# DYNAMIC RESPONSE AND INJURY MECHANISM IN THE HUMAN FOOT AND ANKLE AND AN ANALYSIS OF DUMMY BIOFIDELITY

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## ABSTRACT

Lower leg injuries are the most frequent severe injury (AIS/3) to car occupants resulting from frontal impacts. Consequently it is important to be able to detect and quantify the risk of these injuries from car impact tests. The need for a more biofidelic instrumented lower leg and the development of well-founded injury criteria has been clearly identified. At present there are insufficient biomechanical data available to aid the design and construction of such a dummy leg or to define injury risk criteria.

Three phases of a research programme are reported. In phase one, a driving simulator trial was carried out using 24 subjects to investigate the positioning and bracing of the driver's lower leg prior to an emergency-braking event. In the second phase of the work, low energy impact tests have been carried out on post mortem human surrogate (PMHS) legs and volunteers. Comparative tests were performed on existing Hybrid III and GM/FTSS dummy feet/ankles. A new method to generate an Achilles force is reported which has been used to enable studies to be carried out into the effect of bracing by plantar flexion, such as emergency braking. Dynamic tests were based on the EEVC foot certification test procedure. The programme included tests to produce inversion and eversion. Additional toe impact tests were performed to assess lower leg behaviour with pre-impact bracing, allowing for future comparisons with the ALEx leg.

Difficulties inherent in the interpretation of PMHS testing are reviewed and comparisons between tests on PMHS subjects and human volunteers are presented.

The final part of the paper reports the findings of an in-depth retrospective accident analysis based on data in the UK Co-operative Crash Injury Study database. Accidents in which the front seat occupant had sustained a lower leg injury of AIS/2 in a frontal collision were examined along with the associated hospital notes and X-rays. The injuries are ranked in terms of frequency, severity and impairment in order to determine which lower leg injuries that are a priority for prevention.

## INTRODUCTION

Historically, regulatory tests have been aimed at reducing the incidence of serious injuries to the body regions where there is a high risk of a fatal outcome. Thus protection criteria for the head and chest have been foremost in the early regulations. Significant improvements have been seen in the protection afforded against fatal injuries to these body regions and more attention is now being paid to the serious and debilitating injuries that contribute markedly to the societal costs of road accidents. For this reason, a requirement for injury protection to the leg and ankle was included in the EU Directive on Frontal Impact [1]. The ankle joint of the Hybrid III dummy was modified to include a soft bump stop at the ends of the articulation range to allow the measurements of the injury criteria to be made more reliably. This was regarded as a first step in the improvement of the protection for leg, foot and ankle injuries.

There have been a number of studies of lower extremity injuries. The long-term consequences of lower extremity fractures have been investigated by Luchter [2]. He analysed the outcome of lower extremity injuries resulting from police reported tow-away motor vehicle crashes in the US in 1993. Lower extremity injuries accounted for 17 percent of the total cost of these collisions but accounted for 41 percent of the life years lost as a result of injury. Miller, Martin and Crandall [3] stated that in police reported highway collisions in the US in 1993, lower limb injuries cost an estimated $21.5 billion in passenger vehicle occupant injury costs.

In France, Salmi et al [4], in a population based cohort study to look at long term post traumatic disablement, concluded that lower extremity injuries are one of the most frequent sources of injury-related hospitalisation and that they result in substantial long term morbidity.

From Australia, Fildes et al [5] have reported that lower limb injury occurs in about 40% of passenger car crashes that result in injury. Three quarters of all fractures occurred below 64 km/h. No simple relationship was apparent between intrusion and fracture of the lower limb and the authors state that it is possible that foot and ankle injuries can occur simply from the excessive loading that comes with bracing.

The increasing need to focus on non-life threatening car crash injuries has resulted in renewed interest in car crash dummy design and a number of research projects have focused on comparative tests on Post Mortem Human Surrogates (PMHS). A further development has been the realisation that PMHS specimens in themselves are not completely biofidelic and that they lack the properties of a physiologically active human being. In particular they lack any active muscle tension that would resist movement in a live human being [6-10].

In 1996, Crandall et al [11] reported a series of sled tests using PMHS specimens and the Hybrid III dummy. They concluded that the Hybrid III records greater axial tibial loads and ankle dorsiflexion than the PMHS in comparable conditions. In 1995 Schueler et al [12] reported a series of 60 dummy tests and 24 PMHS specimen tests using a co-axial impactor. They concluded that improved dummy lower legs were required for more reliable injury prediction. They recommended that improvements were needed in the following areas:

1. The kinematics of the foot and the knee joint.
2. The mass distribution and relationship of bony and muscular structures.
3. The location of measuring devices as close to the injury point as possible.

### The Hybrid III Leg, Foot and Ankle

The standard Hybrid III dummy is based on a fiftieth percentile male. Within the dummy the pivot point of the knee is designed to be 49 cm from the ground and with the tibial plateau 44 cm from the sole of the foot with the ankle in the neutral position. The foot measures 26 cm in length and 10 cm in breadth. The lower leg assembly weighs 5.7 Kg. The foot and ankle assembly of the Hybrid III comprises a ball and socket type joint that is intended to simulate the movements of both the ankle and subtalar joints (Figure 1). Initial contact of the ankle shaft and ankle bumper occurs at 36° of dorsiflexion movement and is fully compressed at 45° of movement in this direction and 35° of plantar flexion. Eversion/inversion initial contact occurs at 20° of movement with a maximum 23° of rotation being achieved in each direction. [13]

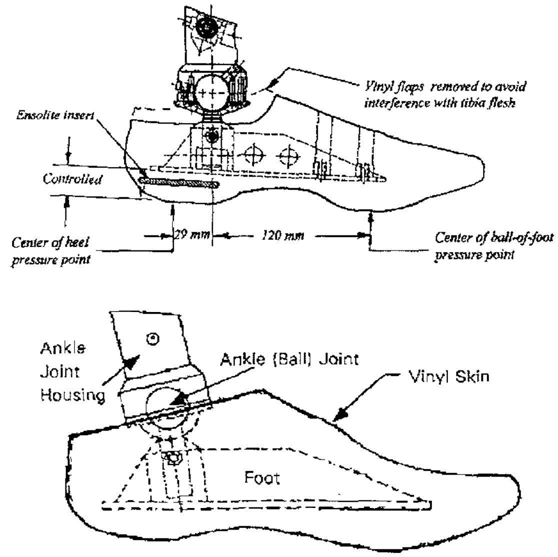


**Figure 1 - The Hybrid III Foot Assembly**

### The GM/FTSS Advanced Foot and Ankle

The GM/FTSS foot has been designed and manufactured to attach to the distal end of the Hybrid III lower leg assembly. In the GM/FTSS foot the ankle joint is represented by a hinge joint, and the subtalar joint by the inclusion of a further ball and socket joint below the ankle hinge (Figure 2). A compliant and continuous joint stop has been included that allows 60° of plantar flexion and 45° of dorsiflexion. In addition 30° of eversion and 35° of inversion are possible. A further joint is positioned in the midfoot position to represent movement within the foot arch, mid-tarsal and metatarsal joints. As with the Hybrid III foot, no instrumentation is included in the foot and ankle assembly.

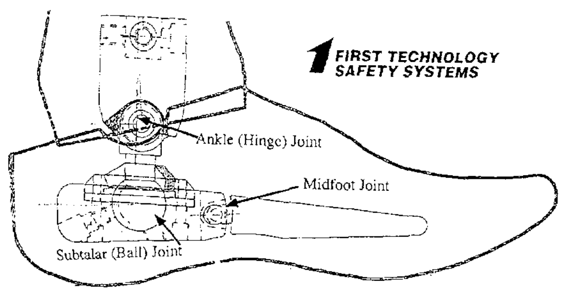


**Figure 2 - The GM/FTSS Foot Assembly**

### Centres of Rotation

The centres of rotation of the hind foot joints for PMHS and dummy joints have been studied by several institutes including Crandall et al [9]. In their PMHS tests they report the centre of rotation for dorsiflexion at a distance of 76 ± 8 mm from the sole of the foot and 61 ± 6 mm from the posterior face of the heel. The centre of rotation in inversion and eversion was 71 ± 12 mm from the sole of the foot and 3 ±12 mm from the long axis of the leg. The Hybrid III foot and ankle moves in the planes of dorsiflexion and plantar flexion and inversion and eversion at the centre of its ball joint 82.6 mm from the sole of the foot. In the GM/FTSS foot the centre for dorsiflexion & plantar flexion has been calculated at 58 mm from the sole of the foot. The centre for inversion and eversion coincides with the position of the simulated sub-talar joint at 39 mm from the sole of the foot (Figure 3).

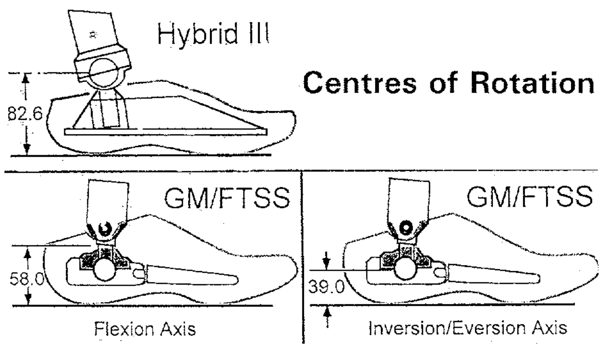


**Figure 3 – Centres of Rotation for Dummies**

The normal ranges of motion for male subjects aged 30-40 years were reported in 1982 by Roaas and Anderson [14]. In this study which was conducted using simple goniometers they reported the following results for the ankle joint. Dorsiflexion 5 to 40°, mean 15.3°; Plantar flexion 10 to 55°, mean 39.6°; Eversion 15 to 50°, mean 27.7°; Inversion 15 to 50 °, mean 27.7°. In PMHS specimens, movement to a greater degree, which would be beyond the pain threshold in a normal human, is possible without apparent damage.

### Mass and Moments of Inertia

A number of laboratories have studied mass and moments of inertia differences for the lower leg [9], and measurement for comparison have been made in the tests reported in this paper. They are compared with other published data in the results section of this paper.

### Summary of Tests to Date

In summarising their collected work, comparing the dummies to PMHS specimens, Crandall et al [9] conclude that the GM/FTSS and other new generation feet (ALEx II and Hybrid III with bump stop) show improvements in many parameters that give them a biofidelic advantage over the original Hybrid III. They highlighted important differences in geometry and consequently stability and response to load. In their work on the dynamic response to load, they highlight the importance of active and passive musculature in determining the axial and rotational responses of the lower leg. The gastrocnemius muscle group was shown to influence both the moment angle joint characteristics as well as the loading in the tibia and fibula during dorsiflexion.

In his thesis Portier [15] summarises both his own work and the collected data of other centres and makes some important conclusions with respect to the physical properties of the lower leg. His results showed substantial differences in centres of pressure, centres of rotation, inertial properties and axial stiffness.

In further work reported at Stapp 1997, Portier et al. [15] reported on the results of further sled tests. They supported the consensus view that the GM/FTSS foot was more biofidelic than the Hybrid III, but highlighted deficiencies in the moment rotation relationship. They firmly conclude that passive and active muscle tension should be included in future dummies to obtain improved ratios between the shear and compressive forces in the ankle joint.

This paper reports work carried out within the LLIMP project (Lower Limb Injuries, Methods of Prevention) which has been performed through the collaboration between the Transport Research Laboratory, the University of Nottingham and the Vehicle Safety Research Centre of the Institute of Consumer Ergonomics. This group has bought together doctors and engineers with a specialist interest in car crash injury prevention.

The first section of the paper reports the details of a series of driving simulation tests designed to assess the kinematics of the lower leg during emergency manoeuvres when driving and to evaluate the forces of braking generated at the same time. This is followed by a report of the biofidelity study comprising a series of low energy impact tests with PMHS specimens (through-knee amputations, above knee amputations and specimens with an applied plantar flexing force). These tests were then repeated with volunteers and with two different dummy feet to assess biofidelity.

Lastly the results of a recent, in-depth, accident analysis are given. This analysis focuses not only on the extent of the problem of lower leg injuries in the UK, but more importantly addresses fracture mechanisms and injury impairment scoring through an analysis conducted jointly by the Vehicle Safety Research Centre and the Department of Orthopaedics at the University of Nottingham.

This paper concludes with an overview of further work in this field currently being performed by the LLIMP collaborative group in the United Kingdom.

## DRIVING SIMULATOR TRIALS

This study was designed to evaluate the kinematics of the driver's lower leg during braking, in a simulated emergency, whilst at the same time measuring the applied brake pedal force. The aim of this part of the study was to provide information on the position and extent of muscular bracing in the lower leg during emergency braking. These are thought to influence the potential risk and mechanism of lower leg injury in car crashes.

A driving simulator with realistic three-dimensional graphics (Silicon Graphics) has been developed at TRL (Figure 4). The simulator is based around a full-scale car (Rover 400, right hand drive configuration). The simulator was fitted with two hidden infrared cameras to record foot movements. A pressure transducer in the vehicle brake line was used to record applied brake pedal pressure. One of the cameras provided a view of the drivers' feet from underneath the drivers' seat and the second a lateral view from under the dashboard. Use of a driving simulator allows safe exposure to realistic driving scenarios, including emergencies.

To enhance the kinematic analysis, which included video recording, subjects were also fitted with a goniometer (Penny and Giles) to their right ankle to record both ankle joint and sub-talar joint movement.

### Subjects

This study was conducted in two phases, each using twelve subjects, six male and six female. In both phases, all subjects had prior experience of driving the simulator. Each subject was led to believe that the driving task was speed control. Before each test each subject's sex, age, height and build were recorded. At the conclusion of each test, the subjects were questioned to ascertain how realistic they considered the simulation to be.

### Test Configurations

In phase 1, two 'static' emergency events, a "military tank" obstructing the road and a fallen tree across the road, were introduced into a thirty-minute driving simulation. The location of the tank and the fallen tree were on a bend in the road, and over the brow of a hill respectively. In the second phase of the trial, the simulation was enhanced to include more traffic on the road during general driving, and the first static event was replaced with a dynamic emergency event. This took the form of a vehicle on the other side of the road overtaking into the path of the driver in the driving simulator. The event was designed such that the likelihood of the driver swerving to avoid the oncoming vehicle rather than braking was minimised. After the second emergency event in both phases of the trial, the simulation was terminated and the subject informed of the true purpose of the trial.

