DEVELOPMENT OF A NEW BIOFIDELIC LEG FOR USE WITH A PEDESTRIAN DUMMY

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ABSTRACT

A new mechanical leg has been designed and developed for use in impact tests simulating pedestrian accidents. The leg has been developed to be part of a new, 50th percentile male pedestrian dummy called Polar. The leg incorporates a knee that has the essential anthropomorphic features of a 50th percentile male knee and a deformable tibia. The leg has been designed to be biofidelic under lateral impact. The knee possesses a human-like geometry that includes the femoral condyles and the tibial plateau. The four principal ligaments, namely, the posterior cruciate, anterior cruciate, the lateral collateral and the medial collateral are included along with a deformable meniscus. The tibia includes a deformable section that provides compliance during lateral loading. The femur and ankle were redesigned to attach to the knee and tibia respectively, and to a standard Hybrid III foot and femur ball joint. The leg is instrumented with two 5-axis load cells at the distal femur and proximal tibia and a 4-axis load cell at the base of the tibia. This paper will present the results of component impact tests performed on the knee and compare them with Post Mortem Human Subject (PMHS) data presented by Kajzer, et al [1997]. In addition, this paper will present results from static and dynamic component tests conducted on the tibia and compare them with PMHS data presented by Yamada [1970] and Nyquist [1985].

KEY WORDS

Leg, pedestrian dummy, knee, tibia, and ligaments.

PEDESTRIAN INJURIES and fatalities are a major concern in many countries. In Japan, pedestrian accidents account for almost 30% of all vehicle related roadside fatalities [Ishikawa, 1991]. In Europe, almost 18% and in the U.S. about 13% of all traffic related deaths arise from pedestrian fatalities. From accident analysis, the most frequently injured body regions are the head and lower extremities. For non-fatal accidents the lower extremities are the most commonly injured body region, accounting for 40% of the injuries, though these were typically AIS 1 and AIS 2 injuries.

In pedestrian crashes, the subject is typically impacted in the lateral direction. In a special study of pedestrian accident cases in the U.S. from 1990-1996, about 40% were impacted on the left side and around 30% on the right side [Jarret, 1998]. About 50% of the injuries resulted from impacts at below 25 km/h and over 75% at impacts below 40 km/h.

The lower extremity injuries of AIS 2 and AIS 3 which were recorded, were usually fractures of the tibia, femur, and pelvis. Direct injuries to the knee are a smaller percentage of the overall lower extremity injuries, though the actual occurrence of ligamentous injuries at the knee are not always recorded [Edwards, 1999]. Also, accident studies indicate that knowledge of the injury potential of the

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lower extremities is important in understanding multiple injuries. In particular, the presence of tibia fracture mitigates the severity of knee and ankle injuries [Edwards, 1999]. On the other hand, tibia fractures seem to increase the likelihood of femur and/or pelvis fractures. The occurrence of ligamentous injuries at the knee was also pointed out in a study by Takeuchi, et al [1998] who studied pedestrian injuries in the Gothenburg area from the period 1985 to 1995.

BIOMECHANICAL REQUIREMENTS

The mechanisms involved in producing injuries at the lower extremities during vehicle impacts have been investigated extensively using post-mortem human specimens (PMHS) [Pritz, 1978; Bunketorp, 1983; Aldman, 1985]. Recently, Kajzer, et al. [1997, 1999] have carried out a series of lateral impact tests to elucidate the response of the knee and concluded that the primary injury mechanisms are due to shearing and bending. Both fractures and ligament damage were generated from these tests. Based on the tests performed by Kajzer and his associates, a new set of biofidelity corridors have been derived by Matsui, et al [1999]. These provide the force response expected during both shear and bending impacts to the lower leg at both low (20 km/h) and high (40 km/h) speeds. They pointed out that the previous corridors which were the basis for the development of the legforms developed by TRL and JARI were based on PMHS data that were not representative of the average 50th percentile male (50AM). These subjects were from an elderly and mainly hospitalized group (average age 78 years). The new data were based on a younger, non-hospitalized group (average age 51 years) which would be more representative of the pedestrian involved in an actual accident. The new corridors developed for knee response during lateral impact are presented in figures 13, 14, 15, and 16.

