Anterior Tibia Impacts: A Biofidelity Study between Post-Mortem Human Subjects and Anthropomorphic Test Devices

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ABSTRACT

Injuries to the posterior cruciate ligament (PCL) commonly result from frontal automobile crashes when the occupant's tibia interacts with the intruding knee bolster of the vehicle. The goal of this research study was to better understand injury to the PCL due to this impact scenario and correlate injury with the response of the knee slider of today's crash test dummies. Over a span of two years, fourteen post-mortem human subject (PMHS) lower extremities were impacted on the anterior aspect of the tibia, just distal to the tibial tuberosity. The impactor was a 24 kg pneumatic ram with a padded face to simulate the stiffness of a knee bolster. The PMHS tests revealed the stiffness of the knee to be a viscoelastic response and the tibia could displace almost 23mm in relation to the femur before the occupant had a 50% chance of tearing their PCL. Following the PMHS testing, five anthropomorphic test device (ATD) lower extremities were subjected to similar impact conditions. The tested ATDs included the 5th female, 50th male standard HIII legs with the ball bearing slider and linear slider, the 5th female FLX and 50th male LX lower extremities. The testing revealed that all of the ATD legs were repeatable and the knees had similar stiffnesses no matter what impact energy level was used. The displacement between the tibia and the femur in the ATDs never reached more than 15 mm.

INTRODUCTION

Knee injuries account for 10% of all injuries sustained in frontal collisions, according to a study using the NASS database (Atkinson 2000). More specifically, knee ligament and tendon injuries are 2.5% of all types of injuries sustained in frontal crashes (Atkinson 2000). This typically occurs as the driver or front seat passenger continues to move forward as the vehicle slows down. The knee bolster of the car comes into contact with the occupant's tibia and the tibia displaces posteriorly relative to the femur. From this impact, the occupant can sustain fractures to the femur, tibia and fibula or injuries to the knee ligaments, particularly the posterior cruciate ligament (PCL). Although knee injuries are typically not life threatening, the injuries can be debilitating and the rehabilitation is expensive (Kuppa 2003).

Initial research on the interaction of the proximal tibia and the vehicle's knee bolster was done by Viano et al. (1978). Following a dynamic impact to the knee joint, five post mortem human subject (PMHS) knees that had no apparent injuries were loaded at a quasi-static rate while the force and deflection values were recorded. The average recorded deflection of the tibia in relation to the femur was 14.4 mm and a stiffness of 1.49 N/mm was calculated. From this testing, an acceptable stiffness for the ATD knee slider was determined to be between 1.26 and 1.72 N/mm. Mertz et al.(1989) scaled this stiffness range for both the 5th percentile female and the 95th percentile male. One limitation of the work by Viano is that the experimental set-up resulted in loading conditions not commonly found during real world automobile crashes. To provide a more realistic loading during laboratory testing, Bartsch (2004) used frontal crash test data to determine the input parameters to the ATD lower limb during the crash. From this data, an isolated PMHS lower limb test was developed that would mimic real world loading conditions. The isolated PMHS tibia was impacted below the tibial tuberosity, while padding was placed on the ram face to match the stiffness of a common knee bolster in a vehicle. These loading conditions were the basis for a study of 14 PMHS lower limbs by Fountain (2007). The PMHS test setup is shown in Figure 1.

Figure 1: PMHS test setup

This study will look at the response of five ATD lower limb models impacted using a similar experimental setup as the PMHS testing done by Fountain. This paper focuses on the initial results of the PMHS and ATD analysis.

METHODS

Five ATD lower limb models were used in the testing: Hybrid III 5th female leg with a ball bearing slider, Thor FLx Leg, Hybrid III 50th male leg with ball bearing slider, Hybrid III 50th male leg with linear slider, and Thor Lx Leg. For the testing, tri-axial accelerometers (Endevco, San Juan Capistrano, CA) were secured on the ATD leg on the distal end of the femur and on the superior aspect of the tibia, beneath the outer skin of the ATD legs. These locations mimicked the anatomical positions of tri-axial accelerometers used in the PMHS testing. The skin was replaced for testing. A six-axis load cell (RA Denton Inc, Rochester, MI) was placed behind the femur to measure the loads and moments that were transmitted through the ATD knee. Figure 2 shows the accelerometers and the load cell with the skin retracted.

