REQUIREMENTS FOR THE EVALUATION OF THE RISK OF INJURY TO THE ANKLE IN CAR IMPACT TESTS

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ABSTRACT

Injuries to the lower leg are still a frequent occurrence in frontal crashes and the most serious injuries have been found to be associated with the ankle region (pilon, calcaneal and talar neck fractures). These injuries are not only of a high severity, they are also associated with long term impairment, which contributes significantly to the societal cost associated with road traffic accidents. In order to reduce these injuries, the ability to determine the potential injury risk in legislative crash tests as well as the capability to assess the performance of proposed enhanced safety measures in the vehicle footwell region is essential. If this is to be achieved a biofidelic assessment tool with appropriate injury criteria is required.

In Europe, the protection afforded against injury in frontal impacts is currently assessed by the Hybrid III dummy in an offset deformable barrier test. For the lower leg the tibia index injury criterion is used, however this relates primarily to the risk of tibia fracture and is not appropriate for the determination of the risk of injury to the ankle.

This paper reports an overview of the biofidelity of existing dummy legs and the results of a series of PMHS tests which have recreated in the laboratory the more serious ankle injuries seen in real world crashes. The limitations of this work in terms of its application to an ankle injury criterion using peak tibia force for the Hybrid III are discussed and proposals for future work using a different approach are made in order to obtain these important criteria.

The results show the significantly enhanced biofidelity of the THORLx lower leg compared to the Hybrid III leg. The paper concludes that the early introduction of a more biofidelic leg such as the THORLx lower extremity is essential if additional and reliable lower leg criteria are to be usefully implemented.

INTRODUCTION

Vehicle safety can be improved through legislative crash safety requirements as well as consumer testing whereby the results of independent crash tests are published and made available to the general public. In order to ensure that vehicle safety is optimised as far as possible for occupants and pedestrians it is important that the performance measures used to assess vehicles in crash testing are continually improved and relate to real world conditions. These improvements need to reflect changes in vehicle design, advanced safety features and other technological developments, as well as ensuring that the test tool used to assess performance is based on the best available data.

Traditionally the focus of safety improvement has been to reduce the numbers of fatal and life threatening injuries sustained by car occupants, and in recent years the advancements in occupant protection through improved crash energy management, seatbelt and airbags have resulted in reductions to serious head, neck and torso injuries. However, the high societal costs associated with road traffic accidents (both economic and social) are not only related to injury frequency and severity, the impairment resulting from some non-life threatening injuries is becoming more significant as people are now surviving accidents that in the past may have proved fatal.

Accident data suggest that injuries to the lower leg are common and are significant in terms of societal costs associated with road traffic accidents [1-4]. Previous studies have indicated that ankle injuries such as pilon, talar neck and calcaneal fractures are the most important injuries for prevention based on their severity and impairment [5-7]. This is compared with injuries such as malleolar fractures which, despite occurring more frequently, generally result in a good clinical outcome for the patient with minimal or no long-term impairment [8].

In order for the risk of lower leg injury to be assessed accurately in legislative testing, a biofidelic dummy leg and appropriate injury criteria are required. The Hybrid III dummy, which is currently specified in ECE Regulation 94 for frontal impact, has a non-biofidelic lower leg and, in addition, the only lower leg assessment criterion in the test is tibia index, which was not intended for the measurement of ankle injury risk.

This paper reports on two phases of an on-going research programme examining lower leg injuries.

The first part of the work examines the biofidelity of current dummy legs, while the second part reports on a programme of work investigating the mechanisms and thresholds of ankle injury using PHMS specimens. The ethical approval for all of the PHMS work in this paper was granted by the relevant organisations in the UK.

BIOFIDELITY OF DUMMY LOWER LEGS

In a previously published paper [9] a set of biofidelity corridors were developed from the series of low energy dynamic impacts tests on PMHS legs. The test procedure used was based on the EEVC 'Tibia and Foot Certification Tests for use with the Hybrid III dummy in the EEVC Offset Deformable Front Impact Test Procedure' [10]. A simple rigid pendulum impactor instrumented with a single axis accelerometer was used to perform the dynamic impacts in this programme (Figure 1).

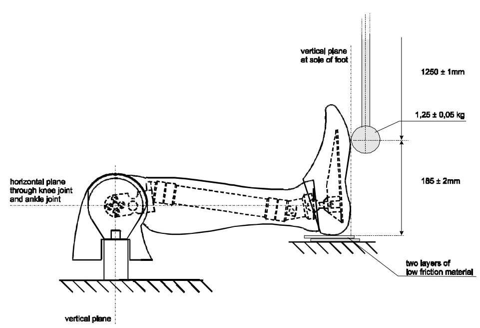


Figure 1. Lower Leg Biofidelity Test Experimental Set-Up.

Impacts were performed to the plantar surface of the foot, at the level of the ball of the foot (toe impacts) and the heel. In order to stay below injury threshold, toe impacts were performed at 2, 4 & 6m/sec, and heel impacts were performed at 2 m/sec and 4 m/sec only.

In addition to flaccid PMHS tests, toe impacts were also performed on instrumented PMHS specimens with an artificial Achilles tension applied (approx. 960N). The muscle group was replaced with a custom built tensioning device [11], and the Achilles tendon was loaded to represent the effect of passive muscle tension. The Achilles loads chosen for the work were based on previous published work [11], and a series of simulator trials [12]. The foot was externally supported, in order to restrict the plantarflexion motion the foot caused by the applied Achilles tension. This was achieved by the use of a stirrup around the forefoot.
As a supplement to the PMHS study, tests were conducted with six different volunteers, aware and unaware of the impending impact. For the aware impacts, the volunteer was asked to resist against the impacting pendulum without plantar-flexing the foot prior to impact. The foot was not restrained in any way to resist against bracing.