BACKGROUND

The frequency of knee injuries has prompted the development of test procedures that aim to assess the aggressiveness of vehicle fronts against pedestrian lower extremities. The European Experimental Vehicles Committee (EEVC) has developed component test procedures which includes a specific subsystem test for impacting a legform against the vehicle bumper [Harris, 1989]. A number of different legforms have been developed to evaluate the injury potential during an impact with a vehicle using the EEVC test procedure. An early effort was the Rotationally Symmetric Pedestrian Dummy (RSPD) leg developed by Chalmers and INRETS [Cesari, 1989]. Currently, the TRL leg and the JARI leg have been used in the legform subsystem tests since they satisfied the original corridors developed for the impact response of the knee. Typically, these legs contain a deformable element connecting the femur and tibia, which acts as a plastic hinge when the applied moment reaches a critical value. It has been pointed out that currently none of the available legs meet the latest biofidelity requirements for knee response to lateral impact [Matsui, 1999].

Recently, GESAC in collaboration with Honda R&D and the Japan Automobile Research Institute (JARI) developed a new pedestrian dummy called Polar [Akiyama, 1999]. Full-scale impact tests have been performed with the dummy and its kinematics have shown good correspondence with PMHS motion [Huang, 1999]. The first version of this dummy, known as Polar I, had a compliant element just below the standard knee joint (which was a standard pin joint allowing only flexion and extension motion). The compliant element was designed to approximately provide the lateral compliance that would be expected during the impact with the vehicle bumper.

The compliant knee in Polar I had some significant limitations. The center of rotation in lateral bending was about 5 cm below the actual knee joint. The bending moment and shear response of the element was not quantified and there was no instrumentation to measure either the loads or the displacements that would be encountered during a lateral impact. In order to address these limitations, and to make both the lower extremity kinematics and the force response more biofidelic, a program was initiated to design and develop a new knee and lower leg component for the Polar dummy.

DESIGN OF NEW KNEE AND TIBIA

The initial goal of the design program was to improve the overall performance of the pedestrian dummy with a knee that more closely represents a human during lateral impact. Early in the design process, it became evident that it would be useful to examine if compliance could also be built into the tibia. The bending of the tibia absorbs part of the initial impact energy during lateral impact, and is thought to help in reducing the loads seen at the knee and in turn modify the overall lower leg kinematics during the impact sequence [Edwards, 1999].

The design process began with the definition of the features that would be required in the new knee. It was decided that, to properly simulate the response of the human knee, significant features of the human knee anthropometry would have to be represented. The purely geometric features included the femoral condyles, the tibial plateau, and meniscus. In order to provide the proper joint resistance, the four principal knee ligaments, namely, the anterior cruciate, the posterior cruciate, the medial collateral, and the lateral collateral ligaments would be represented. Also the meniscus would have to have appropriate force-deflection characteristics to provide proper femur - tibia contact response. The presence of the patellar tendon was kept as an optional element, since it would not be a primary factor in lateral impact.

The design was based on a 50th percentile American male human knee, and began with a polyurethane surgical knee model used by orthopedics to illustrate the shape, range of motion, and location of the ligaments of the knee. Basic dimensional measurements were made off the model and compared with the 50th percentile AATD data developed by Schneider [1983]. The overall dimensions such as the breadth and depth of the condyles and tibial plateau corresponded well with the Schneider data.

COMPUTER MODELING - Before beginning the actual design phase, a series of computer simulations were performed to verify the placement of the origin and insertion points for the four ligaments, and the general force-deflection characteristics of the ligaments and the meniscus. The lumped mass modeling was performed using the simulation program DYNAMAN [Shams, 1993]. The general model was similar to that developed by Yang, et al [1995] in simulating the response of the knee during lateral impact. The simulations modeled the lower leg impact tests performed by Kajzer [1997].

The femoral condyles were represented using ellipsoids and the tibial plateau simplified using planes. The ligaments were modeled as spring-dampers. The input geometry of the knee was based on the polyurethane surgical human knee model described earlier. The input stiffness characteristics of ligaments and the force-deflection characteristics of the meniscus were based on published values [Yang et al, 1995].

Table 1. Ligament stiffness (Yang et al. 1995)

	ACL	PCL	MCL	LCL
Stiffness (N/mm)	90	120	65	60
Damping (N-sec/mm)*	0.09	0.09	0.09	0.09

*Damping characteristics were estimated.