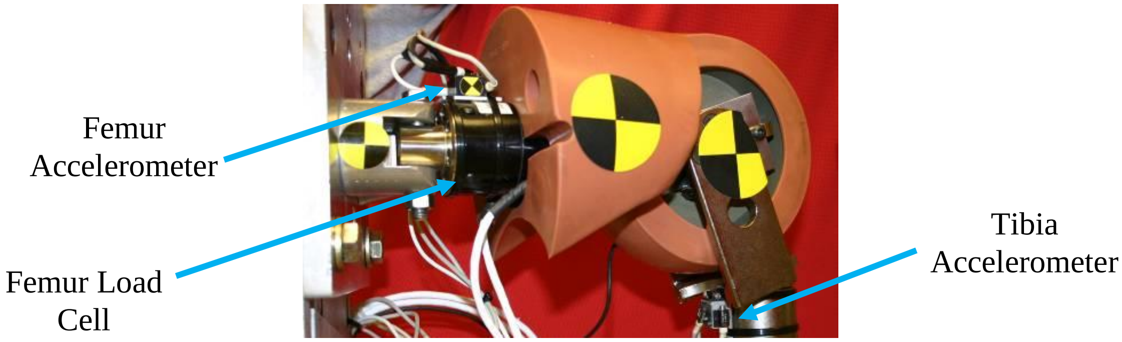

Figure 2: Location of instrumentation for ATD testing

The ATD leg was attached to the test PMHS fixture as shown in Figure 3. The ATD knee slider was used to measure the relative tibia displacement. Photo targets were place on the ATD leg to track the relative motion between the tibia and the femur during the impact by high speed cameras which recorded at 1,000 frames per second. Similar to the relative lower limb positioning in full body testing, the test fixture held the femur at six-degrees of medial rotation about the z-axis. The leg was adjusted so the skin over the anterior tibia formed a 90 degree angle with the ground. A foot strap was used to secure the foot during testing and load cells were used to measure the force through the heel in both x-axis and z-axis directions during testing. Once the leg was in position, points on the leg and instrumentation markers were digitized using a FaroArm (FARO Technologies Inc, Lake Mary, FL).


Figure 3: Experimental test setup for impacts of ATD lower limbs

The ram was positioned to impact the tibia in a similar location to the tibial tuberosity used on the PMHS lower limbs. For each test, a piece of Last-A-Foam® FR-7104 (General Plastics Manufacturing Company, Tacoma, WA) padding was attached to the ram face, as was used in the PMHS testing. During the impact, the data was collected using a 32-channel data acquisition system (Yokogawa Corporation of America). A sampling rate of 20,000 Hz was used and the data was filtered and zeroed according to SAE J211 guidelines. The impact speeds for each lower limb ATD model are given in Table 1.

Table 1: Test Matrix for ATD testing

ATD Model	Size	Type of Slider	Test Velocities (m/s)					
			1.4	2.9	3.2	3.5	3.9	4.2
Hybrid III Dummy	5th Female	Ball Bearing	x	x	x	x	x	
	50th Male	Ball Bearing	x	x				x
		Linear Slider	x	x				x
Thor Dummy	5th Female (FLx)	Ball Bearing	x	x	x	x		
	50th Male (Lx)	Ball Bearing	x	x				x

For analysis and comparison of stiffness between the PMHS and ATDs, the internal knee force was calculated. For the ATDs, the internal knee force was considered to be the force measured in the femur load cell because no other forces acted on the femur of the ATD during impact. For the PMHS, internal knee force in the x-direction was calculated as follows:

$$\textit{Internal knee force} = \textit{Force in femur load cell} - \textit{Force in quadriceps} \bullet \cos(\textit{quad angle})$$

The relative tibia displacement was determined for both the ATDs and the PMHS using image analysis software to track the photo targets on the anterior tibia and the femur load cell. Additionally, the ATD displacement was measured using the knee slider transducer.

RESULTS AND DISCUSSION

Displacement Comparison

The relative tibia displacements of the ATDs are shown in comparison to the tested PMHS lower limbs in Figure 4. The PMHS bars are also shown with plus or minus one standard deviation markings and the injurious results are the peak displacement up to time of injury.