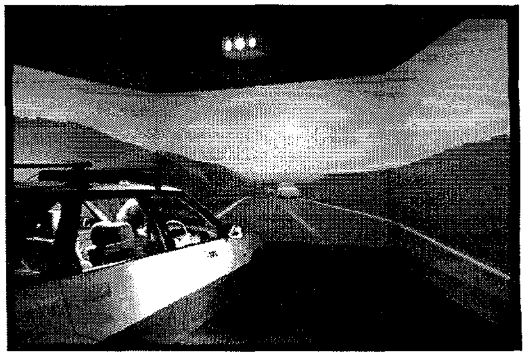

**Figure 4 – The TRL driving simulator**

Table 1 shows a summary of the physical data collected for each subject in the trial, and an indication as to the realism of the simulation as experienced by the subject. The recorded maximum braking force for both emergency events and the degree of plantar flexion of the right foot during the first emergency event is also presented in Table 1.

In both phases of the trial, a visual inspection of the video recordings showed that most of the subjects drove with their heel longitudinally in line with the brake pedal or longitudinally in line with a point between the brake pedal and accelerator. The foot was rotated clockwise and slightly plantar flexed to allow depression of the accelerator. Only four subjects drove with their heel in-line with the accelerator while driving. When responding to the emergency event, two different movements were used to depress the brake pedal. The first was lifting the foot off the floor and stamping on the brake pedal, the second involved rotating the whole foot onto the brake pedal and plantar flexing the foot with the heel still in contact with the floor. In addition it was observed that, although the heel was in contact with the floor as braking was initiated, in just less than 50 percent of all cases, heel to floor contact was lost as the brake pedal force was increased.

**Table 1.**
**Simulator Trial Data for Phases 1 and 2**

| Subject No. | Sex | Age | Height (m) | Realistic Driving? | Max. Braking Force (N) (Event 1) | Max. Braking Force (N) (Event 2) | Plantar flexion (°) | Heel in contact during braking? Ev1 | Ev2 |
|---|---|---|---|---|---|---|---|---|---|
| **Phase 1** | | | | | | | | | |
| 1 | F | 20-30 | 1.65 | No | 820 | 821 | 43 | Yes | Yes |
| 2* | M | 20-30 | 1.84 | Yes | - | 143 | - | No | No |
| 3 | F | 40-50 | 1.73 | Yes | 716 | 587 | 6 | Yes | Yes |
| 4** | F | 40-50 | 1.69 | No | 807 | - | 1 | No | No |
| 5 | F | 40-50 | 1.58 | - | 817 | 927 | 17 | No | No |
| 6 | M | 40-50 | 1.94 | Yes | 1474 | 1003 | 14 | Yes | Yes |
| 7* | M | 40-50 | 1.73 | Yes | 775 | 884 | - | - | Yes |
| 8 | M | 20-30 | 1.76 | Yes | 735 | 466 | 26 | No | No |
| 9 | F | 40-50 | 1.73 | Yes | 1474 | 862 | 16 | No | No |
| 10 | F | 20-30 | 1.64 | Yes | 620 | 928 | 15 | Yes | Yes |
| 11 | M | 40-50 | 1.7 | Yes | 331[+] | 252 | 23[+] | Yes[+] | Yes |
| 12 | M | 20-30 | 1.72 | Yes | 636 | 431 | 21 | No | No |
| **Phase 2** | | | | | | | | | |
| 13 | F | 40-50 | 1.69 | No | 124 | 593 | -5 | No | No |
| 14 | F | 40-50 | 1.68 | Yes | 401 | 554 | 9 | No | No |
| 15 | F | 20-30 | 1.72 | Yes | 249 | 108 | 14 | No | No |
| 16 | M | 40-50 | 1.83 | Yes | 119[+] | 84[+] | -12[+] | Yes[+] | Yes[+] |
| 17* | M | 20-30 | 1.82 | No | - | 108 | - | - | Yes |
| 18 | F | 20-30 | 1.71 | Yes | 214 | 88 | 6 | Yes | Yes |
| 19 | F | 20-30 | 1.67 | Yes | 404 | 200[+] | 44 | No | No[+] |
| 20 | M | 20-30 | 1.84 | No | 1207 | 339 | 23 | Yes | No |
| 21 | M | 20-30 | 1.84 | No | 181 | 552 | 14 | Yes | Yes |
| 22** | M | 40-50 | 1.85 | No | 550 | - | 12 | No | - |
| 23 | F | 40-50 | 1.7 | Yes | 482 | 1171 | -2 | Yes | Yes |
| 24 | M | 20-30 | 1.76 | Yes | 1150 | 561 | 18 | Yes | Yes |

* - No data acquired for event 1 due to instrumentation problems
** - No data acquired for event 2 due to instrumentation problems
\+ - Data not included in analysis because speed prior to braking was less than 35mph

**Combined Phase 1 & Phase 2 Results**

For the purposes of this study the results of both phases of the trial have been combined. The mean peak brake force was determined to be 630N (standard deviation, $\sigma$ 366N), with the ankle joint at a mean angle of 15° ($\sigma$ 13°) of plantar flexion at the point of the applied peak brake force.

All the subjects in the trial used the ball of their foot to apply force to the brake pedal. The ball of the foot is represented on radiographs as the first metatarsal head. Analysis of the radiographs obtained from the PMHS specimens used for biomechanical testing in the LLIMP project indicated a 12:5 ratio in the distance between the first metatarsal head, the ankle joint centre and the insertion of the Achilles tendon. If the ankle joint is considered to act as a single axis hinge for the purpose of this estimate and it is assumed that all of the plantar flexing force is generated through the Achilles tendon, it is possible to calculate the mean plantar flexing force in the lower leg plantar flexing muscle group. Based on this assumption it was calculated that approximately 1.5kN ($\sigma$ 900N) acting at the Achilles attachment would be required to recreate the mean maximum braking force seen in the two phases of the trial.

If males and females are considered separately for the two phases of the trial, the mean peak brake applied force was 636N ($\sigma$ 389N) for males and 626N ($\sigma$356N) for females. For the purposes of this study, statistical significance is taken at the fifth percent level, thus this result is not statistically significant ($p=0.932$). There was also no significant difference in the mean angle of plantar flexion between males (18°) and females (14°) ($p=0.464$).

In comparing the combined results of phase 1 and 2 for event 1 and event 2, the data show that for event 1 the mean peak applied brake force was 692N ($\sigma$ 397N) compared with 569N ($\sigma$ 331N) for event 2, which was not statistically significant ($p=0.295$). The difference in the mean speed before event 1 (47mph $\sigma$ 6.6mph) and event 2 (47.51mph $\sigma$ 6.01mph) was also found to be not significant ($p=0.846$).

There was no significant difference ($p=0.394$) between the mean height of the subjects whose heel was in contact with the floor during braking (1.76m $\sigma$ $9.12\times10^{-2}$ m) and those subjects who lifted their heel during braking (1.72m $\sigma$ $7.65\times10^{-2}$ m).

In analysing the post-trial questionnaires, there was no significant difference ($p=0.513$) in the mean applied brake pedal force for subjects who thought that the driving simulation was realistic (643N $\sigma$ 384N) and those who did not think that it was realistic (555 $\sigma$ 347N).

A linear correlation calculation was carried out in order to determine whether there was a relationship between the subject's height and peak force applied to the brake pedal or the angle of plantar flexion. In both tests, no significant correlation was found with $r=-0.003$, $p=0.99$ for brake force and $r=-0.089$, $p=0.752$ for angle of plantar flexion. No significant correlation was found between the applied brake force and the angle of plantar flexion ($r=0.231$, $p=0.342$). An examination of the relationship between the speed prior to braking and the maximum applied brake pedal force showed a weak correlation ($r=0.289$, $p=0.071$).

### Discussion of Simulator Trials

The work that has been reported here indicates that the driving simulator developed at TRL can be effectively used to evoke an emergency braking response. Our study indicated that the mean peak force applied to the brake pedal with the ball of the foot was 630N and that the right foot is plantar flexed by approximately 15° when the maximum braking force is achieved. It is not known at what point in time during an accident the ankle or foot is injured. If an injury is pedal induced and occurs during braking, as suggested by a recent accident analysis [16], then these results indicate that it would be preferable to study ankle injury with the foot in an initial plantar flexed position. The observation that for 50 percent of the drivers, the heel left the floor during the braking actions suggests that in accident investigations, firm evidence of foot position should be sought when ascertaining the mechanism of a lower leg injury. However, it is recognised that this would be extremely difficult to achieve in practice. In a similar way, biomechanical evaluations of fracture mechanisms should encompass both initial heel to floor contact and heel to floor separation. This study indicates that in future legislative crash testing the initial position of the lower leg should be better specified, particularly if the risk of lower leg injury is to be assessed.

The mean forefoot force of 630N reported in this study is significantly greater than the force used by other researchers [7] in their work to ascertain the effect of bracing on the lower leg. The simulator trial results indicate that in simulated realistic emergency situations, drivers can apply a force to the brake pedal far in excess of that required to obtain maximum braking power for their vehicles. The recent accident analysis of Taylor et al. [16] identified cases where drivers had deformed the brake pedal due to severe braking.

## BIOFIDELITY TESTS

### Introduction

With an increasing and ever more complex range of dummy components available for testing there is a need to be able to quantify how biofidelic each component is and how realistic the data obtained in impact testing are compared to that for a live human subject. Impact testing to injury levels cannot be undertaken on live human subjects, so conventionally PMHS specimens are used as a compromise to provide reference data. At the same time, repeated crash testing in real cars is expensive and resources are limited. There is therefore a strong desire to ensure that any data obtained in such tests are as realistic as possible.

The aim of any biofidelity research must allow for realistic modifications to dummies that will provide useful data as opposed to an attempt to recreate a surrogate so complex and diverse in its design that it becomes practical only as research tool and inappropriate for legislative testing.

Extensive work in the USA [17] has proved that, if the results of cadaveric biomechanical testing are to be repeatable and biofidelic, they can only be performed on fresh frozen specimens. Fresh frozen is a term commonly used to imply that the specimen is frozen fresh soon after retrieval and thawed at a later date prior to testing. All major international laboratories performing biomechanical studies now use fresh frozen cadavers. For the purposes of this study, fresh frozen cadaveric material has been used throughout.

Ethical approval for this study was granted by the relevant organisations in the UK.

PMHS specimen testing usually involves small numbers, most often taken from a relatively elderly population. In order that some comparison of results from different laboratories may be made, the collection of relevant anthropometric data and other physical properties is essential. In this study, a new method for normalisation of results to the fiftieth percentile using finite element modelling is described.

The tests were a series of simple sub-injury tests based around the dummy foot certification procedure [1]. Tests to examine the biofidelity of dummies in dynamic inversion and eversion impacts were introduced, in addition to the standard heel and toe impacts. While it is recognised that the currently available dummy lower legs are not all designed to incorporate a biofidelic inversion/eversion response, it is well known that these movements are involved in the mechanisms of lower leg injury [18]. It is assumed that a dummy ankle that behaves correctly in inversion/eversion would transmit loads to the leg via a more accurate load pathway to the tibia.

In the light of so many recent papers [15] highlighting the importance of the difference between the response to impact seen in a live individual and that in a flaccid cadaver, this study also reports on results of an impact analysis with volunteers. The tests in this part of the study were also conducted according to the EEVC new dummy foot certification procedure [1]. The energy delivered in each case was known to be of a sub-injury level from the results of the tests with the PMHS specimens. The results of further tests using pre-tensed or braced PMHS specimens are also reported in this paper.

No PMHS could ever be instrumented in such a way that natural muscle tension could be reproduced in all of the muscle groups acting across the knee, ankle, subtalar and other foot joints. It is also accepted that it is impossible to get a volunteer to relax and not actively contract his/her lower leg musculature when confronted with a simulated impact.

The biofidelity tests aim to provide comparative data that can be used to assess dummy leg performance compared to actual human lower leg behaviour. Full instrumentation of live human lower legs is impossible and most evaluations are made using data obtained from PMHS specimens. It is however, possible to study the kinematic response and impactor response of a live subject to a sub-injury impact.

### Methods for PMHS Specimen Testing

**Storage of PMHS Specimens** - All specimens were initially frozen at -80°C and subsequently stored at -26°C. Each specimen was allowed to thaw for over 24 hours prior to instrumentation and subsequent testing.

**Pre-Test Radiological Imaging** - Standard lower leg radiographs were taken prior to the testing: An orthopaedic surgeon reviewed all radiographs for pre-existing anomalies, prior to testing. All specimens were analysed for Bone Mineral Density (BMD) using a DXA scanner.

**Specimens** - The specimens were all retrieved as above knee amputations, however for most tests in this study they were disarticulated through the tibio-femoral (knee) joint, and then cut cleanly at a consistently identifiable landmark just below the knee. A further additional series of tests was carried out on an above knee specimen to allow a direct comparison to be made between the two specimen types.

The PMHS specimens were all worked to allow a normal range of articulation.

**Basic Physical Measurements** - The following basic physical measurements were taken from the PMHS:

1. Mass of lower leg and mass of foot.
2. Dimensions:
   a) Lower leg length (tibial plateau to sole of foot with ankle at 90°)
   b) Foot length (end of first toe - dorsal edge of heel)
3. Volume of lower leg and foot.
4. Centre of Gravity.
5. Heel flesh depth at the ball and heel of the foot.
6. Moment of inertia about the y-axis for the leg and the y and z-axis for the foot. These were measured using a two-wire torsion tray based rig.

**Potting Procedure** - For the below knee specimens, the upper 5 cm of soft tissue were removed from the PMHS specimen by sharp dissection. To preserve the proximal tibia/fibula joint, the upper end of the fibula was secured to the tibia by means of a length of 5mm studding passed through a 6mm-drilled hole. The specimens were potted in standard polyester-based filler.

**Tibial Load Cell** - A Denton™ five axis load cell, which was a modification of the load cell used in the Hybrid III dummy, measured forces in three directions ($F_x$, $F_y$, $F_z$) as well as two bending moments ($M_x$ and $M_y$). For this data to be comparable across all PMHS specimen tests the load cell had to be mounted not only in the correct orientation, but also maintaining the exact alignment of the tibial shaft in all planes. A special installation technique was developed which allowed exact alignment of the load cell whilst maintaining the integrity of the fibula without employing an external fixator (Figure 5). The full details of this technique are given in Appendix 1.

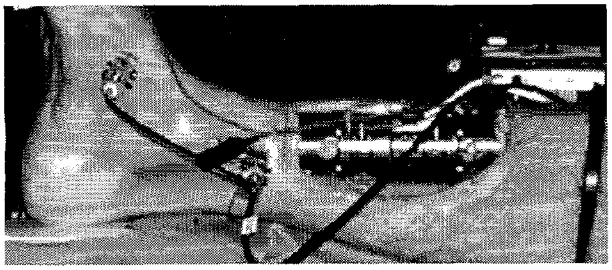

**Figure 5 - Instrumented PMHS Specimen**

__Acoustic Transducer__ – An acoustic transducer was attached to a metatarsal with the object of detecting failures during the test (see end of Appendix 1).