In each PMHS tests an implantable load cell was used to record the forces and bending moments in the tibia (Fx, Fy, Fz, Mx, My), in addition the pendulum acceleration and the angle of dorsiflexion were recorded. For the volunteer tests, only the pendulum response and dorsiflexion of the foot were measured.

Dummy Legs

In order to make an assessment of dummy leg biofidelity, comparative toe and heel impacts were performed on the Hybrid III lower leg fitted with a 45° articulating ankle and soft bump stop (specified in current European legislation (Regulation 94), the Hybrid III lower leg fitted with the GM/FTSS ankle and foot, and two Thor-Lx prototypes. Three tests in each condition were performed with a recovery period of 30 minutes between each test. The leg was attached via the knee clevis to a rigid back-plate to make the tests directly comparable to the PMHS tests.

Comparison of Dummy Legs with Biofidelity Corridors

The biofidelity corridors produced were for tibial force (Fz), tibial bending moment (My), and pendulum acceleration (Ap). In a comprehensive accident analysis [13], axial loading was highlighted as a significant cause of injury for the most disabling of ankle fractures (pilon, calcaneal and talar neck). Emphasis has therefore been placed on tibia force and, in particular, the magnitude of the peak for the assessment of biofidelity. The pendulum acceleration response gives an indication of the interaction of the foot with the vehicle pedal and floorpan, and therefore is also important in terms of biofidelity. The bending moment response is an additional method of assessing biofidelity and could potentially be used for injury prediction. The kinematic response of the ankle in terms of dorsiflexion indicates the joint stiffness and therefore the peak dorsiflexion during toe impacts was also assessed.

Toe Corridors (Flaccid PMHS) Figure 2 shows the tibia Fz response for the different dummies in a toe impact at 6m/sec compared with the corridor for flaccid PMHS. The responses of both prototypes of Thor-Lx were close to the flaccid PMHS corridors for tibia force. The response was good in terms of peak force but occurred slightly earlier than that in the PMHS specimens and is consequently outside the corridor.

In contrast, the tibial forces measured by both the Hybrid III and the GM/FTSS legs were very different from the corridors in terms of their overall shape and magnitude. One of the fundamental differences between the human leg, compared to the design of the Hybrid III, is that the dummy tibia is offset in its alignment between the knee and ankle joints. This aspect of the dummy design has historically been an important factor in the interpretation of the load cell data. This offset has been corrected in the Thor-Lx, which may partially account for the notable improvement in response.

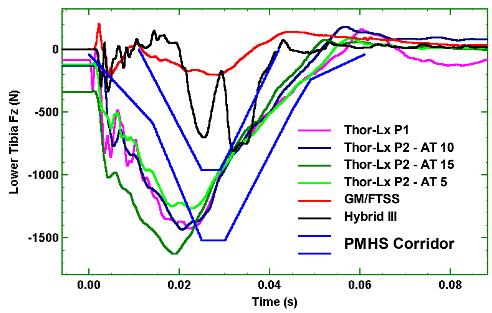


Figure 2. Tibial Force for Toe Impact (6m/sec) with Flaccid PMHS Corridor

None of the dummy legs tested was within the flaccid PMHS corridor for pendulum acceleration (see Figure 3), as all exhibited substantially higher peak responses, although the Thor-Lx was the closest to the corridor.

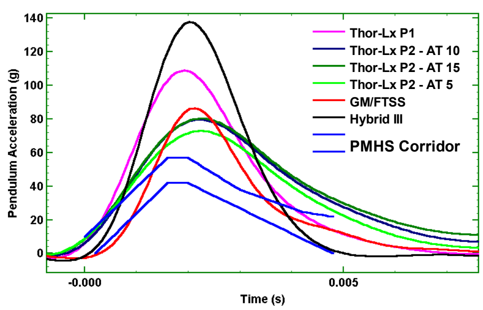


Figure 3. Pendulum Acceleration for Toe Impact 6m/sec with Flaccid PMHS Corridor

For tibia bending moment, the GM/FTSS foot and ankle was the only dummy leg to show some correlation with PMHS specimens and the mean response fitted within the first section of the corridor only.

Toe Corridors (PMHS with Achilles Tension) The same dummy results were compared with the corridors for PMHS with an applied Achilles tension and both prototypes of Thor-Lx were found to be close to the peak value of the corridor for tibia force (see Figure 4). However, the pulse duration for the dummy component was longer than for the PMHS results. Again it was observed that the pulse shape and duration for both the Hybrid III and GM/FTSS was very different to the corridor.

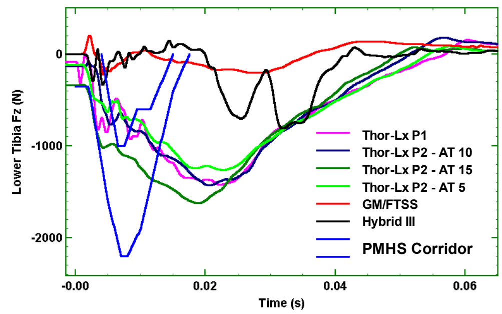


Figure 4. Tibial Force for Toe Impact (6m/sec) with Applied Achilles Tension PMHS Corridor

For pendulum acceleration, the Hybrid III was almost within the corridor, but with the peak acceleration occurring slightly earlier than for the PMHS with applied Achilles tension. The response for the ThorLx and GM/FTSS were both significantly lower than the corridors. It was anticipated that the stirrup on the ball of the foot (used to constrain the plantarflexion of the foot in the PMHS test with applied Achilles tension) may have influenced the localised stiffness of the foot and, as such the corridors may be unrealistically high in this instance).