The condyle interface had a stiffness of 750 N/mm and a friction coefficient of .05 (from Yamada 1970, and Yang et al. 1995). Some hysteresis was also added as part of the force-deflection function. The impactor was modeled according to the specifications given by Kajzer [1997]. It had contact dimensions of 100 mm x 120 mm with a mass of 6.25 kg. Impact speeds of 20 km/h and 40 km/h were modeled. The impact point, for the shear type tests, was 84 mm below the knee joint.

The output from the simulation is compared with the output from an average high-speed shear test. It is seen that the maximum stretch is generated in the ACL and MCL, which are generally the ligaments liable to be damaged.

Table 2. Comparison of DYNAMAN knee model and Kajzer tests (1997, 40 km/h)

	Shear Force (N)	Shear Displacement (mm)	Max. Ligament Force (N) / Displ (mm)			
			ACL	PCL	MCL	LCL
DYNAMAN Knee Model	3100 (6 ms)	17 (6 ms)	2400 N 27 mm	2100 N 17.5 mm	1200 N 18.5 mm	2300 N 38 mm
Kajzer et al. (1997, 7 tests)	2500 ! 400 (5.0 ! 1.0 ms)	20 ! 1.0 (5.0 ! 1.0 ms)	-	-	-	-

TESTING OF PHYSICAL MODEL - For the initial concept testing, the polyurethane surgical human knee model was used. The model is made of a stiff polyurethane, and was determined to be strong enough to withstand normal impact loads. Synthetic ligaments were used to model the actual knee ligaments and a series of static and dynamic tests were performed to determine the response of the system.

At this time, a number of candidate materials were evaluated for use as ligaments. Materials used in actual knee prostheses were obtained and static tests were performed to compare their tensile characteristics with human data. Two materials that had potentially useful properties are shown in the table below.

Table 3. Tensile data on materials tested for use as ligament.

Material	Length	Load vs. Deflection	Ultimate Load
3/16" Double Braided Dacron Rope	95 mm (3.75 in)	154 N/mm (880 lb/in)	4316 N (970 lb)
5 x 0.1" Twisted Strands of Nylon Rope	51 mm (2.0 in)	150 N/mm (855 lb/in)	7965 N (1790 lb)

Test Fixture - For the initial tests, a test fixture was designed to hold the physical knee model. The test fixture differed somewhat in its configuration with the original setup of Kajzer [1997]. Instead of using a full dummy, the knee model used was only a section of the lower extremity from mid-femur to mid-tibia. It was felt that since the femur was fixed to a stationary system with fixation screws in Kajzer tests, a component test with the femur rigidly attached to a large plate would be adequate to test the knee model response in a manner similar to that of the PMHS tests. At the mid-tibia, a mass block was attached to provide the effective mass of 40 N to represent the portion of the leg below the knee. In between the mass block and the tibia, at the base and the sides, a rubber pad (Buna N Rubber, 40 Shore A) was used to add compliance to allow the knee to slightly bend. The test fixture allows preloading to be applied so that the force felt at the knee would correspond to that of the PMHS.

The knee was instrumented with a load cell connected to the femur to measure the shearing force. An LVDT was attached to the mass block of the unstruck side to measure shear isplacement. The impact location corresponded to the Kajzer shear test setup. It was felt that since the mass block was attached a short distance from the knee, limited knee bending will result therefore only shear displacement was measured.

The physical model was impacted at slow speed and the results were used to validate the DYNAMAN knee model described previously. The actual force-deflection characteristics of the rope used in testing were used in the simulation model. Figure 1 compares the impact force from testing with the simulation model.