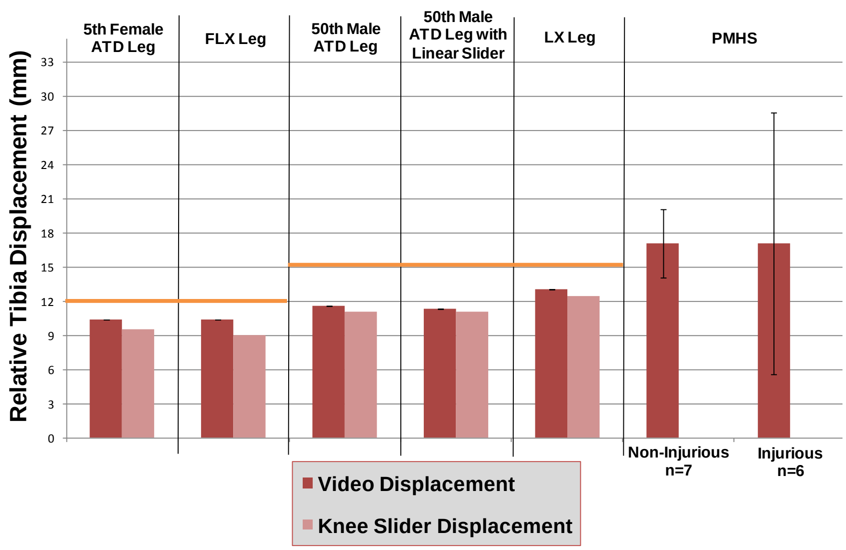

Figure 4: Relative Tibia Displacement of ATD and PMHS at an impact speed of 2.9 m/s, displacement measured using photo targets

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From this graph, it can be seen that the knee slider displacement is consistently slightly lower than the video data due to the soft skin compression on the ATDs. The orange line on the graph at 12 mm and 15 mm is a proposed value of knee slider displacement for the 5th and 50th ATD, respectively, to indicate injury to the knee complex (Mertz 1984). At 2.9 m/s, none of the ATDs reached the 12 mm or 15 mm of displacement as measured by the knee slider transducer or by the video analysis. However, the PMHS lower limbs saw an average peak displacement higher than 17 mm and six of the thirteen subjects were injured from the impact. The PMHS from the impacts completed by Fountain (2007) also had higher displacements than the 14.4 mm of displacement measured in the knees loaded at quasi-static rates by Viano et al. (1978).

Stiffness Comparison

During the PMHS testing by Fountain, different loads were applied to the quadriceps muscle during the impact to simulate in vivo loading on the muscle and to hold the patella in place. To explore how the quadriceps load affects the knee stiffness, the internal knee force vs. displacement curves were plotted for each load as shown in Figure 5. Similar to the method used by Shaw et al. (2006) for thoracic stiffness, targets for each load were also plotted using the standard deviation for the displacement and internal knee force. One PMHS test did not have displacement data due to a camera failure and is not included in the comparison. As the load on the quadriceps muscle increases, the average stiffness increases. However, the standard deviation for the 222 N load tests shown in green, overlaps significantly with the 111 N and 444 N load tests. Since there are no distinct separations between the test conditions, all will be considered together for analysis.

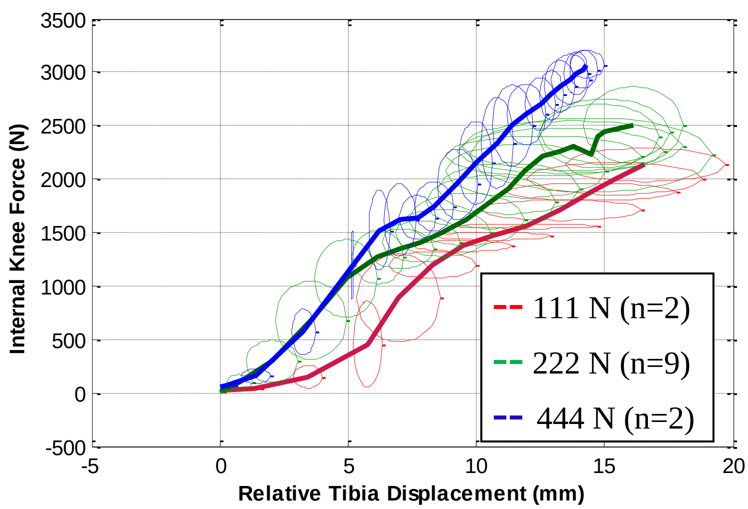

Figure 5: PMHS Corridors for various quad loads at an impact velocity of 2.9 m/sec