__Shoe Representation__ - Each PMHS was fitted with a representative sole of a shoe, which was secured by Ethibond sutures. The sole was made of industrial black Velbex Flexible 6290 plasticised PVC sheet, 3mm thick.

__Physical Properties of the Instrumented Leg__ – The physical properties of the whole-instrumented PMHS specimen (PMHS, clevis clamp, tibial load cell, acoustic transducers and shoe representation) were measured according to the list above, excluding the volume measurements.

__Pendulum Rig__ - A simple rigid pendulum impactor rig was used to perform the dynamic tests for this study. The design of this rig was based on the EEVC new dummy foot certification procedure [1] (Figure 6). It consisted of a horizontal cylinder of radius 25±1mm and a lightweight support arm. The cylinder had a mass of 1.25±0.02kg including instrumentation and the arm has a mass of 285±5g. The pendulum was supported by a scaffold structure, which had been adapted to allow the adjustments needed to position PMHS specimens of differing lengths on the rig.

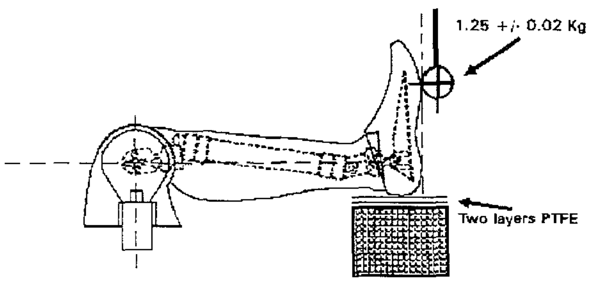


**Figure 6 - Pendulum Rig Configuration**

__Pendulum Accelerometer__ - One single axis piezo-resistive accelerometer was mounted on the pendulum.

__High Speed Film__ - Two 16mm high-speed cine cameras (lateral and overhead perspectives) operating at 400 frames per second, were used to provide a visual record of the tests.

**Testing Specification**

All testing for this study was designed around a derivation of the EEVC 'Tibia and Foot Certification Tests for use with the Hybrid III dummy in the EEVC Offset Deformable Front Impact Test Procedure' [1]. For the purposes of foot tests the procedure requires that (in the case of a dummy) the line joining the knee clevis joint and the ankle attachment is horizontal ± 3°, with the heel resting on two layers of low friction material. The underside (sole) of the foot should be vertical ± 3°

__PMHS Specimen Orientation on the Rig__ - The proximal end of the specimen was attached to the back plate of the pendulum rig, using a clevis load cell. For heel and toe impacts the foot was orientated such that the second metatarsal pointed vertically upwards, with the ankle at 90° (neutral position). For inversion and eversion impacts the specimen was orientated such that the foot was resting on its medial or lateral side respectively, with the ankle remaining in the neutral position. The back plate was rigidly secured to prevent movement during impact. The specimen was orientated on the rig such that the line between the ankle joint (mid-malleolar point) and the knee joint was horizontal. This was checked by the alignment of the tibial load cell.

__Lower Foot (Heel) Impact Test__ - The height at which the cylinder impacted the foot (as measured from the base of the heel) was calculated according to the relevant dummy test defined in the EEVC foot certification procedure. The EEVC foot certification procedure defines a height of 62mm for a total foot length of 265mm. Laterally, the centre of the cylinder was aligned with the axis of the second metatarsal. The heel impact tests were performed at 2m/sec and 4m/sec. At these impact velocities, the energy delivered is normally below the injury threshold.

__Upper Foot (Toe) Impact Test__ - The EEVC foot certification procedure defines a height of 185mm for a total foot length of 265mm.

The toe impact tests were performed at 2m/sec, 4m/sec and 6m/sec. At these impact velocities, the energy delivered is normally below the injury threshold.

__Inversion Impacts__ - The specimen was orientated and mounted lying on its lateral side. As above, the specimen was mounted such that the longitudinal axis was horizontal and the foot was in the neutral position. The cylinder was aligned such that it impacted the medial side of the foot, causing it to invert during impact. The height at which the cylinder impacts the foot was defined by impacting the first metatarsal head.

The eversion impact tests were performed at 2m/sec and 4m/sec. At these impact velocities, the energy delivered is normally below the injury threshold.

**Eversion Impacts** - The specimen was orientated and mounted lying on its medial side. As above, the specimen was mounted such that the longitudinal axis was horizontal and the foot was in the neutral position. The cylinder was aligned such that it impacted the lateral side of the foot, causing it to evert during impact. The height at which the cylinder impacted the foot was set so that it impacted the fifth metatarsal head

The inversion impact tests were performed at 2m/sec and 4m/sec. At these impact velocities, the energy delivered is normally below the injury threshold.

**Test Order** - The test order was changed for each specimen. This was to ensure that there was no bias in the results, due to possible changes in the articulation of the ankle, sub-talar and mid-foot joints with repeated testing.

**Test Termination Conditions** - All eight tests were performed unless a physical examination by an orthopaedic surgeon or the acoustic data suggested that an injury might have occurred during testing.

### Post Test Procedures

After testing, the standard lower leg radiographs referred to above were re-taken. The radiographs were examined in detail by both an orthopaedic surgeon and a consultant radiologist. In addition to a clinical examination, each specimen was dissected at the conclusion of testing to exclude soft tissue and bony injuries. The physical properties of the foot were measured, after dis-articulation, for the normalisation process.

### Methods - Dummy Comparative Tests

The Hybrid III lower leg fitted with the 45° articulating ankle and soft bump stop and the Hybrid III lower leg fitted with the GM/FTSS ankle and foot were used for these tests. The testing protocol was designed to mimic the tests carried out on the PMHS specimens exactly and to adhere to the guidelines of the EEVC dummy foot certification procedure [1] (Figure 7). For each dummy type and for each test configuration three identical tests were carried out. A recovery period of 30 minutes was allowed between each test as specified in the EEVC procedure.

**Shoe Representation** - In a similar fashion to the PMHS testing each dummy foot was fitted with a representative sole of a shoe.

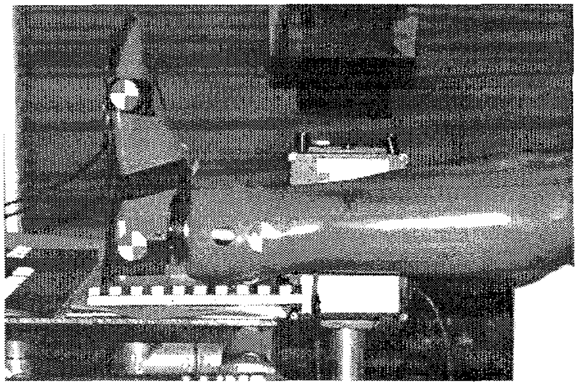

**Figure 7 - Dummy Test Configuration**

### Methods - Volunteer Tests

**Pendulum Rig** - The same pendulum rig as used for the PMHS tests was used for six different volunteers. This rig was adapted to allow the volunteer to be comfortably reclined with their right lower leg in the position defined in the EEVC dummy foot certification procedure.

**Instrumentation** - For the tests with volunteers the instrumentation used was the pendulum accelerometer (as for PMHS and dummy tests) and a goniometer fitted to the volunteer's leg to measure the degree of dorsiflexion/ plantar flexion of the foot/ankle in response to the pendulum impact.

As with the PMHS and dummy tests, a shoe representation was fitted on to the sole of the foot and high speed cine cameras were used to film the impact.

**Test Specification** - The tests with volunteers were conducted both with the volunteer unaware of the impending impact to their foot and conversely with the volunteer fully aware of the impending impact. In the case of the volunteer being fully aware of the test, he/she was told to brace and resist the impacting pendulum. The volunteers were only exposed to the toe impacts and not the heel, inversion or eversion tests.

The volunteers were made unaware of the impending impact by wearing headphones and by being blind-folded for the duration of the testing session.

**Volunteer Leg Orientation on Rig** - The volunteers were positioned on a surgical couch with their foot in a similar position to that of the dummy leg in a dummy test. Two sheets of low friction (PTFE sheet) were placed under the heel. The volunteer's leg was orientated such that the foot pointed vertically upwards for the impacts, and such that the line between the ankle and the knee joint was horizontal.

The height at which the pendulum impacted the foot was calculated according to the relevant dummy test defined in the EEVC foot certification procedure. The toe impact tests were performed at 2m/sec, 4m/sec and 6m/sec, as for the PMHS tests (Figure 8).

**Test Termination Conditions** - All the required tests were performed unless the volunteer requested that the testing should stop.

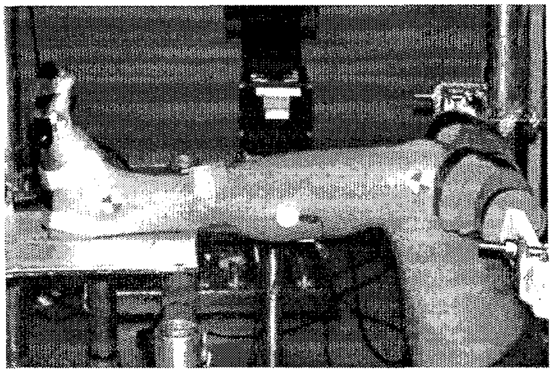

**Figure 8 - Volunteer Test Configuration**

**Methods - Braced PMHS Specimen Testing**

**The Loading Cylinder** - There have been attempts to load the Achilles tendon in human surrogate testing but the majority of these have failed to date due to an inability to adequately grip the tendon. In 1997 the authors have reported a series of tests [6] where a small hydraulic cylinder and clevis clamp had been successfully used. This method has been used again to apply a plantar flexing/bracing load in these tests.

This technique involved the replacement of the muscle group with a custom built tensioning device based on a hydraulic cylinder (Figure 9).

This cylinder is capable of developing loads of up to 4kN and is hydraulically driven through an air/oil intensifier, which is supplied by an air compressor. Control is achieved via fine pressure control valves on inlet and exhaust. One end of the cylinder is attached to the back plate via a turnbuckle (which allows for gross tension adjustments prior to testing).

The attachment to the back plate is adjustable so that the cylinder will pull in line with the long axis of the tibia thus preventing the introduction of artificial bending moments. The piston end is attached to the Achilles tendon and was clamped in a clevis bracket, on the end of the piston rod (Figure 10). The length of stroke of the cylinder was 70mm.

A load cell was located in line with the cylinder to measure the load applied and a further load cell was positioned in line with the stirrup to measure the force applied at the forefoot.

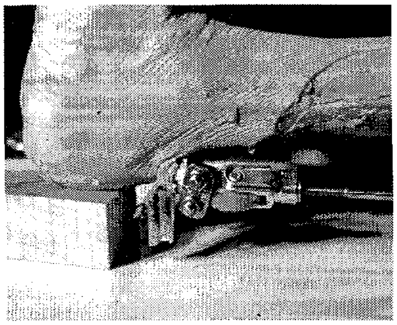

**Figure 10 - Achilles Tendon Clamp**

**Instrumentation of Specimens** - The specimens were prepared according to the method described in reference [6]

Active tension applied to the Achilles tendon resulted in plantar flexion of the unsupported foot. In order to restrict this motion the foot had to be externally supported. This was achieved by the use of a stirrup around the forefoot. The Achilles loading cylinder is schematically shown below the tibia and fibula (Figure 11). The general position of the tibia load cell is also illustrated. The Achilles loads chosen for this work were based on previous published work [9] and the results of the simulator trials described in this paper.

The chosen loading conditions were:

1. The loading cylinder in place, but with no tension applied to the Achilles tendon.

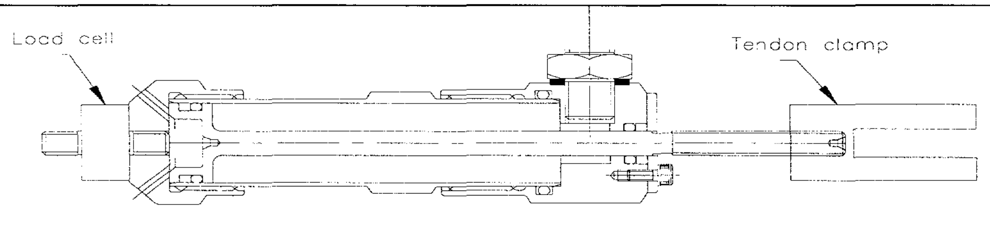


**Figure 9 - The Achilles Tension Cylinder**

2. Approximately 960N of tension (an estimate of the natural tension within the tendon in a live subject) applied to the Achilles tendon.
3. An "active" tension applied to the Achilles tendon based on the results of the simulator trials

The force in the tendon was increased until the desired active force was reached. When the tension was being increased the state of the tendon and clamping system was visually monitored to ensure that no tendon slippage through the fixing clamps occurred. The impact was initiated as soon as the tension remained stable. The foot was allowed to dorsiflex since the piston was held at a constant pressure with a large volume of air in the intensifier. (The 'passive' resistance of the piston and fluid is approximately 25N. With the intensifier activated the passive resistance increases to approximately 120-150N).

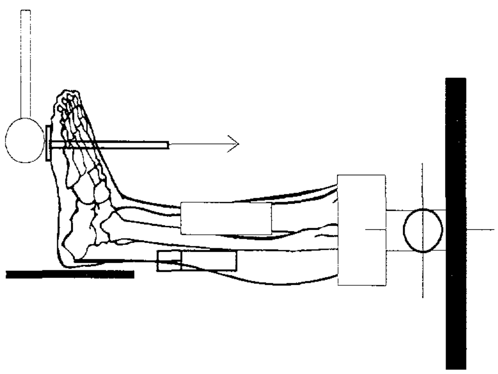

**Figure 11 – Schematic of PMHS Specimen with Achilles Loading Cylinder In-Situ**

**Testing Specification** -The PMHS specimens with an applied Achilles force were subjected to the toe impact tests from the EEVC certification procedure. The specimens were mounted and orientated as detailed above in the methods for the PMHS specimens. Impact tests were conducted at 4 and 6 m/sec. At these impact velocities, the energy delivered is normally below the injury threshold.

**Test Order** - The test order was kept the same for each specimen. The 'passive' 960N tests were performed first, then 'active' tension tests with 1800N or more, and finally the tests with no Achilles force.