Volunteer corridors were constructed for pendulum acceleration only. In general, the dummy leg responses peaked before the volunteers. The GM/FTSS foot and ankle and both Thor-Lx prototypes were close to the corridors for aware volunteers and would fit within the corridors if they were time shifted (see Figure 5).

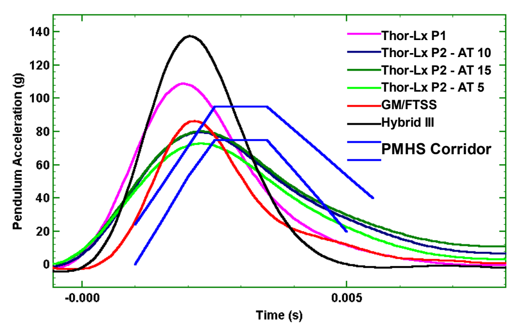


Figure 5. Pendulum Acceleration for Toe Impact 6m/sec with Aware Volunteer Corridor

Dorsiflexion The peak dorsiflexion for each of the specimen types was recorded in the toe impact tests (Figure 6). The PMHS specimens consisted of seven specimens that had been sectioned through the knee and one that had been sectioned above the knee and the response of each type is shown separately in the figure. An increasing trend of humanlike behaviour can be observed from the Hybrid III foot, to the GM/FTSS foot, then the through knee PMHS specimens, the above knee specimen, unaware volunteers, the Thor-Lx prototypes and finally the aware volunteers. The GM/FTSS foot exhibits very similar behaviour to the through knee PMHS specimens. In a similar way, the responses of the Thor-Lx prototypes show an extremely good correlation with the unaware volunteers and both are comparable to those of the above knee specimen.

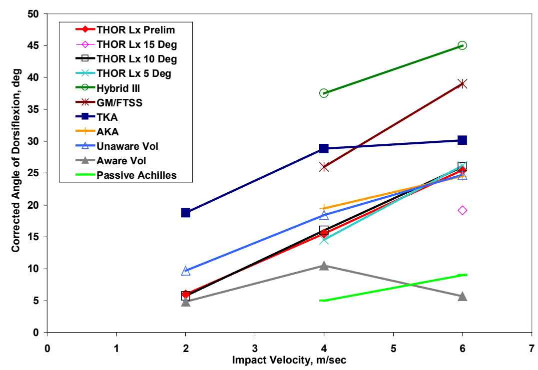


Figure 6. Maximum Dorsiflexion of Foot under Toe Impact

Although the GM/FTSS was close to the flaccid through knee PMHS results, both of the two existing dummies showed a higher dorsiflexion response than the remaining human surrogates. The kinematic behaviour of the Thor-Lx prototypes in dorsiflexion was considerably more biofidelic. The dorsiflexion results for PMHS with 960N of applied Achilles tension showed some correlation with the aware volunteers, but less motion was seen compared to unaware volunteers. It has been suggested that 960N may be too high as a representation of passive muscle tension. This may be so if all of the force transmitted to the foot through all of the tendons normally present is applied only through the Achilles tendon, which is not representative of the real-life situation.

Ankle stiffness and the tension in the lower leg plantar flexing muscle group influence peak dorsiflexion under impact to the toe. The stiffness of the ankle joint will determine, in part, the bending moment and force transmitted to the tibia. If these are to be recorded accurately in a dummy tibia then these data illustrate the importance of incorporating a biofidelic ankle in a car crash dummy.

<u>Heel Corridors</u> The dummy legs tested were compared only to flaccid PMHS specimens during heel impact, as Achilles tension has little effect on direct axial loading to the heel (Figure 7). Thor-Lx showed promising results for tibial force but peaked substantially earlier than PMHS specimens and consequently was outside that corridor.

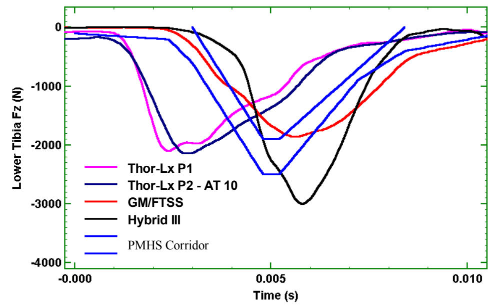


Figure 7. Tibial Force for Heel Impacts at 4m/sec with Flaccid PMHS Corridor

The GM/FTSS foot and ankle showed the closest response and almost fitted into the corridor for pendulum acceleration and tibial force. The responses of the Hybrid III and Thor-Lx were well outside the corridor for pendulum acceleration (Figure 8).

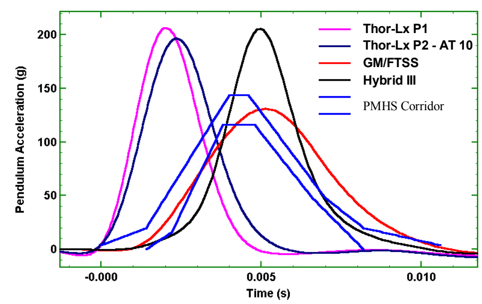


Figure 8. Pendulum Acceleration for Heel Impact at 4m/sec with Flaccid PMHS Corridor

Summary

Dummy leg response should ideally be validated against results from live volunteers. However, since it is not possible to obtain internal tibia force for volunteers, the dummy legs are also compared with data from instrumented PMHS specimens. Flaccid specimens do not exhibit the same biomechanical behaviour as live specimens due to the lack of physiological muscle function, and therefore the dummy response is also compared with impacts to PMHS with artificially applied PMHS.