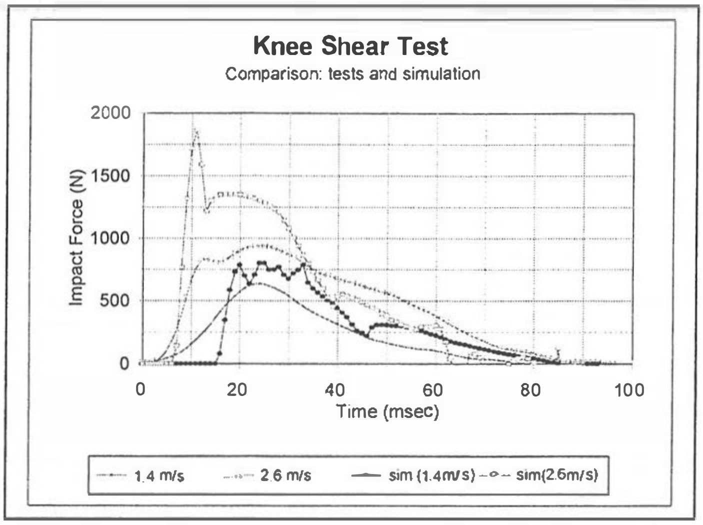
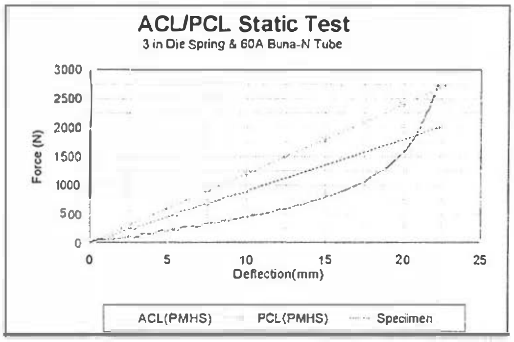
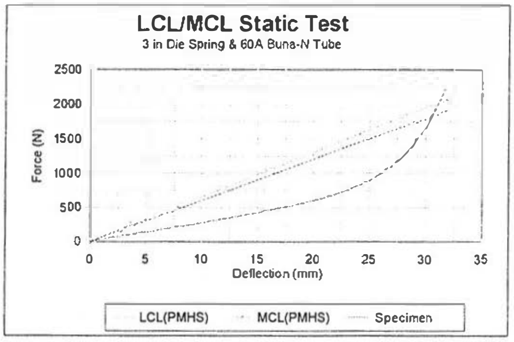


Figure 1. Comparison of impact force from testing with the simulation model.

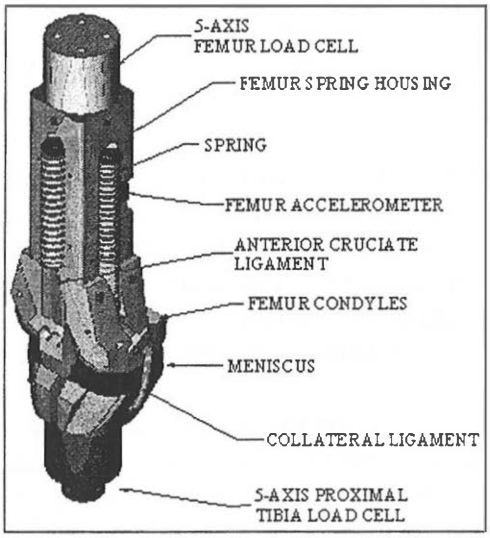
DEVELOPMENT OF KNEE DESIGN - Once it was established that the physical knee model would provide responses qualitatively similar to that seen in the actual PMHS tests, the physical knee model was digitized and the DXF file imported into AutoCAD. Within AutoCAD, the irregularities in the digitized shapes of the condyles and tibia plateau were smoothed into elliptical shapes with left/right symmetry. The meniscus was cast in Urethane and was made thicker than the actual human meniscus to provide for durability and intercondylar eminence was made much broader for the same reason. The durometer selected for the meniscus was to allow the component to have the effective stiffness of the actual human meniscus.

The ligaments were represented by non-linear spring system of steel springs and rubber/Urethane tubes. The stiffness of the spring tubes was selected to match the values provided by the Dynaman simulation model. Since the force-deflection response of the steel and rubber combination is highly non-linear, the stiffness of the rubber was selected to match the force at the point of rupture of the PMHS. The ligament path and attachment points were based on the surgical knee model. To simplify the design, the lateral and medial ligaments were made symmetric with the same stiffness. Similarly, the anterior and posterior cruciate ligaments have the same stiffness. The stiffness characteristics of the ligaments are shown in figures 2 and 3. A three-dimensional front and rear view of the new knee design is shown in figures 4 and 5 respectively.





Figures 2 & 3. Force-deflection characteristics of the artificial ACL/PCL and MCL/LCL ligaments.



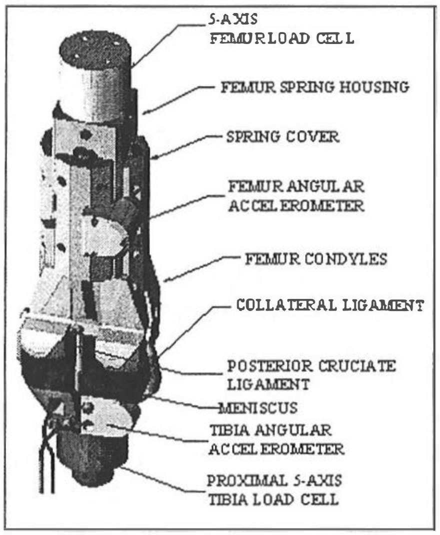


Figure 4. A front view of the new knee. **Figure 5.** A rear view of the new knee.