**Biofidelity Results**

Overall, 78 tests were performed on 7 'through knee' amputation PMHS specimens, 7 tests were performed with an 'above knee' amputation specimen, 32 tests were performed with 6 volunteers and 60 comparable tests were performed on the two different dummy feet. The data from the important results are presented in this section and are discussed later in the paper.

**Physical Properties of PHMS Specimens** - The physical property data collected from the PMHS specimens used in this study are presented in Table 2 with comparisons to data collected in other institutes and for the dummy components.

**High-Speed Film Analysis** – A kinematic analysis from the high-speed films was carried out for all toe impacts on PMHS specimens, volunteers and the dummies. The films were analysed for dorsiflexion and time to peak dorsiflexion. The same two people analysed every PMHS, dummy and volunteer test to achieve consistency in the analysis. The dorsiflexion illustrated for each test type is the actual dorsiflexion corrected to a neutral starting point in each case. The high-speed film analysis is presented graphically, and on each graph, a one standard deviation corridor has been added.

PMHS Specimens - Figure 12 illustrates dorsiflexion for 'through knee amputation' specimens showing up to 34° of movement at 6 m/sec. Figure 13 illustrates the 'above knee amputated' specimen with up to 25° of movement. (Dotted lines show 61 standard deviation)

Volunteers - The results from tests with the volunteers are shown in Figures 14 and 15. Figure 14 illustrates the results where volunteers were unaware of the impending impact with up to 28° of movement at 6 m/sec shown. Figure 15 shows the results with the volunteer aware of the impending impact. In these tests the greatest range of movement was observed at 4 m/sec and was recorded to be 16° of dorsiflexion.

Dummies - Figure 16 illustrates the curves for the toe impact tests using the Hybrid III foot/ankle with up to 45° of movement shown and the GM/FTSS foot/ankle with up to 39° of movement shown.

Figure 17 shows an overlay of the results, where movement at the ankle can be compared for the volunteers, dummies and the PMHS specimens.

**Instrumentation Output** – The output from the pendulum accelerometer and the response of the leg, as measured by the tibia load cell, are presented as a series of overlay graphs in Figures 18-35. The graphs show the mean response of the PMHS specimens, GM/FTSS, Hybrid III and, where appropriate, the volunteers over-laid on the same axes. The graphs have been time-shifted and aligned on the principal peak in each case. Each curve is a plot of the mean response, which has been derived from the results of all the individual similar tests. Figures 18-23 show the pendulum acceleration, tibia force (Fz) and tibia bending moment (My) for the toe impact tests at 4m/sec and 6m/sec. Figures 24 & 25 show pendulum acceleration and tibia force (Fz) for heel impacts at 4m/sec. Figures 26-31 show pendulum acceleration, tibia force (Fz) and tibia bending moment (Mx) for inversion and eversion impacts at 4m/sec.

The graphs in Figures 32-35 show the mean peak values for pendulum acceleration, tibia force (Fz), bending moment (My) and dorsiflexion for the toe impacts, overlaid for comparison. In these graphs the parameters presented graphically are for toe impacts only.

A table of peaks of the relevant parameters for each test configuration is given in Appendix 2.

**Table 2.**
**Basic Physical Properties**

| Parameter | This Study | Crandall et al. | Hybrid III | GM/FTSS |
|---|---|---|---|---|
| **Age** | 66.75 σ 6.9 | | | |
| **Leg Length (mm)** | 454 σ 35 | 470 σ 41 | 442 | 442 |
| **Leg Volume (Litres)** | 2.629 σ 0.73 | | | |
| **Leg Mass (kg)** | 2.92 σ 0.75 | 3.19 σ 0.33 | 3.6 | 3.2 |
| **Foot Length (mm)** | 243 σ 11 | 244 σ 1.5 | 265 | 265 |
| **Foot Volume (Litres)** | 0.70 σ 0.15 | | | |
| **Foot Mass (kg)** | 0.75 σ 0.14 | 0.99 σ 0.12 | 1.48 | 1.11 |
| **Heel Flesh (mm)** | 16.86 σ 3.94 | | | |
| **Ball Foot Flesh (mm)** | 7.92 σ 0.689 | | | |
| **MoI Leg** | 0.07695 σ 0.02764 | | | |
| **MoI Instrumented leg** | 0.16468 σ 0.06154 | | | |
| **MoI Foot yy ($kg/m^2$)** | 0.00343 σ 0.00074 | 0.00474 | 0.0064 | 0.00409 |
| **MoI Foot zz ($kg/m^2$)** | 0.00342 σ 0.00076 | 0.00076 | 0.006325 | 0.0043 |

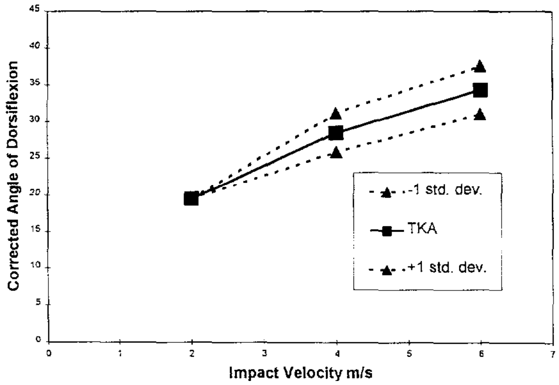


Figure 12 - Toe Impacts 'Through Knee Amputation'

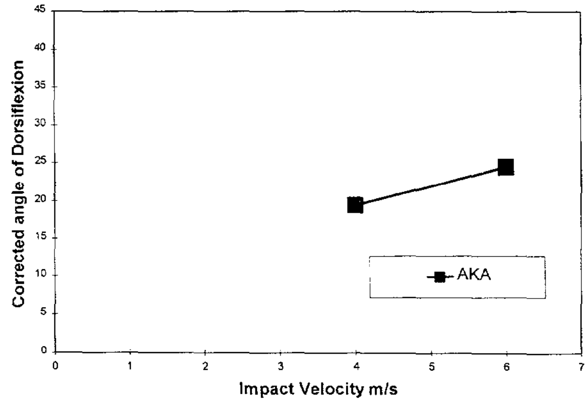


Figure 13 - Toe Impacts 'Above Knee Amputation'