For toe and heel impacts the Thor-Lx was found to be capable of recording tibial force more accurately than existing dummy legs. Tibial force is considered to be the most important parameter of those compared in the programme in relation to injury risk prediction. Looking at the pendulum acceleration for toe impacts, the Thor-Lx and GM/FTSS foot would appear to be most similar to the human in terms of the way in which the lower leg may interact with the toepan or pedals. For the pendulum response in the heel impacts the GM/FTSS foot was found to be closest to the corridors with both the Thor-Lx and Hybrid III having a much higher peak response.

In order to assess accurately the risk of injury in legislative tests, the biofidelity of he dummy is of crucial importance. The results of these low energy dynamic impacts tests indicate that the Thor-Lx more accurately represents the human lower extremity and that it has the additional benefit of a more comprehensive instrumentation package, compared with existing dummy leg.

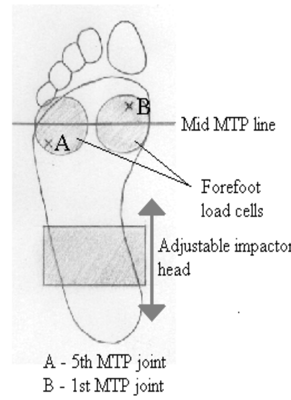
INJURY RECREATION TESTS

If injury risk is to be assessed reliably in impact tests, it is important to understand the factors affecting the generation of the priority injury types so that, where appropriate, the relevant parameters can be measured on a dummy or alternative test tool. The most disabling ankle injuries (calcaneal, pilon and talar neck fractures) are generally considered to result from primarily axial loading to the lower leg [13]. However it is not known whether different loading configurations result in different injury types. This part of the study was designed to determine whether there were important characteristics that needed to be recorded on a dummy in order to ensure a reliable measure of injury risk or whether axial force alone would be a sufficient measure of injury risk.

Test Protocol

For this test series a linear impactor test device was used to load the sole of the foot dynamically at a range of impact forces. A schematic of the layout of the linear impactor rig can be found at Appendix 1. The PMHS leg was potted in a mounting cup and attached via a knee clevis to a support frame. The PMHS foot was positioned and pre-loaded against three tri-axial load cells. Each load cell was protected by a nylon cap, which provided an appropriate contacting surface to position against the foot. The impacting force was applied through the lower load cell assembly. The upper two load cells were positioned against the forefoot. These remained static and measured the reactive force at the forefoot (Figure 9).

The position of the impactor load cell relative to the upper two could be varied to accommodate foot size and different impact positions. The impacting load cell was instrumented with an accelerometer and linear potentiometer. Between the foot and the load cells was interposed a 3mm thick rubberised sheet (Velbex) to reduce the risk of injury due to concentrated point loading and also to serve as a basic shoe representation.



(MTP= Metatarso-Phalangeal)

Figure 9. Position of Load Cells on Foot

Each PMHS leg was attached, via the mounting cup, to the support frame and was adjusted so that the tibia axis was parallel to the impactor with the ankle in the neutral position. Pre-loads were then applied to the leg to represent more realistically the forces experienced by the leg in a real world crash.

A hydraulic Achilles tendon tensioning device, which enabled a measurable artificial force to be applied though the Achilles, was used to simulate the braking force applied through plantarflexion of the foot on the brake pedal. In order to maintain the ankle at neutral and the heel in contact with the impactor head under the application of Achilles tension, the PMHS were pre-loaded via a 'jacking' plate located at the proximal end of the tibia.

Each leg was pre-loaded with 1.5kN-2kN of Achilles tension, which would equate to applying the brake pedal forces that were reported in paper on emergency braking trials using volunteers in a driving simulator [11, 12].

Instrumentation

Each of the three load cells in contact with the plantar surface of the foot were used to measured the load in three axes (x, y & z). A linear potentiometer was used to measure the displacement of the lower impactor head assembly. A single axis load cell, positioned between the 'potted' proximal end of the PMHS leg and the supporting frame, recorded the resultant load generated from pre-loading the tibia through the impactor rig's frame and from the simulated Achilles tension.

An acoustic transducer was also fitted to the distal tibia in order to help establish the time of failure. Goniometers were fitted to the ankle joint to measure the rotation, inversion/eversion and plantar flexion/dorsiflexion. The Achilles tension device was also fitted with a single axis load cell in order to quantify the applied Achilles loads to be quantified.

All data were acquired at 20kHz except for the acoustic transducer data, which were acquired at 100kHz. The sample rate and filtering was carried out in accordance with the guidelines set out in ISO 6487.2000 [14]. Each impact was recorded on high-speed film running at 400 frames/sec, to provide additional information about the kinematics of the foot during impact and to substantiate the goniometer measurements.

Each PMHS leg underwent a detailed anthropometric assessment and regional Bone Mineral Density (BMD) analysis before testing. The regional BMD studies measured the bone density in three places in the distal tibia, one approximation of the BMD for the whole foot and two measurements from the Calcaneus. Following each test the legs were examined, X-rayed and a necropsy was performed. Injuries were been described using the Orthopaedic Trauma Association (OTA) classification system [15].