DEVELOPMENT OF TIBIA DESIGN - As mentioned in the beginning of the paper, a deformable tibia was thought to provide a more realistic response during the initial contact phase with the lower extremity. The design requirements for a deformable tibia were:

1. It should have human-like force-deflection characteristics up to the point of fracture in quasi-static lateromedial loading response as seen in PMHS testing performed by Yamada et al [1970].
2. It should have human-like force-deflection characteristics up to the point of fracture in dynamic lateromedial loading response as seen in PMHS testing performed by Nyquist et al [1985].
3. It should meet the above biomechanical requirements in addition to being a reusable component.

The biomechanical design requirements were placed because it was felt that the deformation of the tibia prior to fracture was the critical information that is needed to be captured. The characteristics of the deformable tibia would control the lower extremity kinematics and the load transfer to the knee. The effect of fracture would have secondary influence on the final kinematics. To validate this assumption, a number of computer simulations were performed in which a deformable element, which could sustain ultimate fracture was used in a pedestrian model. The results showed that there were no significant influence on the final body trajectory, once the energy absorbed in the deformation is accounted for.

The reusable design requirement was placed to make the dummy easier to certify and maintain. In general, the use of a frangible component would have some uncertainty in its characteristics because the actual component would become damaged if calibrated dynamically. Instead, a similar component is used in calibration testing, with the idea that the actual component will have similar performance characteristics. Moreover, a reusable element would be easier to maintain, since it would not have to be replaced after each test.

The design features of the tibia are described below and an illustration of the tibia is shown in figure 6:

- The external surface was constructed of a hard urethane (75D) hollow rod. The selection of the hard urethane was based on its deformation characteristics and impact strength. In order to maintain anthropometry, while meeting the required static and dynamic response characteristics, a stiffer material namely a nylon/kevlar rod was inserted in the middle of the urethane. The combination of a softer and stiffer material allows the tibia to bend similar to a human tibia, up to the point of fracture, without reaching its elastic limit.
- The proximal and distal ends of the tibia were tapered to resemble a human tibia and to relieve high stress concentration accumulation.

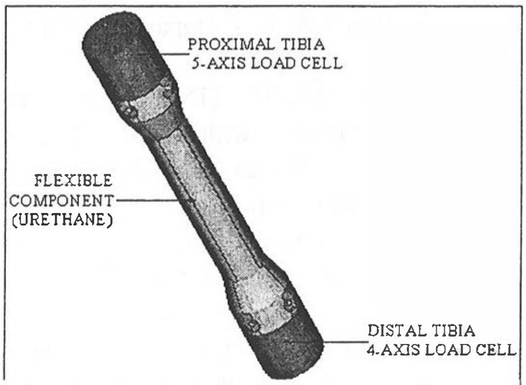


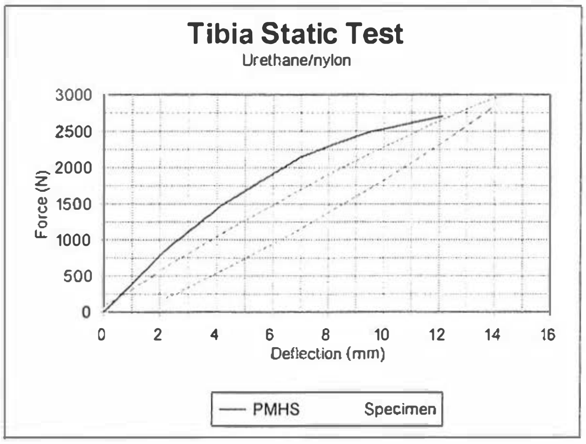
Figure 6. A 3-D view of the new tibia.