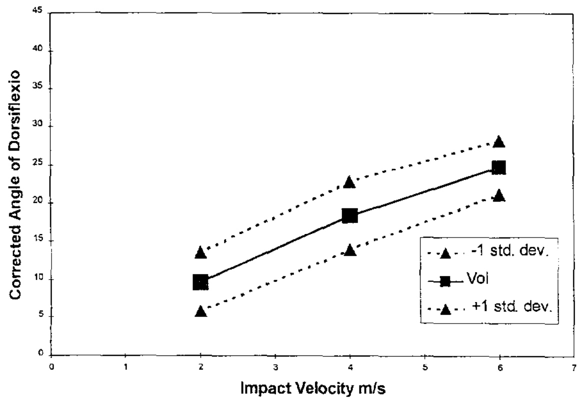


Figure 14 - Toe - Impacts Unaware Volunteers

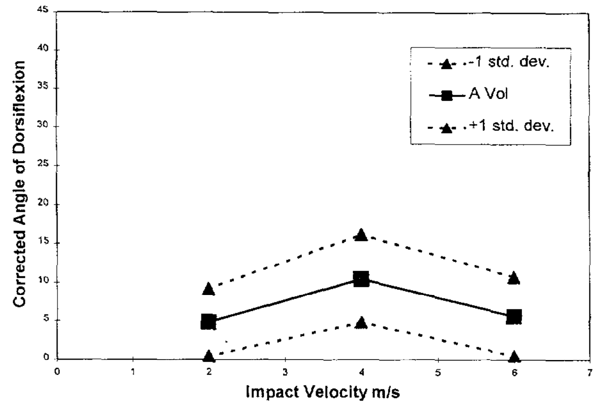


Figure 15 - Toe Impacts Aware Volunteers

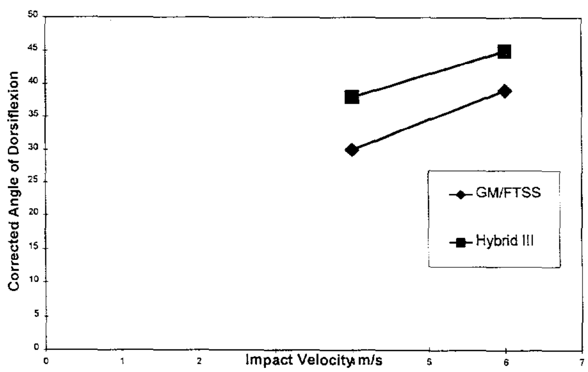


Figure 16 - Dummy Toe Impacts Dummies

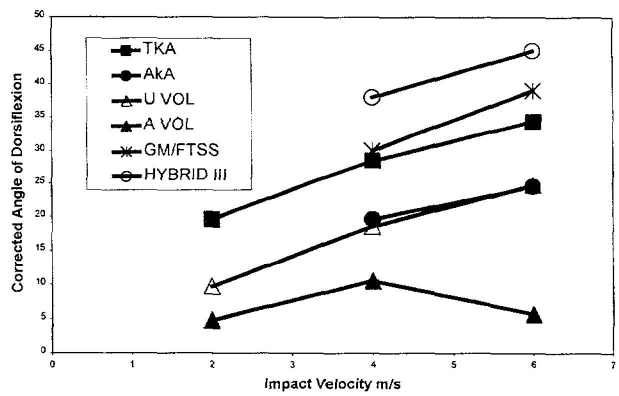


Figure 17 - Toe Impacts - All

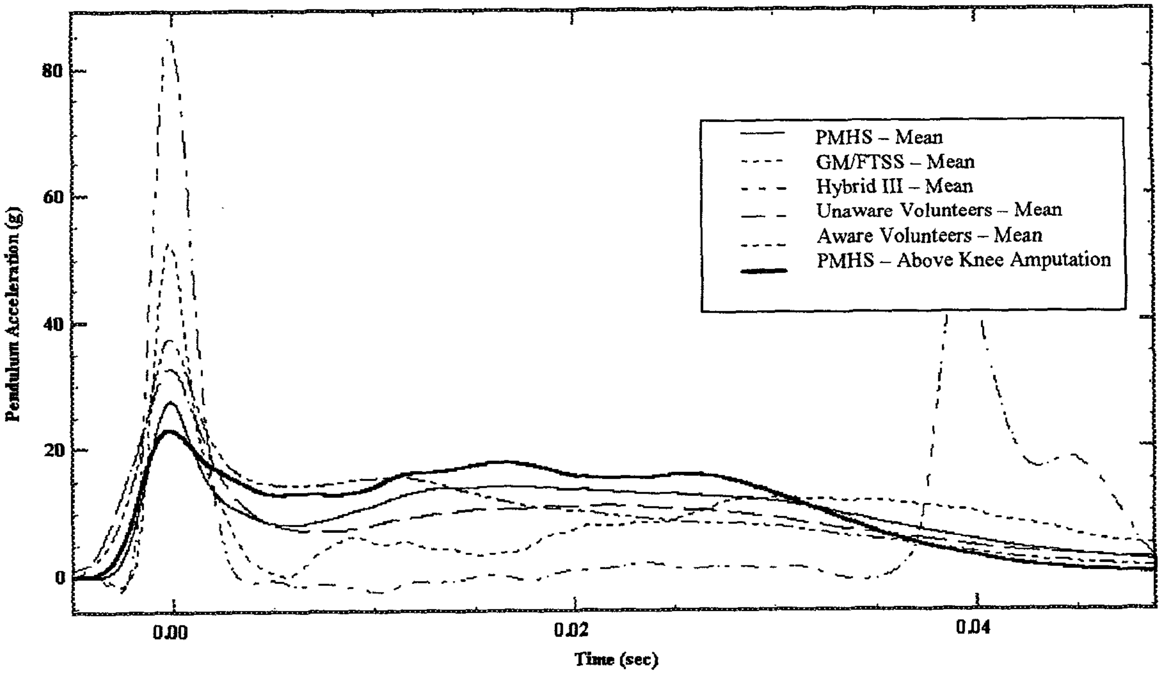


**Figure 18 - Pendulum Acceleration at 4m/sec - Toe Impacts**

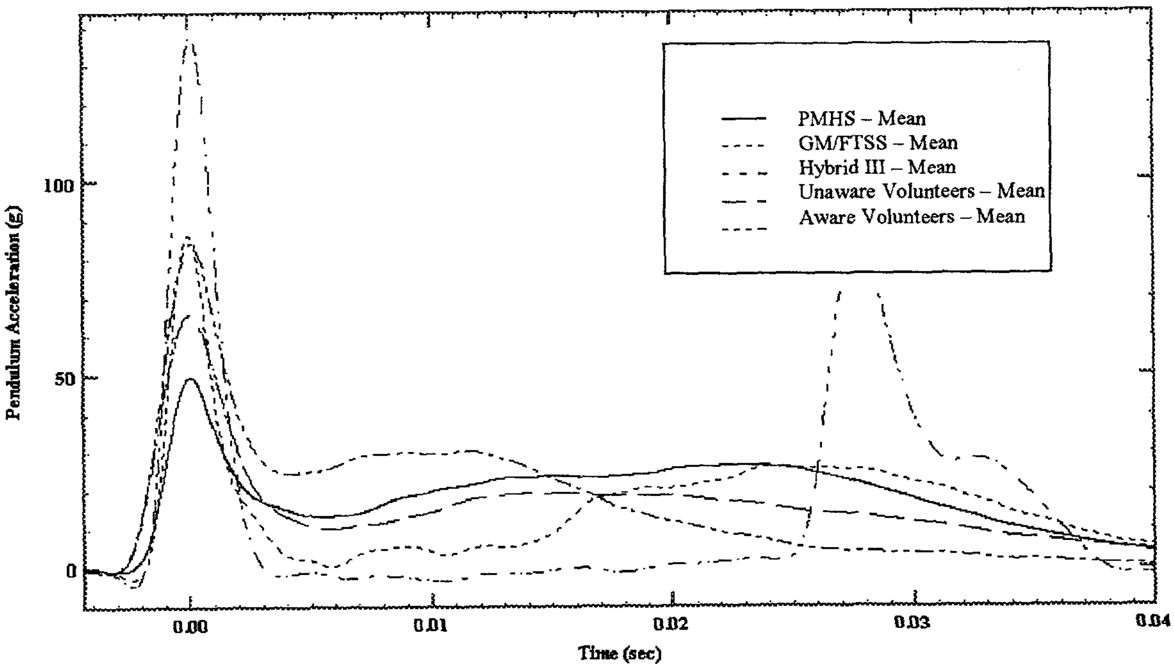


**Figure 19 - Pendulum Acceleration at 6m/sec - Toe Impacts**

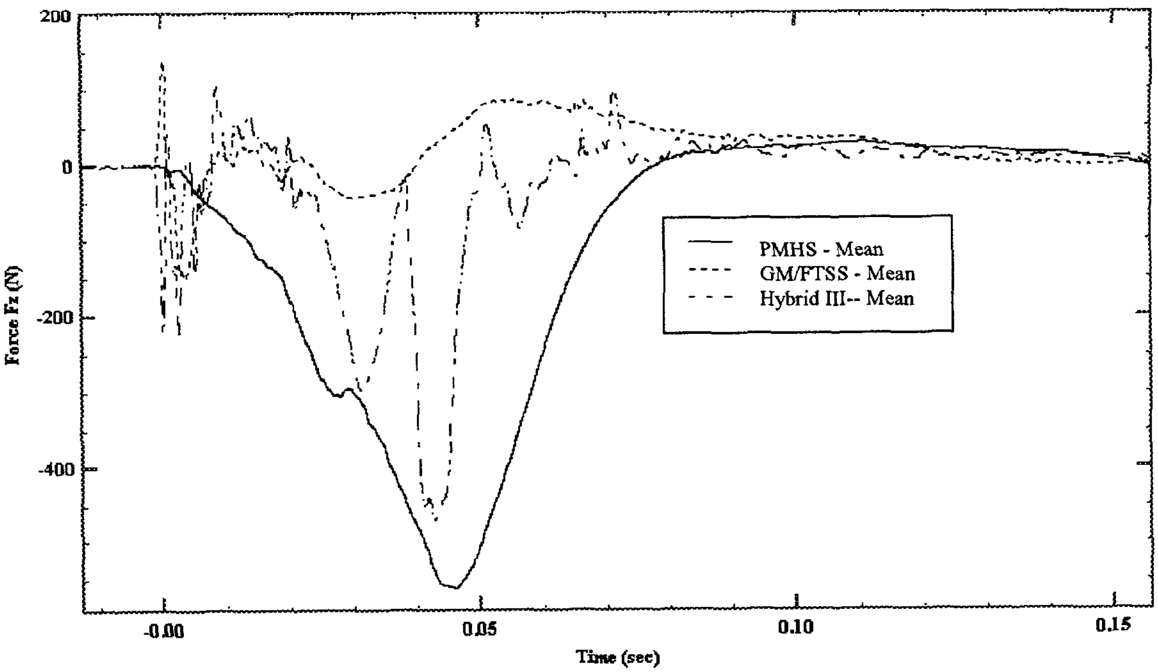


**Figure 20 – Tibial Force Fz at 4m/sec - Toe Impacts**

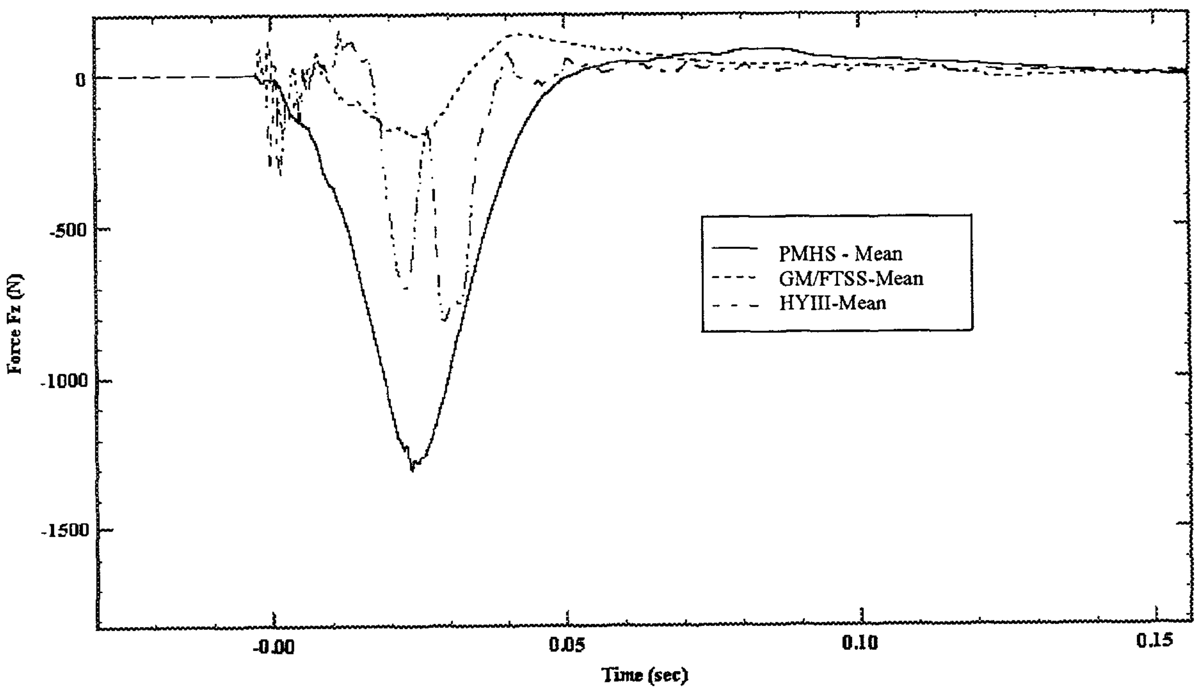


**Figure 21 – Tibial Force Fz at 6 m/sec - Toe Impacts**

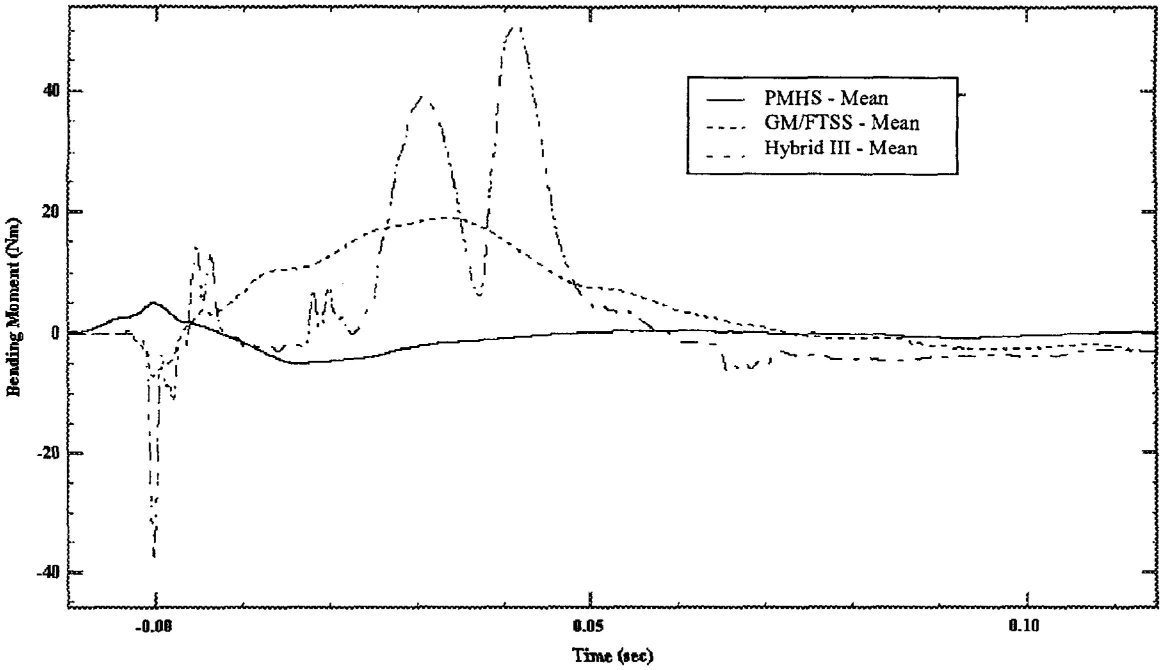


**Figure 22 – Tibial Bending Moment My at 4 m/sec - Toe Impacts**

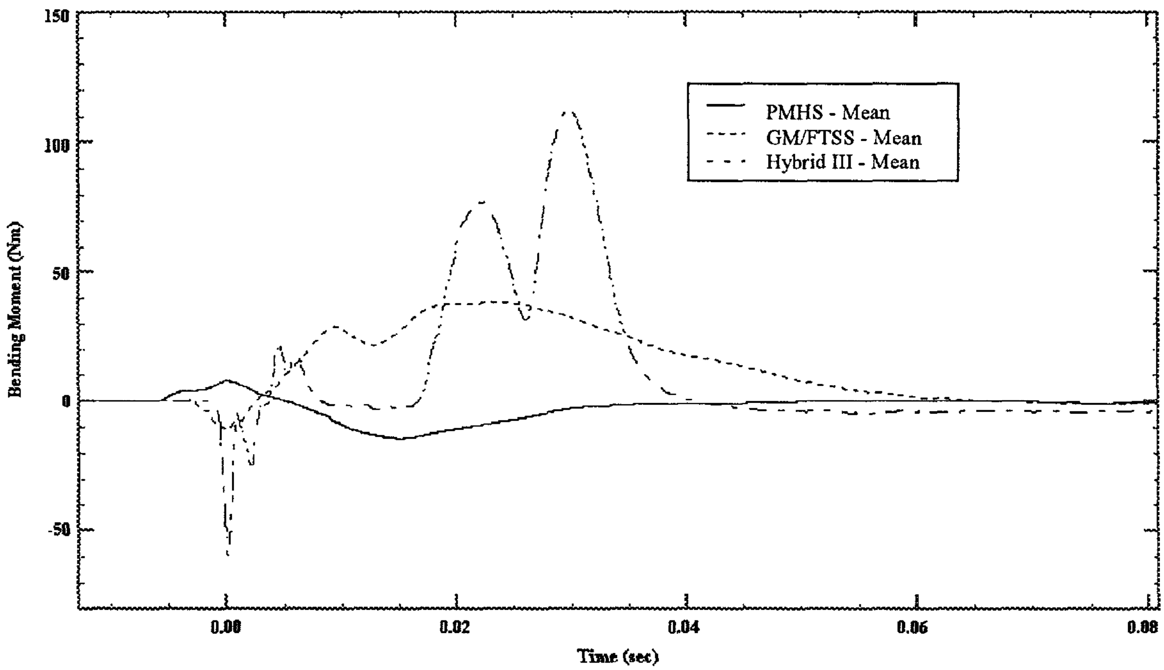


**Figure 23 – Tibial Bending Moment My at 6 m/sec - Toe Impacts**

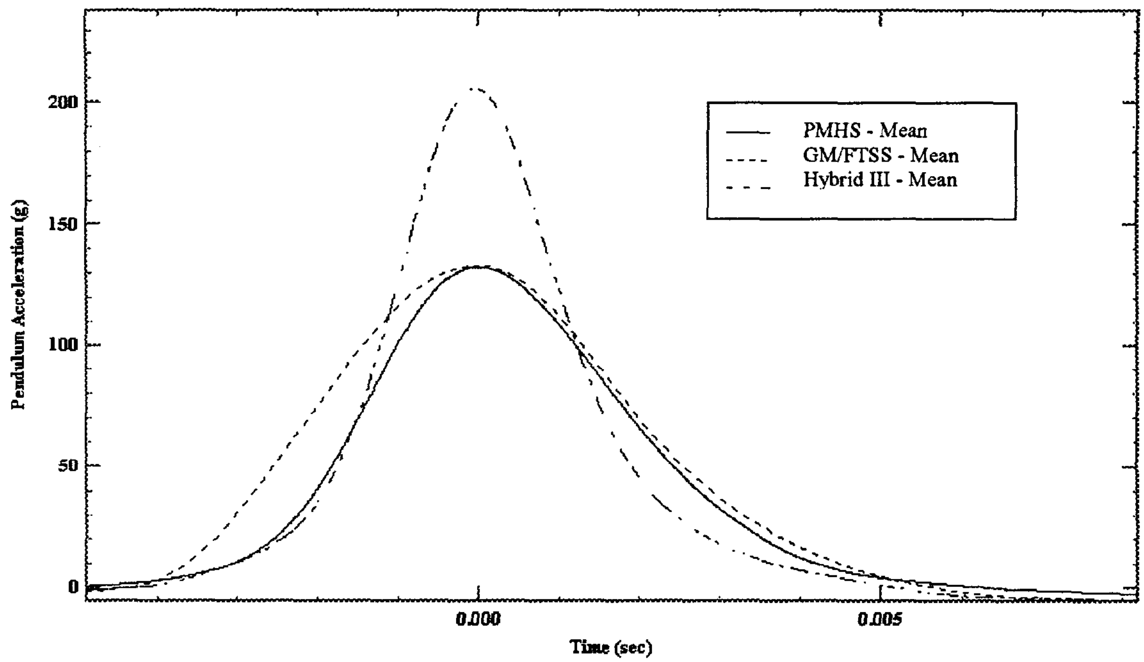


**Figure 24 – Tibial Pendulum Acceleration at 4 m/sec, Heel Impacts**

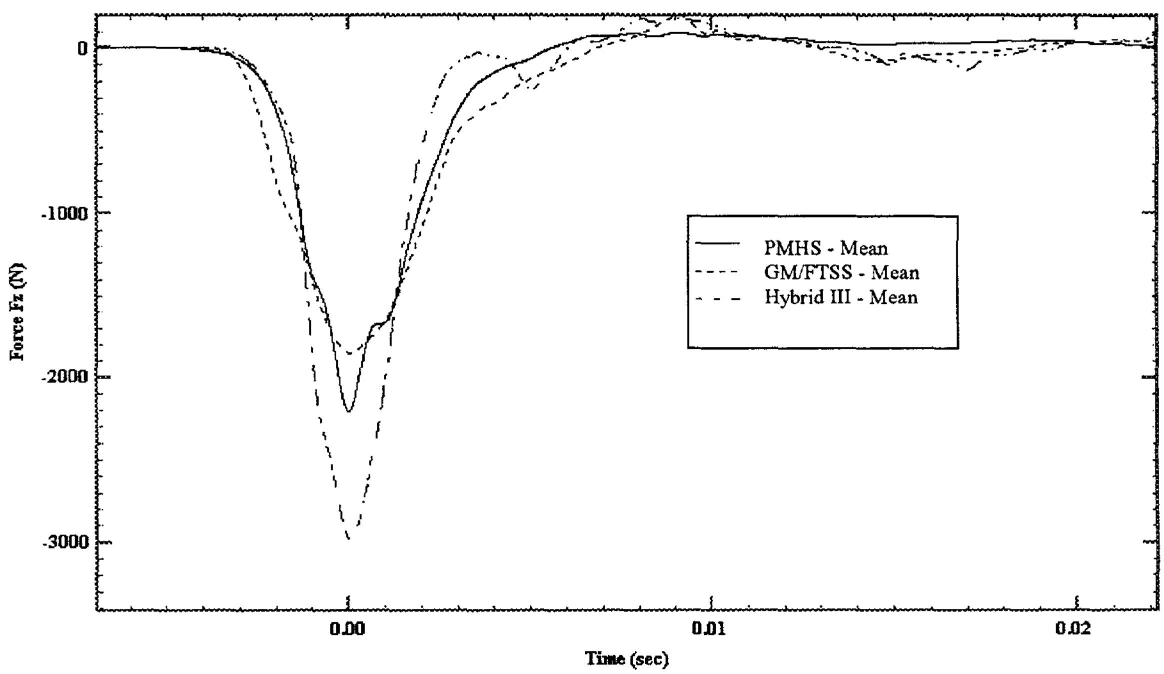


**Figure 25 – Tibial Force Fz at 4 m/sec, Heel Impacts**

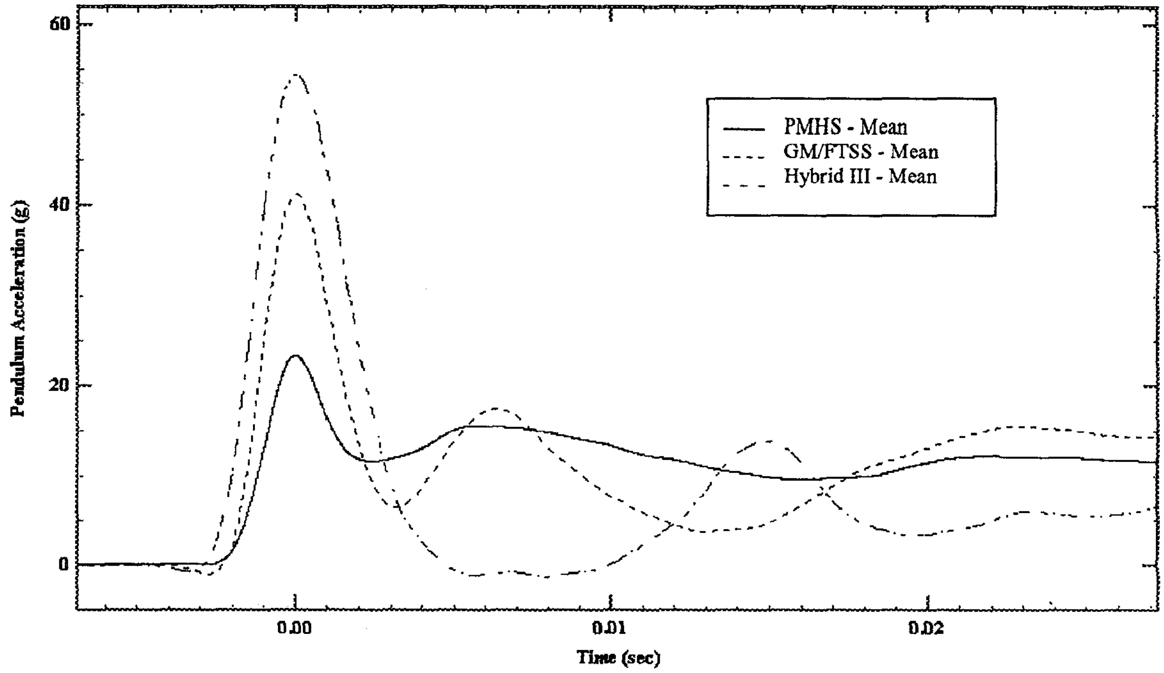


**Figure 26 - Pendulum Acceleration at 4 m/sec, Inversion Impacts**

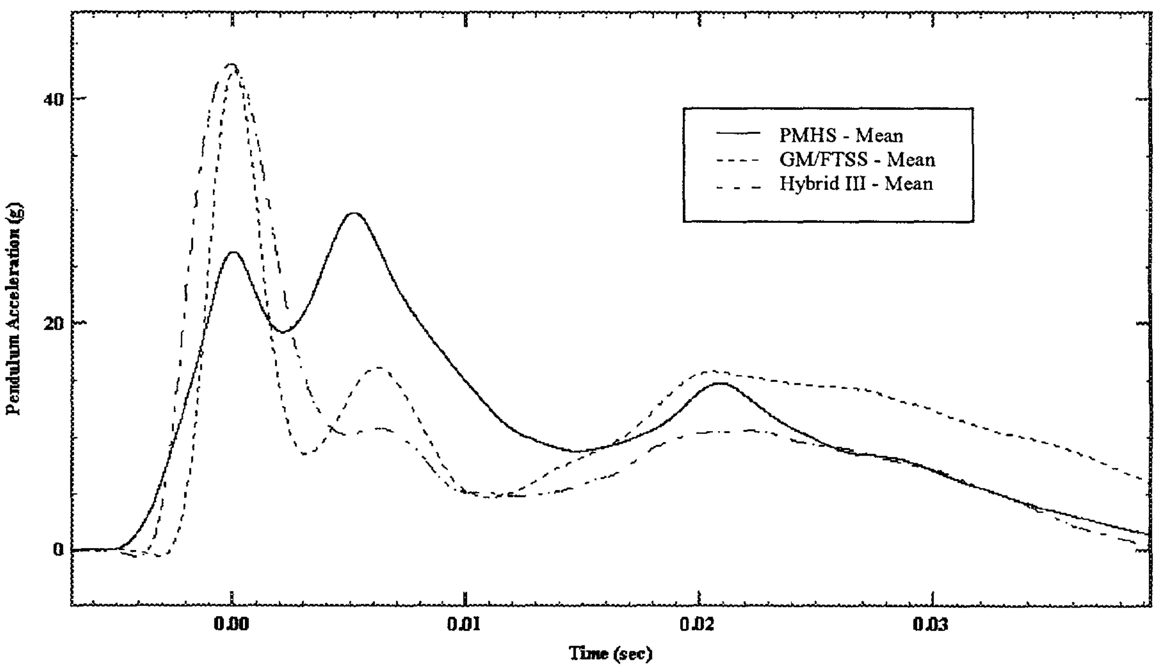


**Figure 27 - Pendulum Acceleration at 4 m/sec, Eversion Impacts**

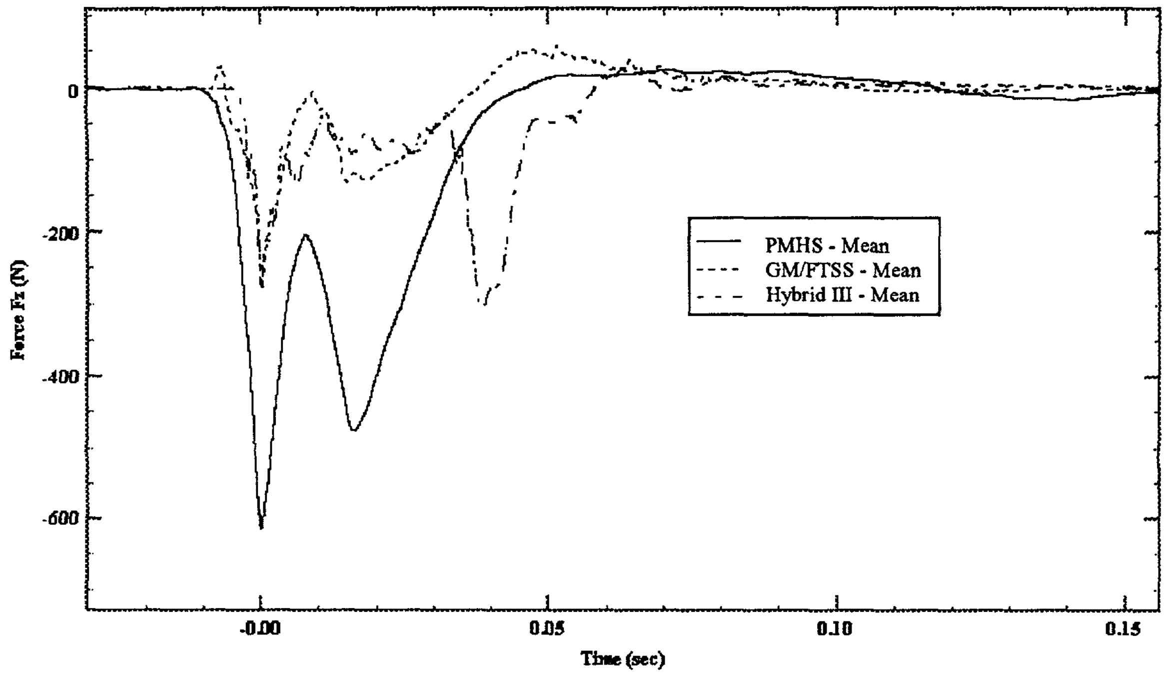


**Figure 28 – Tibial Force Fz at 4 m/sec Eversion Impacts**

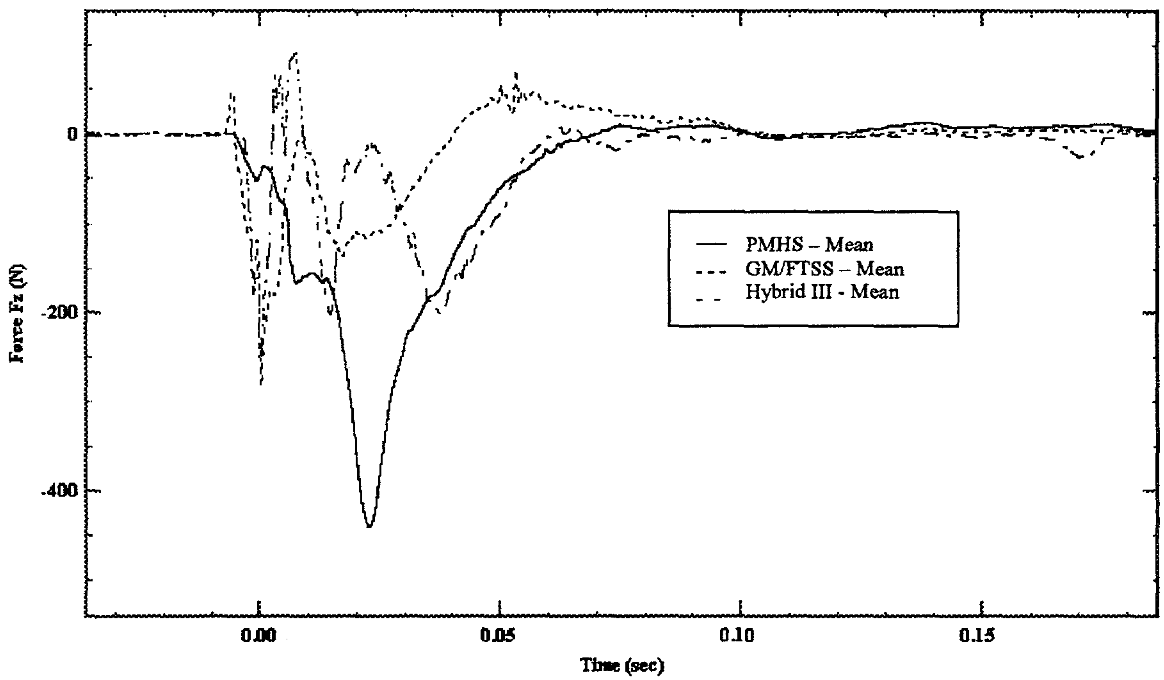


**Figure 29 – Tibial Force Fz at 4 m/sec Inversion Impacts**

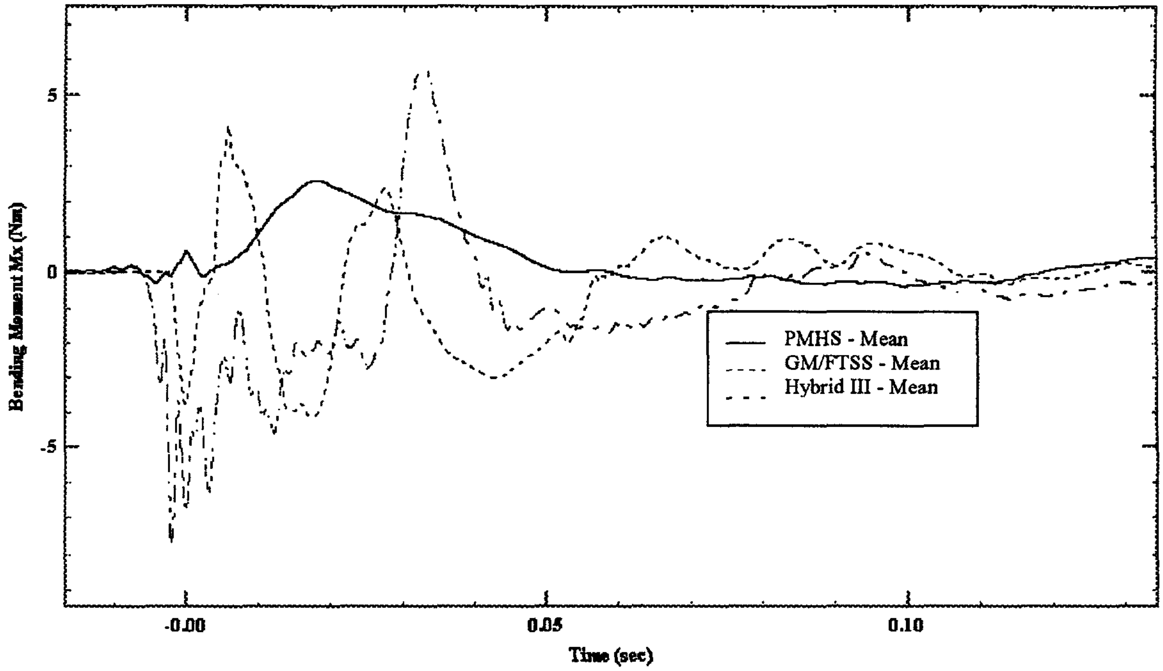


**Figure 30 – Tibial Bending Moment Mx at 4 m/sec - Eversion Impacts**

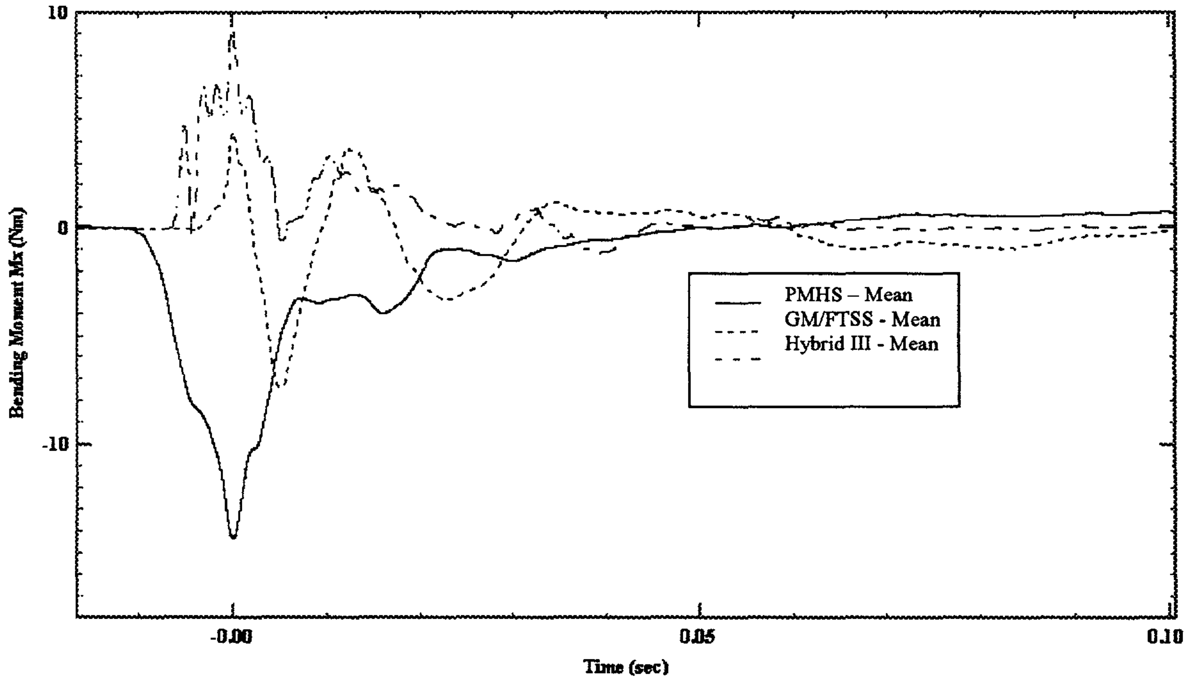