Testing

Three test configurations were used:

- Position A - The impactor head was centred in line with the tibial axis (Figure 10)

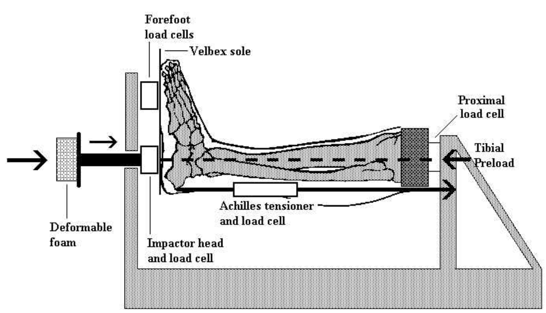


Figure 10. Position A

- Position B -The impactor was centred on the anterior tibial margin, parallel with the tibial axis (Figure 11)

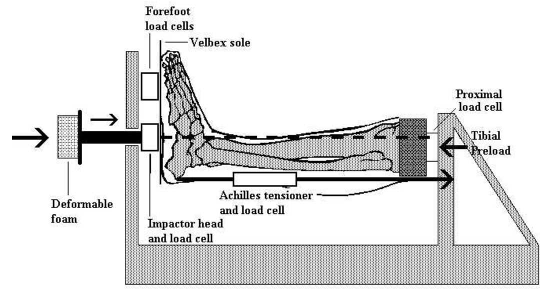


Figure 11. Position B

- Position C - The impactor was centred 2.5cm anterior to the anterior tibial margin, parallel with the tibial axis (Figure 12)

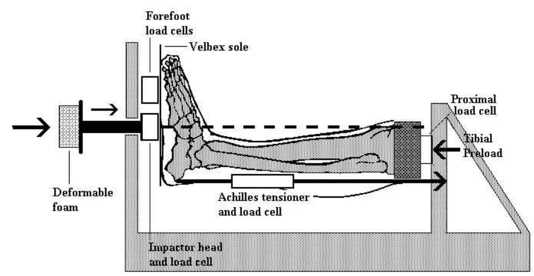


Figure 12. Position C

Results

A total of twenty-three PMHS legs were impacted in the three impacts positions A, B and C (n=3, 9 and 11 respectively). The injuries generated were nine inter-articular calcaneal fractures, one talar neck fracture, two talar body fractures, two malleolar fractures and three soft tissue injuries. The remaining three legs had no detectable injury. Table A1 in the Appendix summarises the main features of each impact test.

One of the calcaneal fractures in Position A and two of the calcaneal fractures generated in Position C were felt to have failed in tension, which is not the mechanism of failure associated with these injuries in automotive accidents. The other 14 injuries that were produced were considered typical of ‘real-world’.

Five legs were impacted twice as no injury occurred during the first impact. It should be noted that data from these impact tests have not been used in the injury analysis.

The processed data were analysed and, based on the acceleration profiles, peak loads and data from the acoustic transducer, it was possible to estimate the time of failure/fracture. Specimen13R sustained a calcaneal fracture when impacted in Position B. The three graphs in Figures 13-15 show the typical outputs from the instrumentation for each test.

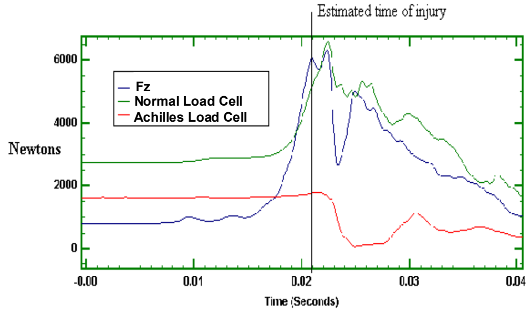


Figure 13. Forces Recorded from Impact during Calcaneal Fracture

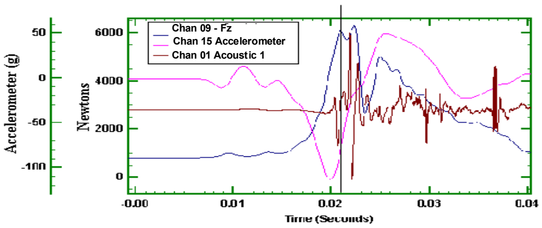


Figure 14. Accelerometer Data from Impact during Calcaneal Fracture

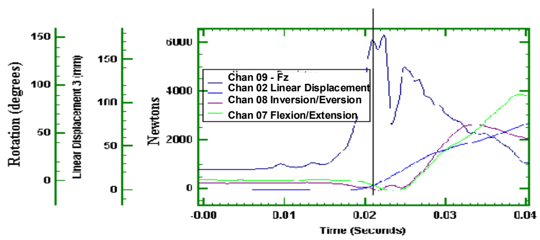


Figure 15. Displacement Data from Impact during Calcaneal Fracture

Analysis of Results

Table 1 shows the injuries in each impact position and compares the pre-impact and failure loads.

Position A generated two calcaneal fractures that were the result of direct loading to the calcaneus. Both fractures were sustained with high levels of pre-loading and involved minimal ankle rotation. Test 08R produced a 'clinical' two part articular fracture and the load cell signals were suggestive of a crushing process. Specimen 12R produced an unusual fracture, which was not typical of the high-energy calcaneal fractures seen in real world crashes. The high-speed film records revealed that this was most likely due to non-physiological pull by the Achilles tensioning device.

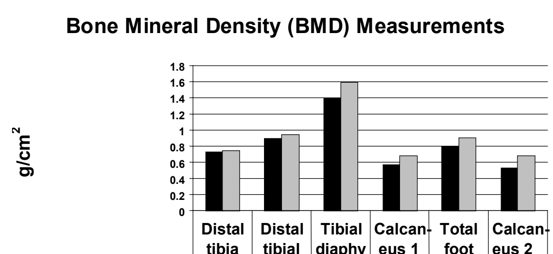
Table 1. Summary of PMHS Impact Test Results

Impactor Position		A	B	C
Number of tests		3	9	11
Mean Pre-load (N)	Forefoot	720	236	424
	Impactor head	199	825	705
	Proximal load cell	2212	2520	2786
	Achilles tension	1612	1342	1796
Mean Failure Load (N)	Impactor head	8622	4914	4468
	Proximal load cell	8146	6022	6365
	Achilles tension	306	1426	2309
Injuries		2 Calcaneal # 1 No injury	4 Calcaneal # 2 Pilon # 1 Talar body # 1 Soft tissue 1 No injury	3 Calcaneal # 2 Malleolar # 2 Soft tissue 1 Pilon # 1 Talar body # 1 Talar neck # 1 No injury

In both positions B and C, the loading of the hindfoot was less direct and therefore the contribution of simulated active plantarflexion was more significant. A review of the loading patterns, injuries and BMDs suggested that the results from positions B and C were comparable.