Figure 6 and Figure 7 show the targets created from the PMHS data plotted against the stiffness of the 50th and 5th ATD knee sliders, respectively. For simplicity, a line showing the average value for the PMHS was not plotted on Figure 6 and Figure 7. It should be noted that the impacts plotted were all conducted at 2.9 m/sec, an impact velocity Bartsch (2004) documented to cause responses in the knee to be similar to real world loading conditions.

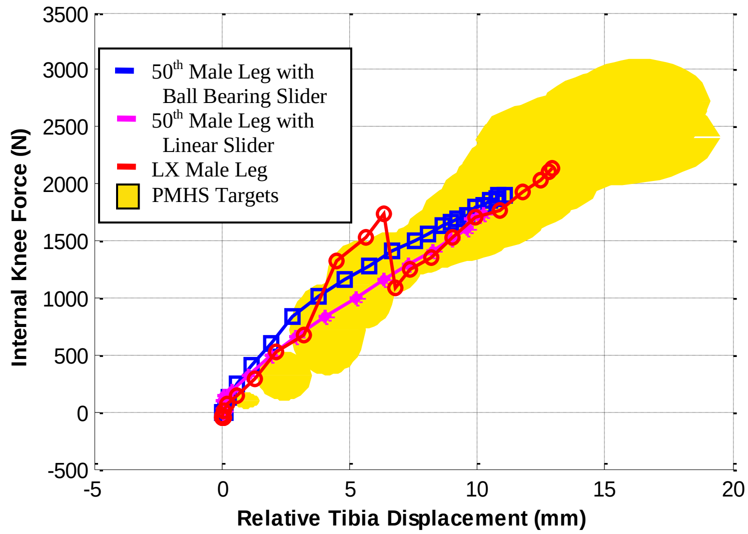

Figure 6: Force vs. displacement target from all PMHS lower limb data versus 50th male ATD lower limbs

Although the initial stiffness of the ATDs was higher than for the PMHS, it can be seen that the ATD stiffnesses are similar to those observed in the PMHS. However, the forces and displacements are lower for the ATDs than for the PMHS. While the Hybrid III legs are nearly linear, the LX leg shows a sharp drop in force around 6 mm of displacement. It is hypothesized that there is yielding in one of the elements of the leg but further investigation is required for conclusive results.

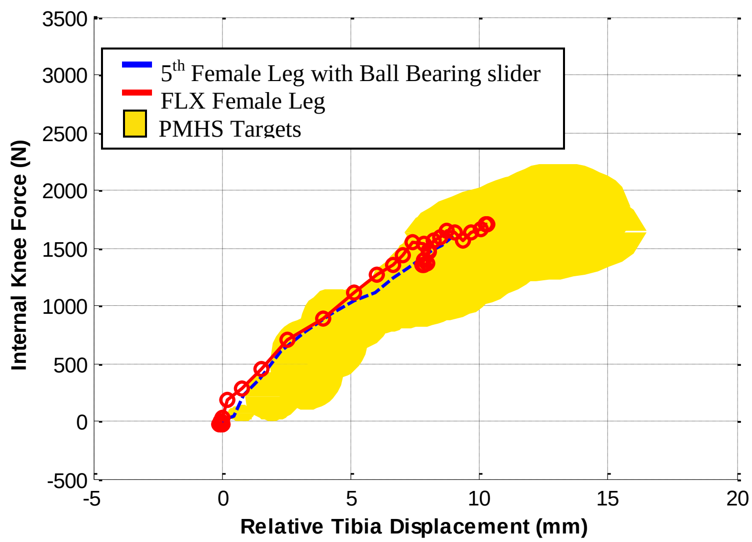

Figure 7: Force vs. displacement target from PMHS lower limb data scaled to a 5th percentile female versus 5th female ATD lower limbs

The above figure shows the targets from all the PMHS tests with the average values scaled for the 5th female using the method detailed by Mertz et al. (1989). It was assumed that the standard deviation for the 5th sized female was the same as the 50th sized male standard deviation. The plot again demonstrates that the stiffnesses are similar between the PMHS and the ATDs.