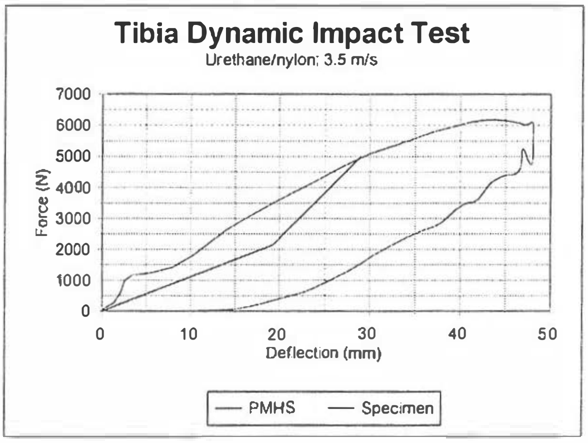
- The proximal and distal ends were reinforced with internal and external steel rings that were bonded to the tibia. The steel rings provide additional strength to the ends of the tibia for attaching to an upper and lower tibia load cell.
- The design provides interchangeability between left and right legs.

Tibia Testing and Results - Quasi-static three-point bending tests were conducted similar to the test setup described by Yamada et al [1970]. The tibia without flesh was simply supported at the proximal and distal ends of the tibia. The blocks were equally separated about the midspan of the tibia a distance of 287 mm. A constant static load was applied by a universal hydraulic static test machine equipped to measure force and relative displacement of the loading head. The loading head was a 25 mm diameter cylindrical rod; 305 mm long oriented with its longitudinal axis perpendicular to the tibia.

Dynamic three point bending tests were conducted using a 32 kg linear impactor similar to the test setup described by Nyquist et al [1985]. The impacting head was a 25 mm diameter rod, 305 mm long, and it was oriented with its longitudinal axis perpendicular to the tibia. The impact velocity was approximately 3.5 m/s. The linear impactor was instrumented with a load cell and an accelerometer. The accelerometer was used to correct the measured force for the inertia of the mass located in front of the load cell. In addition, an LVDT measured relative displacement of the loading head.

The tibia was simply supported near the ends against the tibia load cells (mock load cells were used in place of the original load cells) approximately 250 mm apart. A portion of the tibia was wrapped with flesh to provide a realistic impactor-to-tibia interface. Tibia deflection was measured with an LVDT (contact duration was measured with a tape switch). Figure 7 and figure 8 show the tibia response under static and dynamic loading and compare them to PMHS targets. It is seen that there is good agreement for static loading conditions and fair agreement under dynamic loading conditions. The intent was that the force-deflection characteristics should agree up to the point of fracture.





Figures 7 & 8. Comparison of a typical quasi-static and dynamic test with PMHS.

LOWER LEG TESTING - A series of tests were conducted with the new lower leg to compare the response of the knee with the corridors presented by Matsui [1999]. The knee response tests are somewhat similar to the test setup described above for testing the polyurethane surgical knee model and are based from the tests performed by Kazjer [1997]. The test fixture was modified to allow testing of the leg assembly from mid-femur to foot.During the initial testing, a rigid, aluminum tibia was used, rather than the compliant tibia, since the response characteristics of the latter in full-scale testing had not been totally determined. Other differences with the PMHS test setup was the mass of the impactor and the foam placed in front of the impactor. The impactor mass in the Kajzer tests was 6.25 kg. Because of limitations of the impactor available, the smallest impactor mass available was about 8.5 kg. Because of the mass difference, the impactor speed was lowered so that the total input energy was the same. In the Kajzer tests, a 100 mm x 120 mm x 50 mm rectangular piece of Styrodure crushable foam was used to minimize damage at the contact area. At the time of testing, this material was not available, and a substitute Styrofoam with similar force-deflection characteristics was used. Figures 9 and 11 illustrates the shearing and bending test setup for the PMHS tests. Figures 10 and 12 illustrates the shearing and bending test setup for the new leg.

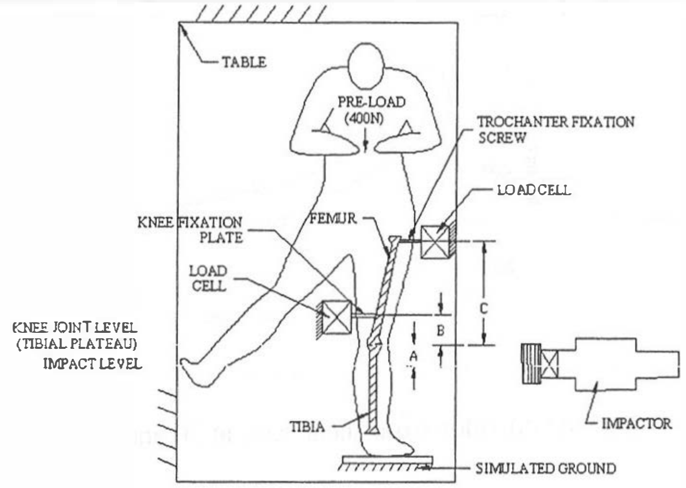


Figure 9. PMHS shearing test setup.