Considering only the injuries generated in positions B and C, the impact conditions of pilon and talar body fractures were compared with the group of calcaneal fractures. The conditions that resulted in proximal fractures e.g. pilon and talar body fractures were of particular interest. Previous axial loading studies have produced relatively few proximal fractures involving the talus and distal tibia but large numbers of calcaneal fractures.

The PMHS legs in the two group were noted to be of roughly equivalent size and mass however there was some difference in the average Bone Mineral Density (BMD) measurements (See Figure 16).



	Distal tibia	Distal tibial	Tibial diaphy	Calcaneus 1	Total foot	Calcaneus 2
Calcaneal fractures	0.732	0.895	1.39	0.569	0.803	0.533
Pilon + talar body fractures	0.748	0.94	1.594	0.686	0.902	0.68

Figure 16. Bone Mineral Density Measurements

The calcaneal fracture group had lower calcaneal BMD measurements than the pilon and talar body group. This difference was statistically significant (p=0.0029 for pilon and p=0.0038 for talar body).

There was a marked difference between the pre-loading and impact loading conditions for the different injury types (see Figures 17 & 18). The legs that had sustained pilon and talar body fractures on average had experienced a much higher pre-load. At the estimated time of fracture, the Achilles tension was found to be lower for the calcaneal fracture group.

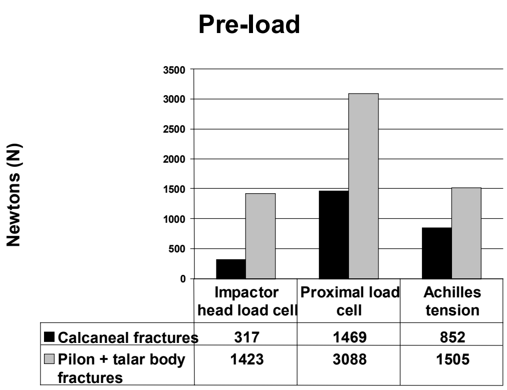


Figure 17. Pre-load Conditions of Leg prior to Impact

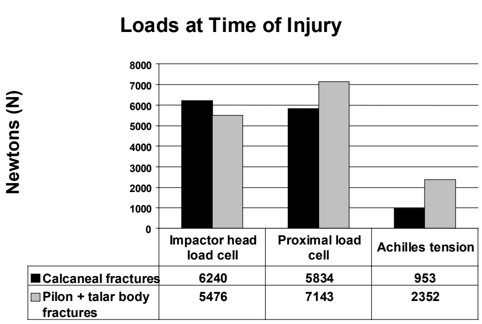


Figure 18. Loads at Time of Injury

Figure 19 indicates the ankle rotation and displacement at the time of injury. Rotations are described as positive or negative dorsiflexion and eversion. Negative eversion and dorsiflexion represents inversion and plantarflexion respectively.

Large ankle rotations and displacements were seen at the point of failure in the soft tissue group. The goniometer readings showed the ankle had dorsiflexed and everted at the time of injury. The PMHS ankles demonstrated gross disruption of the medial (deltoid) ligament complex and were in keeping with this mechanism of injury. Otherwise the injury groups showed, only small amounts of rotation prior to failure, and there was no significant difference in impactor head displacement or ankle rotation at the time of injury.

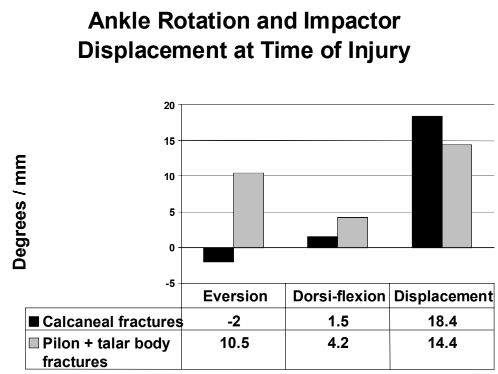


Figure 19. Ankle Rotation and Impactor Head Displacement at Time of Injury

Discussion

The fracture patterns and soft tissue injuries generated in positions A, B & C were generally clinically representative of the lower leg injuries seen in real world accidents.

The results have demonstrated that calcaneal fractures can result without direct loading to the body of the calcaneus. The results also indicated that, while Achilles tension might affect the calcaneal fracture pattern, the extent of this effect was difficult to determine. In contrast, pilon and talar body fractures occurred only in specimens where a high level of Achilles tension was maintained prior to and during the impact phase.

Only one talar neck fracture was generated in this test series. One mechanism that has been proposed for talar neck fracture is hyper-dorsiflexion of the ankle with an applied axial load to the plantar surface of the foot. This causes the talar neck to impact on the anterior tibial margin. The talar neck fracture in this study appeared to occur before hyper-dorsiflexion was achieved.