**Figure 31 – Tibial Bending Moment Mx at 4 m/sec - Inversion Impacts**

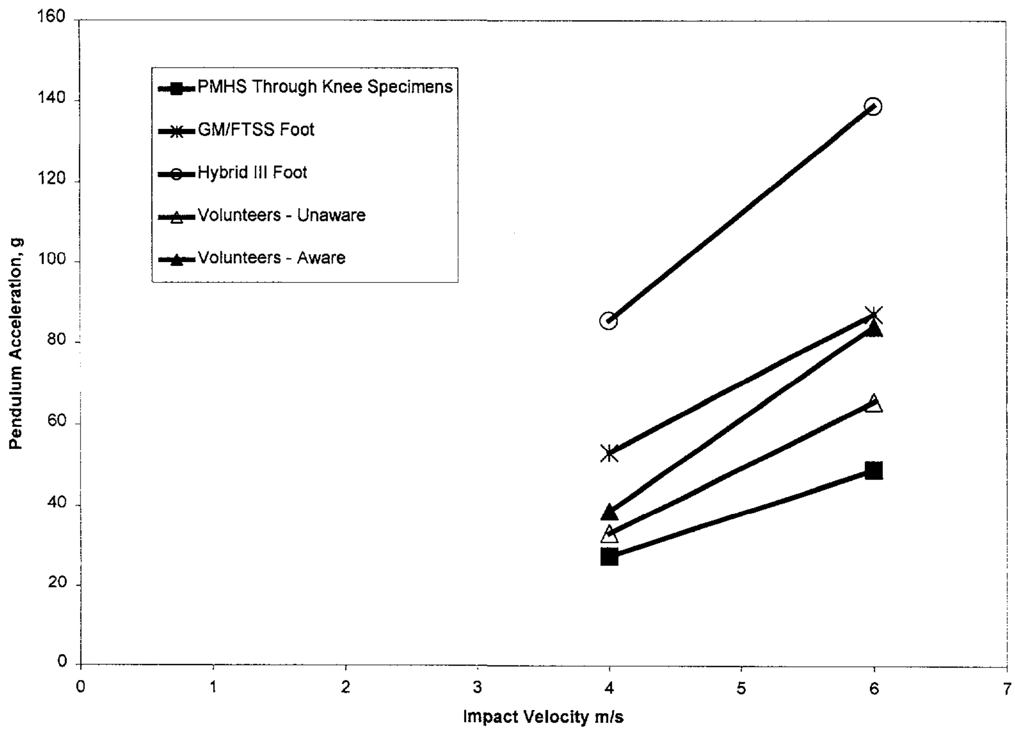


**Figure 32 – Mean Peak Responses for Toe Impacts – Pendulum Acceleration**

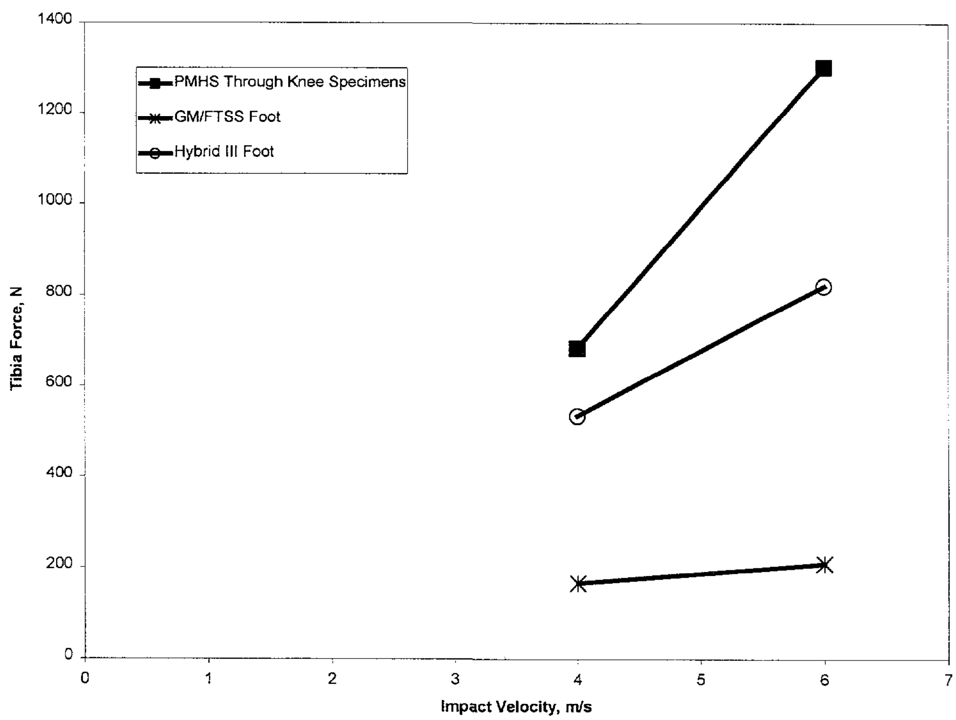


**Figure 33 – Mean Peak Responses for Toe Impacts – Tibia Force Fz**

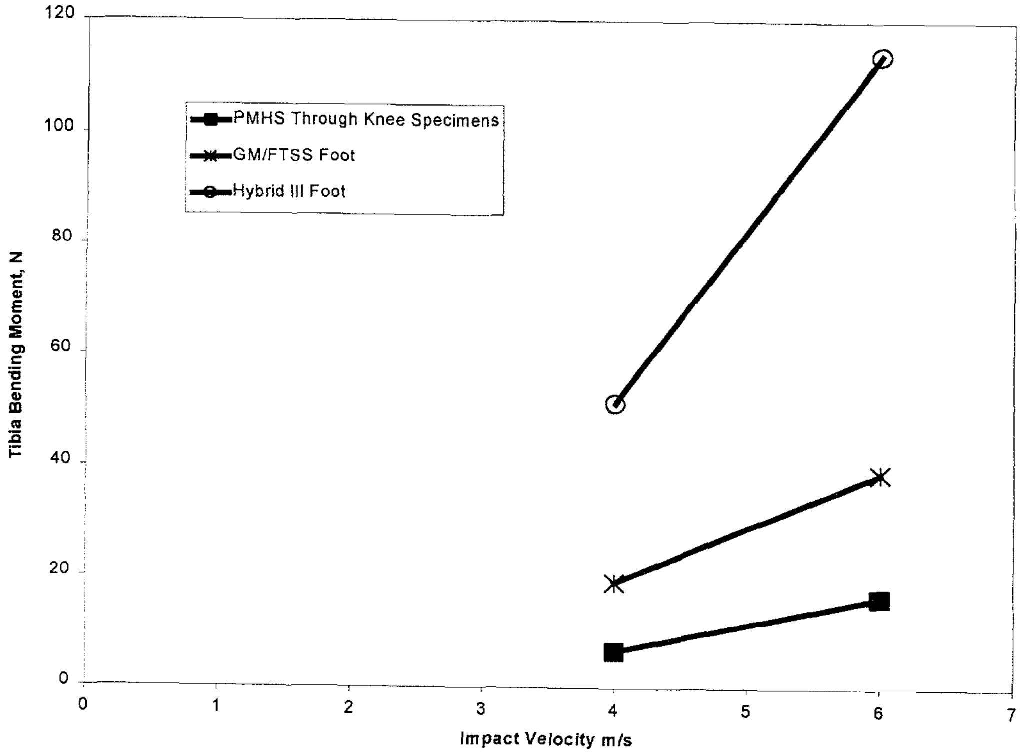


Figure 34 – Mean Peak Responses for Toe Impacts – Tibia Bending Moment My

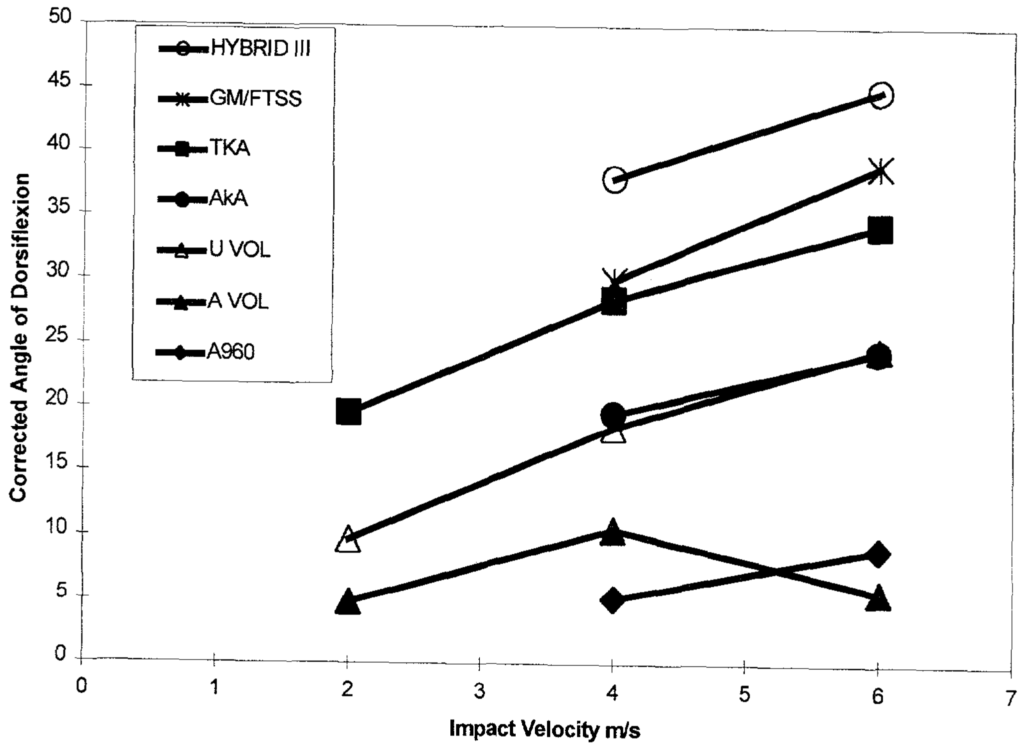


Figure 35 – Mean Peak Responses for Toe Impacts – Dorsiflexion

## Normalisation

The results of any PMHS specimen tests will vary somewhat due to differences in mass, dimensions and physiologically related parameters. To achieve an estimation of a mean response from a PMHS specimen in this study, a process of normalisation to the 50th percentile was attempted.

The approach taken for this work was to adapt a simple FE skeletal leg model in order to simulate the test conditions. Each PMHS specimen test was then modelled so that a comparison could be made to a similar test on a fiftieth percentile specimen. By comparing the results of the two models a function could be developed that could be applied to the actual test results, in order to normalise them to the fiftieth percentile.

The model comprised a rigid body representation of the leg and foot bones, with accurate geometry and four joints: hip, knee, ankle and sub-talar joints. Each of these joints was represented by simple rotation about a fixed axis or a fixed point, in the case of the spherical hip joint. The joints were each assigned an appropriate stiffness characteristic.

The model included, a tibial load cell, a spring/damper system to represent the plantar flexing muscle groups and the Achilles tendon. Flesh was added to the sole of the foot with similar properties to those of 'Confor foam'.

After the model had been prepared and having verified it against a number of individual tests, the PMHS specimen data was normalised as follows. Tests with each PMHS were modelled, in terms of their geometry, test conditions (impactor velocity, position and orientation), mass and tibial load cell location. The responses from the model ($F_Z$, $M_y$, $M_X$ and pendulum acceleration) were then obtained. A fiftieth percentile specimen was then modelled in the same test conditions. For each specimen tested, the results of the PMHS model were compared with the corresponding fiftieth percentile model. By noting the relationship of peak values between the two sets of model results, scaling factors were calculated to bring the test specimen model results to those of the fiftieth percentile model. These scaling factors were then applied to the actual test results to obtain normalised data.

## Normalisation Results and Discussion

The normalisation procedure was designed to factor the results of the toe and heel impact tests to represent those of the fiftieth percentile. However, the initial output from the process indicated that the results were not converging as expected. Some examples of the effect of the procedure described are shown in Figures 36-38.

Figure36 shows a relatively successful normalisation of the pendulum acceleration magnitude from a toe impact test. The scatter in the results has been significantly reduced as intended. Figure 37 indicates some degree of normalisation on time scale for a similar test. However, in some cases such as that shown in Figure 38, the normalisation process did not appear to have any effect on the scatter of the results.