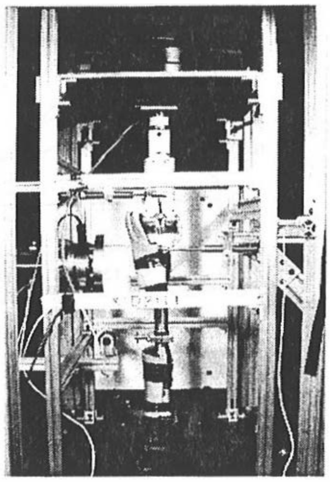


Figure 10. New leg shearing test setup.

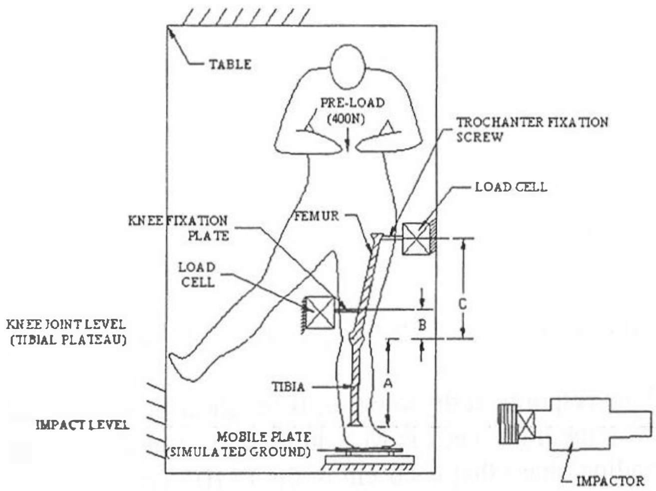


Figure 11. PMHS bending test setup.

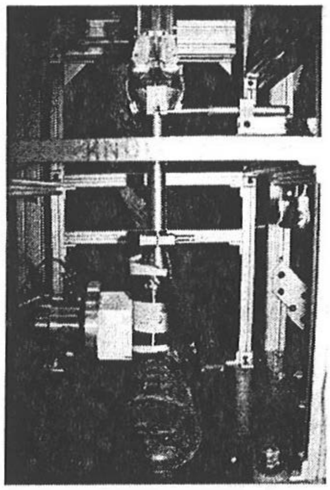
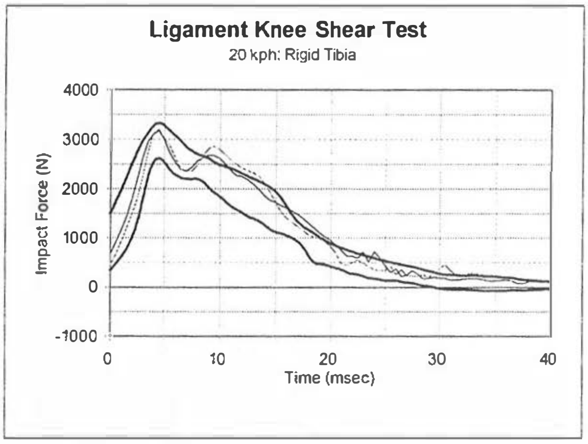
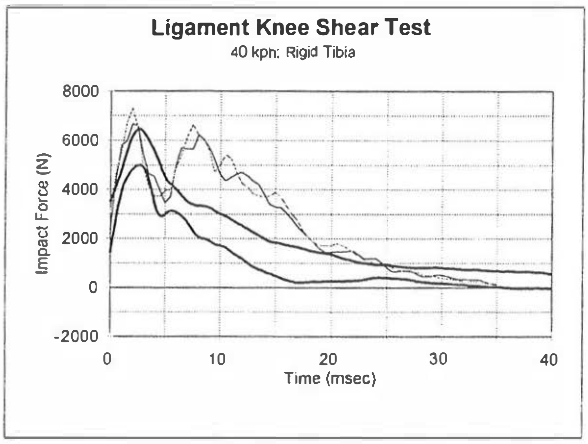