The results suggested that changes to the load distribution relative to the foot did influence injury outcome, but that the influence of Achilles tension was more significant with regard to the type of injury produced. For legislative test purposes, the dummy used will not be required to measure the risk of individual specific injuries, but rather the risk of any severely impairing lower leg injury. As such it was concluded that a requirement based on the axial load in the tibia would be an appropriate basis for ankle injury criteria. The orientation of the ankle joint may also be an influential factor in some cases (e.g. talar neck fracture) but the results of this study indicated minimal articulation of the ankle at the time of injury.

It has been demonstrated that the influence of Achilles tension on injury outcome is significant. Thus in a regulatory test where it is not practical to generate pre-impact braking forces in the dummy, the performance criteria will need to take account of the reduced additional force required to cause injury under pre-braced conditions.

Correlation of Dummy Response with PMHS Tests

The ability to predict severely impairing lower leg injuries is paramount in assessing safety of the footwell region of the car. To evaluate the injury risks in vehicle impact tests, the response of the dummy needs to be correlated to injury risks observed in PMHS specimens. As a paradigm of how future PMHS research tests may be used to develop injury criteria for the lower leg, the methodology below is proposed. A series of comparative tests on the Hybrid III leg were carried out in an attempt to correlate the forces recorded in the tibia with the injuries observed in the PMHS tests. (The Hybrid III was used as the Thor-Lx prototype was unavailable at the time of testing).

Method The leg was rigidly mounted via the knee clevis such that the line between the centre of ankle rotation, the knee clevis, and the axis of the impactor head, was horizontal (Figure 20). As the Hybrid III has no means of applying or measuring Achilles tension, the foot was placed in contact with the lower impactor head only, and was held in place by friction. No pre-load was applied.

Figure 20: Dummy Test Configuration

A 3mm thick rubberised sole was interposed between the sole of the foot and the impactor head to reduce point loading and to replicate the PMHS tests. The leg was impacted over a range of forces and velocities. Two impacts were performed under each test condition and a recovery period of 30 minutes was allowed between impacts. The impact force, Fz_3, was measured in the load cell impactor head. Tibial force and bending moments at the upper and lower tibia were recorded via a four-axis tibial load cell. The acceleration and displacement of the impactor head and velocity of the trolley were also recorded.

Hybrid III Results For tests with the Hybrid III, which is known to have limited biofidelity, the force recorded in the impactor head, Fz3, varied between 4809N and 10 909N and that measured in the lower tibia varied between 4973N and 10639N.

In order to be able to use these results to guide the development of performance criteria for impact tests, the injury risk needed to be correlated with the response of the dummy legs under the same input conditions.

With the experimental arrangement used in the research programme (attachment of the leg at the knee), the overall response of the dummy was vastly different from that seen in the PMHS tests. The tibia force was found to vary with the way in which the peak force was generated. A relationship was established between the output, Fz_{tibia}, the input force Fz_3, and the time to peak of the input force and this was used to develop the tibia force equivalent to a known applied force.

Table A1 in the Appendix includes the calculated equivalent tibial force measured in the Hybrid III for each of the PMHS tests. In each case the force predicted to have been measured in the Hybrid III tibia has been compared to the resulting PMHS injury from each of the tests.

A dose response statistical analysis was carried out to compare the force in the tibia with the injury in order to create and injury. Due to the limited sample size the sample was enhanced by making assumptions about the injury response of the samples at force inputs beyond one standard deviation of the fracture load. Data were selected only when the injury outcome was known over an entire range of force. The number of tests that fulfilled these criteria was 16. These data were fitted to a probit model and the resulting curve shown in Figure 21.

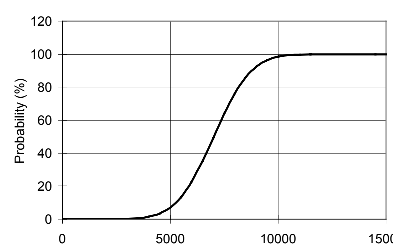


Figure 21. Injury Risk Curve Hybrid III

Limitations It should be noted that this was a first attempt at creating an ankle injury risk criteria for the Hybrid III. The methodology used in the PMHS tests did not lend itself to simple replication with the current dummy foot. Since the legs were effectively locked at the knee, the dynamic response of the system relied on the characteristics of that part of the body and no effect of inertia could occur because the legs were unable to move in response to the impact. Due to these and the assumptions made in the statistical analysis the results should be treated with caution.

SUMMARY

Biofidelity of Dummy Legs

- The response of the Hybrid III, GM/FTSS and Thor-Lx dummy components have been compared with PMHS and volunteer data from low energy dynamic impacts to the ball of the foot and heel.

Toe Impacts:

- Tibial force measured by Thor-Lx was more biofidelic than for existing dummy legs.

- The pendulum acceleration response for Thor-lx and the GM/FTSS foot were similar to aware volunteers.

Heel Impacts:

- THORLx was found to be the most biofidelic for tibial force (Fz), however the GM/FTSS was also quite close to corridor.

- The GM/FTSS response was closest to PMHS results for pendulum acceleration.

- THORLx & Hybrid III showed much higher peak response for pendulum acceleration. Some modification to the design may be required to address this.

General:

- THORLx was more biofidelic than existing dummy lower legs.

- Better biofidelity and comprehensive instrumentation allows a more accurate measure of injury risk.

- Potential to reduce incidence of severely disabling lower leg injuries in frontal crashes.

Injury Generation

- Twenty-three PMHS legs were impacted using simplified loading patterns designed to replicate those seen in real world frontal collisions.

- The injuries generated from the impacts were nine intra-articular calcaneal fractures; one talar neck and two talar body fractures; three intra-articular distal tibial (pilon) fractures; two malleolar fractures, and three soft tissue injuries. The remaining three legs had no detectable injury.