Visco-elastic Response Comparison

The response of knee ligaments has been shown to be visco-elastic and rate dependent (van Dommelen 2005). Therefore, it was hypothesized that the stiffness of the PMHS would show a similar visco-elastic response. While there was a limited number of samples, Figure 8a shows the average response of five PMHS lower limbs at impact speeds of 2.9, 3.2 and 3.5 m/s and two of the same legs at 4.8 m/s. The graph shows a general trend of increasing stiffness at higher impact speeds. In comparison, Figure 8b shows the response of the 5th female ATD with a ball bearing slider. The stiffness at these same increasing impacts speeds is nearly identical. A similar trend was seen in all the models of ATD lower limbs that were tested.

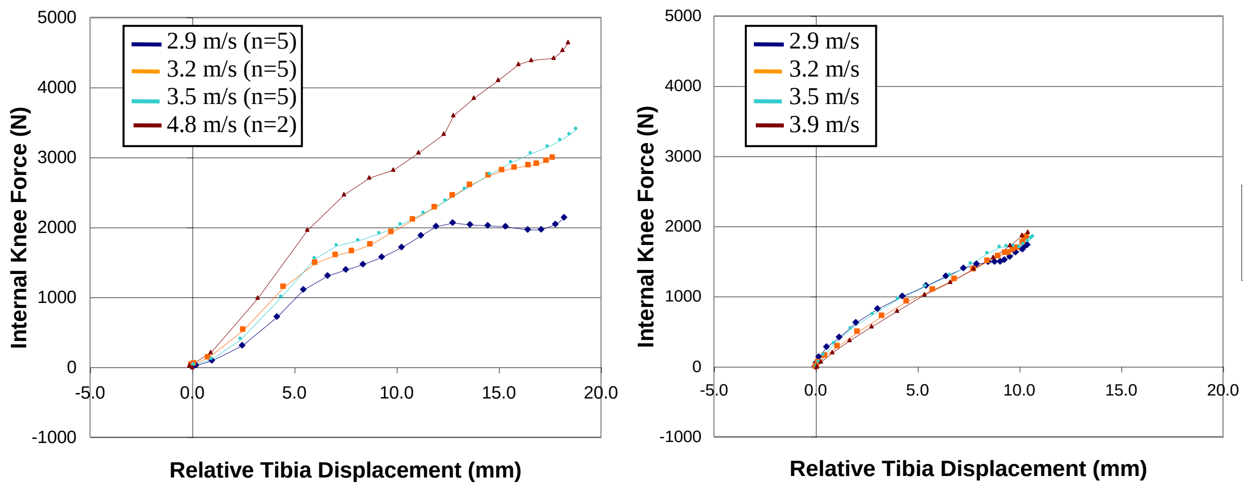

Figure 8: (a) Average Force vs. Displacement for PMHS Lower Limbs
(b) Average Force vs. Displacement for ATD 5th Female Lower Limbs

Limitations and Future Work

One limitation of the work is that the PMHS data was not normalized. Initial efforts at standard normalization techniques of the PMHS data such as those detailed by Mertz et al. (1989) and Eppinger et al. (1984) have not shown any reduction in variation. Further work is needed to determine a better normalization technique.

Another limitation of the ATDs, the knee sliders were difficult to calibrate and four out of the five models of legs had slightly higher stiffnesses than the calibration specifications. More testing with the same setup will be done using knee sliders that pass calibration. Additional research should also be done to understand why the PMHS and ATDs show similar stiffnesses but different displacements under the same impact conditions.

CONCLUSIONS

From this research, the following conclusions can be drawn:

- At 2.9 m/s impact velocity, the 5th female and 50th male ATDs did not reach the 12 mm and 15 mm displacements, respectively, which were proposed as indications of injury to the knee complex. However, six out of thirteen PMHS subjects had injuries at this impact velocity.

- The stiffnesses of the PMHS and ATDs were similar but the ATDs did not displace as much or see as much force. The transmission of force through the ATD leg should be further investigated.
- The PMHS tests showed a visco-elastic response so the impact velocity is important to consider when determining an injury criteria.

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