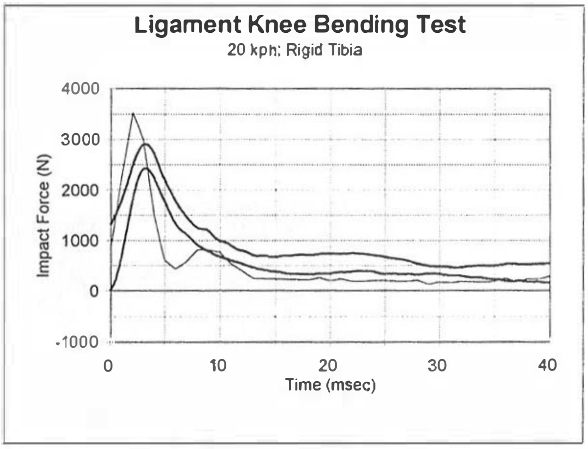
Figure 12. New leg bending test setup.

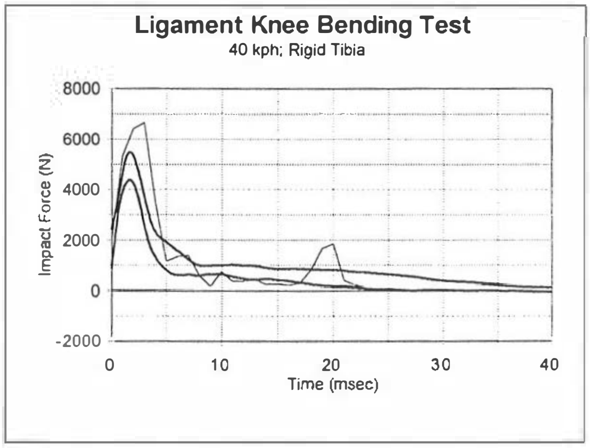
Lower Leg Test Results - Both shearing and bending tests at the nominal impact speeds of 20 km/h and 40 km/h were conducted on the new lower extremity. The results from the tests are shown in the following graphs.





Figures 13 & 14. Comparison of impact force with corridor from shear tests at 20 and 40 km/h.





Figures 15 & 16. Comparison of impact force with corridor for bending tests at 20 and 40 km/h.

It is seen that there is good agreement of the response at the low speeds for shearing impact, and the primary and initial peak of the shear response at the high speed is well duplicated. But for the high speed, there is a secondary peak during the unloading phase that is absent in the PMHS corridor. The source of the secondary peak is probably the non-destructive nature of the dummy knee. During a PMHS test, there would probably be ligament damage following the primary peak that would lower the subsequent forces seen in the system. In the mechanical knee, the spring/tubes probably begin to reload following the initial shearing effect when a femoral condyle is able to overcome the resistance of the intercondylar eminence, which results in the secondary peak. In the high speed test, the primary peak is seen to be somewhat above the corridor.

In the bending tests, there is no secondary peak seen, since in this case the curved surface of the condyle rotates more smoothly around the mating concavity of the meniscus. Thus there is good agreement in timing and shape of the response of the knee with the PMHS corridors. Here also, the maximum force seen in the high speed test is somewhat higher than the corridor. It is thought that part of the reason for the higher forces is the use of the rigid tibia. Future tests are planned using the deformable tibia. It is thought that the magnitudes of the peaks for the high speed will be lowered when such a tibia is used. In addition, the tension in the cables attached to the spring/tubes for each ligament can be adjusted to provide some tuning.

DISCUSSION AND CONCLUSION

The results from both the leg impact tests and the separate tibia loading tests indicate that the new knee is performing well. The response curves from the leg impact tests, show that the general features of the PMHS response are well duplicated, include peak timing, duration of the principal loading curve, and the general magnitude of the peak force. The forces generated for the high speed shear and bending tests are somewhat higher than the PMHS corridor, but it is felt that this can be controlled by relieving the tension in the spring and cable system that represent each ligament and by the use of a deformable tibia.

The tests on the tibia component indicate that the stiff Urethane with a Kevlar interior provides a good model for the response characteristics in static and dynamic three-point loading tests. The tests at high speeds show that the tibia is durable and repeatable under multiple impact conditions at high speed up to 7.5 m/s.

Full-scale tests using the new knee and tibia structures have been recently concluded at JARI. Data from these tests indicated that these new components are durable over the whole test series (of about 10 tests) including a high severity test at 50 km/h. Details of these tests will be presented in the near future. We believe that the knee and tibia components can be used successfully in both the pedestrian dummy and should be useable as part of a legform in component testing.

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