- Severe ankle and hindfoot injuries can occur without significant ankle rotation.

- Achilles tension was noted to be significantly higher in fractures affecting the talus and distal tibia compared to the group that sustained calcaneal fractures. It is proposed that Achilles tension pre-loads proximal structures through physiological pathways and reduces failure threshold. The influence of Achilles tension on load distribution across the foot and ankle increases with impacts anterior to the ankle centre.

- Fracture patterns were influenced by regional variations in the bone mineral density (BMD) of individual PMHS specimens.

- Severe ankle and hindfoot injuries occur under specific and individual circumstances which make it difficult to identify one parameter that will be predictive for these injuries.

- Comparative testing of the Hybrid III dummy leg has allowed an initial estimate of Hybrid III calibration against injury data. However, due to the need for the transfer function and the assumptions made, this should be treated with caution.

RECOMMENDATIONS

The biofidelity study has indicated that the Thor-Lx lower leg is the most biofidelic of the currently available dummy components, and that it also benefits from an enhanced instrumentation package. As such it is felt that its incorporation into legislative or consumer tests should be encouraged. However, suitable injury criteria need to be developed.

In the injury generation research programme presented here, the objective was to try to determine the critical features and loading regimes, which would result in the three main impairing ankle injury types. To control the test environment as much as possible, the PMHS legs were attached and prevented from translating at the knee joint. Reproduction of these tests with the Hybrid III leg resulted in unrealistic forces which would not have been reached in an in-vehicle situation since the foot and leg would have translated away from the loading surface before such high forces had been generated.

For the purposes of regulatory or consumer testing, it is not important to predict which of the major injury types will occur. It is sufficient to be able to measure the risk of any major injury in that region. The focus of the next phase of work will be the development of injury criteria for the ankle/lower leg will rather than the discrete injury mechanisms, which were the subject of this study. As such the test equipment will be designed to provide impact conditions which will allow the foot, ankle and leg to move in a more realistic manner.

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APPENDICES

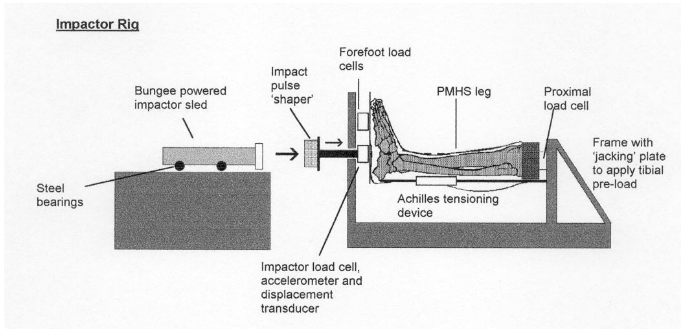

Figure 1a. Linear Impactor Rig Test Set-Up

Table A1.
Summary of PMHS Lower Leg Test Results and Loading Conditions

		Pre-load			Load at estimated time of injury			Fztotal Peak (N)	Time to Peak (msec)	Equiv Fz Tibia (N)	Injury	OTA classification
	Impactor position	Impactor head	Proximal load cell	Achilles tension	Impactor head	Proximal load cell	Achilles tension					
08R	A	387	2576	1693	9894	7907	87	10713	23	11355	Calcaneal fracture	73-C1.1
10L	A	120	1854	1529	_	_	_	10197	33.7	10142	Nil	0
12R	A	90	2206	1614	7121	6766	612	8391	29.8	8469	Calcaneal fracture	73-B3.3
06L	B	78	1135	na	8485	8317	na	8452	29.2	8557	Pilon fracture	43-B1.2
08L	B	2	531	334	7151	7052	596	7154	28	7293	Calcaneal fracture	73-C1.1
09L	B	1056	2629	1111	5535	5447	2083	5598	23.3	5916	Pilon fracture	43-C2.2
09R	B	2406	749	na	5601	4333	na	5623	26	5810	Calcaneal fracture	73-C1.3
10R	B	1789	3546	1900	5416	8839	2622	6510	23.6	6861	Talar body fracture	72-B2.1
11R	B	1061	2376	1457	3626	4343	990	6208	23.5	6549	Rupture of deltoid ligament	0
13L	B	148	1122	593	5492	5297	493	5396	27.3	5525	Calcaneal fracture	73-C3
13R	B	800	2754	1629	6078	5152	1769	6375	22.3	6803	Calcaneal fracture	73-C2.1
20R	B	922	4685	2373	_	_	_	7162	26.7	7363	Nil	0
02L	C	605	1371	697	5510	4101	1347	5543	23.3	5858	Calcaneal fracture	73-C1.1
02R	C	195	2650	1759	4883	6094	3	4955	22.8	5262	Calcaneal fracture	73-C3
03L	C	1488	3046	1647	5640	7231	3114	5848	22.4	6235	Talar neck fracture	72-A1.1
03R	C	916	2904	1662	5312	6899	2799	5345	22.6	5687	Talar body fracture	72-B1.1
11L	C	251	2554	1695	_	_	_	_	_	_	Nil	0
12L	C	853	3582	2393	6241	6842	2657	6356	21.9	6811	Calcaneal fracture	73-C3
14L	C	316	2405	1644	1230	3069	1854	4180	22.8	4439	Rupture of deltoid ligament	0
14R	C	-139	718	na	2108	3525	na	2723	30.2	2743	Medial malleolus fracture	MM
20L	C	112	4078	2489	4955	7249	3029				Medial malleolus fracture	0
21L	C	1153	3565	2075	5935	10374	3419				Rupture of deltoid ligament	0
21R	C	2013	3779	1895	7331	8263	2561				Pilon fracture	43-B3.1