loading is still under question. The legitimacy of the test boundary conditions should be evaluated by comparing the resulting injuries with real world crash data. In lateral-medial impacts of the limb, current pedestrian injury data indicate the need to reproduce MCL damage, tibial plateau fractures, and ACL ruptures. The bending tests performed at UVA (Kerrigan et al., 2003) were able to repeatedly produce the first injury listed, MCL damage. Since it has been hypothesized that tibial plateau fractures occur due to valgus knee loading in the presence of compression, the effect of including an axial compressive load in knee bending tests needs to be studied in future tests. ACL ruptures were produced in the shear tests performed. However, these were always accompanied by osteo-chondral fractures from tibial spine gouging/plowing into the femoral condyles, which have never been reported as an injury mechanism in knee joint tests. While, the use of an axial compressive force (equivalent to full body weight) may be partly responsible, it is unlikely that tibial spine gouging can be completely eliminated in shear tests. Thus, pure shear loading of the knee joint may be an unrealistic scenario and it is possible that shear is always accompanied by knee joint bending in real world pedestrian impacts.

In the pedestrian injury data analyzed by Teresinski et al. (2001), isolated ACL injuries were rare in lateral impacts. Among the bending tests performed at UVA, a partial rupture of the ACL was observed in addition to MCL damage in one test (Test 1.1). These observations agree with the recommendations by Takahashi et al. (2001), that a combined knee bending and shear criteria should be used for knee injury thresholds. Thus an alternative test procedure for PMHS knees should be considered, in which the relative fraction of shear and bending load on the knee joint is systematically varied. One approach for such testing would be to perform three point bend tests on knee joints by varying the location along the limb at which the bending force is applied.

CONCLUSIONS

In summary, the following conclusions can be drawn from this study:

PMHS Knee Joint Tests:

- In most PMHS leg tests, early fracture of the impacted bone results in injuries outside the knee joint. Such data should not be used to determine the tolerance of the knee joint.

- Dynamic, non-impact (displacement controlled boundary conditions) knee tests (UVA tests, Kerrigan et al., 2003) ensure that there are no extra-articular injuries.
- Knee bending tests are capable of reproducing real world pedestrian injuries.
- Pure shear of the knee joint is an extreme case that does not occur in real world pedestrian crashes.
- Additional knee tests are urgently needed to characterize the response of the pedestrian knee. These tests should seek to replicate real-world pedestrian injuries.

Mechanical Knee Joint Tests:

- Both the TRL legform and the POLAR II knee joint are stiffer than the PMHS knee in pure bending. However, the relative differences using the POLAR II knee are much smaller.
- Both the TRL legform and the POLAR II knee are much stiffer than the PMHS knee in shear loading. However, pure shear of knee joints is an extreme loading environment that occurs rarely in real world impacts.

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