The premise for normalisation of results in the fashion described is that the results are related to the properties of the PMHS specimen in some way. For instance, it might be assumed that the tibia bending moment is proportional to a combination of bone dimensions, masses and the properties of the flesh, underneath the pendulum impact point. Thus by including these parameters in the normalisation process, their effect can be taken into account and the results aligned. However, if there were no such relationship, it would not be possible to normalise the results. A regression analysis was carried out between the results and the normalisation parameters. While some correlations did result from the analysis, relationships that were expected to be strong (such as that between pendulum acceleration and foot mass) did not occur but other relationships had unexpectedly high correlations. Thus the results remained inconclusive.

Other factors may have influenced the success of the technique as well. The model may have been over-sensitive to a particular parameter and therefore the response of the model would have been unrealistically influenced by this one parameter. This issue was addressed through a sensitivity study on the model. The results of which showed the model to be not unduly sensitive to any one parameter. The normalisation process would also fail, if the test results were being influenced by a parameter that was not taken into account in the normalisation process, such as joint stiffness. The parameters that were included in the procedure had been chosen as being those thought most likely to influence the results, but with the proviso that it was possible to measure them on the PMHS specimens. Some potentially influential factors such as joint stiffness were not included.

It would be surprising if there were no relationship at all between the results and the PMHS specimen properties. However, this relationship may be non-linear and variable and hence it may have been over-optimistic to expect to be able to remove the variability from the results.

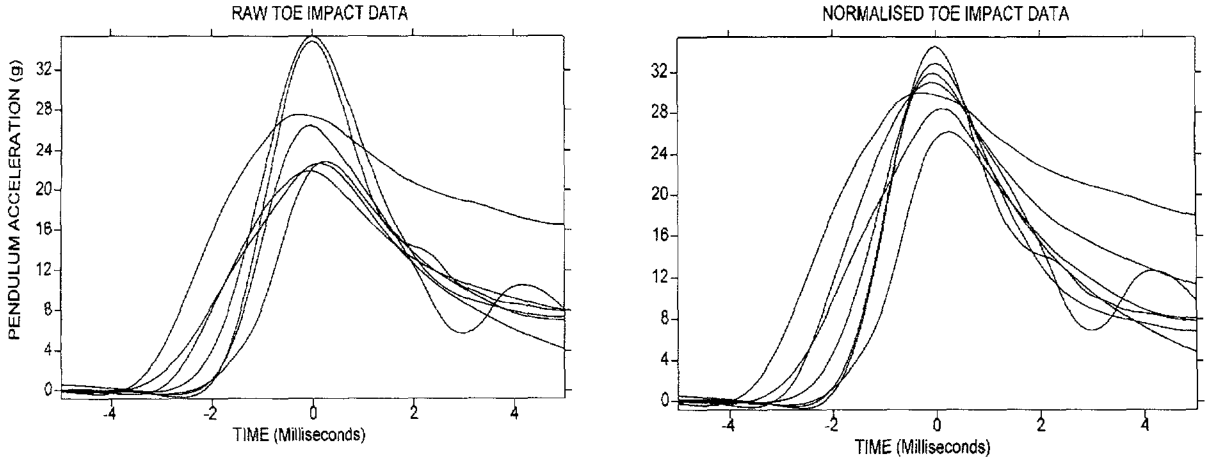


**Figure 36 – Normalisation of Peaks with Respect to Magnitude for Toe Impact Pendulum Acceleration at 4m/sec**

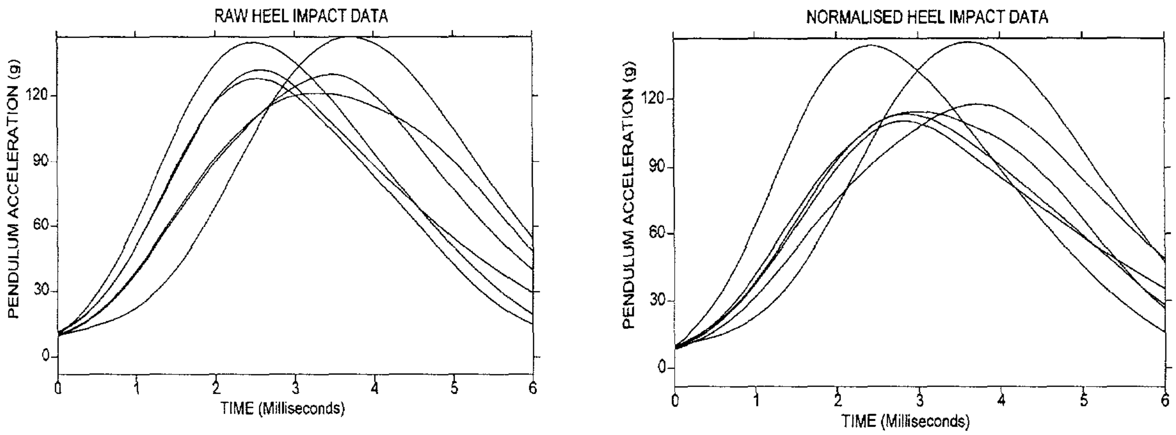


**Figure 37 – Normalisation of Peaks with Respect to Time for Toe Impact Pendulum Acceleration at 6m/sec**

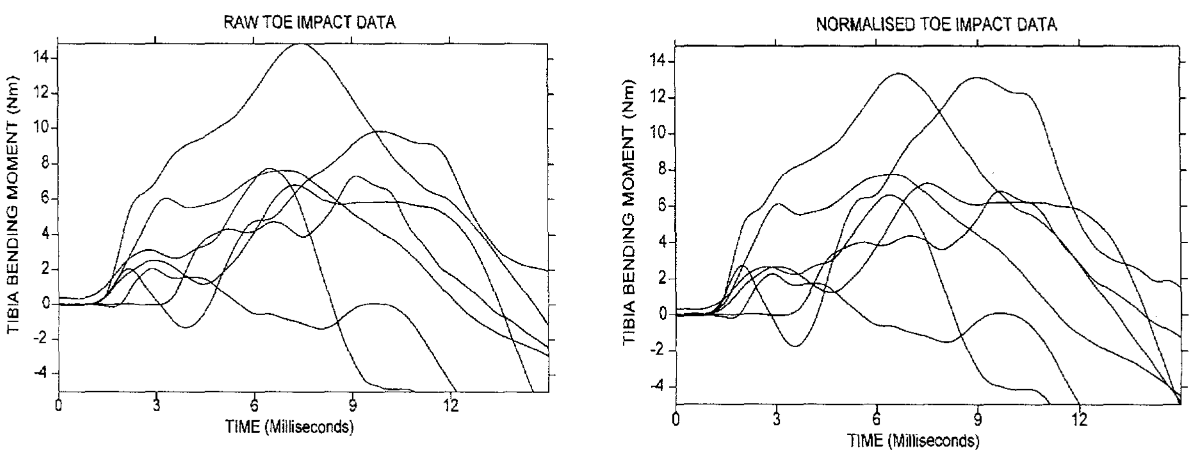


**Figure 38 – An Example Showing the Normalisation Resulting in a Limited Effect on the Scatter of Results**

**Influence of Achilles Tension**

The data for tests with an applied Achilles tension is presented here as a peak value analysis. This has enabled paired-t test comparisons to be made as the same specimens were used both with and without an applied tension. Only toe impacts were performed with the Achilles loading cylinder in-situ.

**Axial Loading Fz** - Table 3 shows the test conditions for which there were significant differences in Fz. A Through Knee Amputation is referred to as TKA, a specimen with the Cylinder In-Situ is referred to as CIS, with the force applied following, e.g. CIS - 960N. The difference between CIS-0 and CIS->1800 is significant only at the 10% level.

**Table 3.**
**Significant Differences in $F_z$**

| Test Conditions | Mean Difference | P Value |
|---|---|---|
| TKA : CIS-960 | 1735N | 0.001 |
| TKA : CIS->1800 | 3234N | 0.039 |
| CIS-0 : CIS-960 | 1507N | 0.007 |
| CIS-0 : CIS->1800 | 2313N | 0.064 |

**Tibial Bending Moment** - There were no apparent significant differences in the peak tibial ($M_y$) bending moment recorded for any of the comparative tests.

**Ankle Dorsiflexion** - Table 4 shows the test conditions for which there was a significant difference in the degree of dorsiflexion.

**Table 4.**
**Significant Differences in Dorsiflexion**

| Test Conditions | Mean Difference | P Value |
|---|---|---|
| TKA : CIS-960 | 24° | 0.001 |
| CIS-0 : CIS-960 | 14° | 0.045 |

**Mean Impactor Force** - Table 5 shows the test conditions for which there were significant differences in mean impactor force.

**Table 5.**
**Significant Differences in Mean Impactor Force**

| Test Conditions | Mean Difference | P Value |
|---|---|---|
| TKA : CIS-960 | 244N | 0.001 |
| CIS-0 : CIS960 | 311N | 0.016 |

**Biofidelity Testing Discussion**

The data presented in Table 2, comparing the basic physical parameters of the specimens used in the testing shows that the specimens used are comparable to those used in other institutes. It is recognised that a population with a mean age of 67 years will not reflect the driving population at large, particularly with respect to bone mineral density. The effect of ageing on bone quality has been well studied and it is recognised that an elderly population will tend to have weaker bones. It is for this reason that the data on bone mineral density has been collected. It is hoped that the continued collection of data with such experiments will allow for a definitive reference range to be established.

In further work, planned within this project, it is hoped to be able to normalise the results of any individual PMHS test and to correct the tolerances established for injury to mean population bone mineral density data.

The data for moments of inertia differ more between institutes and it is probable that this reflects differences in measuring techniques, as differences in dummy feet measurements have been observed compared to published results.

Over 170 different tests are reported in this paper all designed essentially to establish appropriate biofidelity requirements and to compare the performance of the two dummy designs with various human surrogates. Repeated PMHS testing was used in this study at a sub-injury level. In every case an orthopaedic surgeon examined the specimen after the conclusion of each test. Careful attention was paid to the ligamentous structures around the ankle joint. Testing with any one specimen was halted immediately that any injury became apparent, however minor.

**Toe Impacts** – Figures 12 – 23 illustrate the response of the foot to toe impacts.

Dorsiflexion - The kinematic analysis of the high speed films illustrated in Figure 17 allows a direct comparison between the different specimen types for ankle articulation in impact to be made. 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 and finally the aware volunteers. It can be observed that the mean GM/FTSS results lie close to the upper one standard deviation limit for the through knee PMHS tests (Figures 12 & 16). In a similar way, the responses of the unaware volunteers are comparable to those of the above knee amputated specimen (Figures 13 & 14). When a tension of 960N was applied to the Achilles tendon of a PMHS specimen, the response is similar to that of an aware volunteer (Figure 35). Although the GM/FTSS is close to the flaccid through knee amputation PMHS results, both dummies show a higher dorsiflexion response than volunteers, the above knee PMHS specimen and the PMHS specimen with a passive (960N) force applied to the Achilles tendon.

The degree of dorsiflexion will to a certain degree be influenced by the stiffness of the ankle joint. In reality, and as the results of the simulator trial have indicated, there is likely to be an active tension in the lower leg plantar flexing muscle group, which will influence the kinematics of the foot. The impactor tests have shown that the application of tension through the Achilles tendon results in a decrease in the amount of dorsiflexion observed for a given impact energy. The unaware volunteers exhibit less dorsiflexion than the flaccid PMHS specimens and the dummy feet. It is very difficult to assess accurately the amount of muscle activity and its influence in human volunteers. It is possible that there may have been some degree of involuntary bracing even in the 'unaware' state. The aware volunteer dorsiflexion results showed some correlation with the PMHS that had 960N of applied force in the Achilles tendon. In PMHS specimens with 960N applied, 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 is applied 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.

Pendulum Acceleration - The profile of the response curves of the dummies, volunteers and PMHS are quite similar. The Hybrid III dummy exhibits a second peak in the response which is characteristic of the foot design (Figures 18 & 19). The first peak results from the pendulum contacting the foot, the second peak arises from the fact that the design of the foot/ankle allows the foot to rotate relatively freely until it impacts the tibia. This response has improved over previous design with the addition of the bump stop and the increase in the joint articulation to 45 degrees.

The response of the GM/FTSS foot was closer to the volunteers and the PMHS results. The difference was not significant when comparing with volunteers, but it was for PMHS specimens ($p>0.01$ for GM/FTSS v. volunteers and $p<0.001$ for GM/FTSS v. PMHS). The Hybrid III results were further away from the volunteers and PMHS results. These differences were significant, except for the comparison with aware volunteers at 4m/sec ($p=0.009$). The difference between the pendulum acceleration observed with the two dummies was also significant ($p<0.001$).

The proposed design of the ALEx II leg incorporates a simulated Achilles tendon, which when tensed will enhance ankle stiffness in a biofidelic manner. It is hoped that the response seen in testing will be closer to the results seen with PMHS specimens and volunteers. At the same time the bending moment imparted to the tibia by the ankle will hopefully remain unchanged. The success of the GM/FTSS foot in simulating the response of the PMHS specimens is due to its rubber bump stop which is incorporated within its ankle hinge.

Tibial Force Fz – Although the peak Hybrid III response was closer to the PMHS results than the GM/FTSS response, the difference between the PMHS response and that seen by the two dummies is striking, when the graphs of Fz for toe impacts are examined (Figures 20 & 21). At 4 m/sec, the difference between the PMHS specimens and the GM/FTSS leg is statistically significant ($p>0.005$), although between the PMHS and the Hybrid III it is not. The difference between the two dummy types is also significant at the fifth percent level ($p<0.001$). The low value seen with the GM/FTSS foot can be attributed in part to the rubber component of the joint. At 6 m/sec a similar pattern of behaviour is seen, however the difference is significant with both dummies, when compared to the PMHS specimen.

The accident analysis that is reported later in this paper, has highlighted axial loading as a significant cause of injury for the most disabling of ankle fractures (pilon, calcaneal and talar neck). It would seem that the advantage of the GM/FTSS foot, with its stiffer ankle, is offset by an adversely recorded peak axial load. If the risk of serious injury is to be assessed, then the ability of future dummy feet designs, to measure accurately Fz biomechanically, will be paramount. The new proposed design for ALEx II will hopefully address these issues by imparting enhanced ankle stiffness without affecting the actual ankle joint.

A series of tests in the same experimental configuration, but with an added Achilles loading of 960N applied, have been reported previously. [6] It has been demonstrated that the observed axial load is significantly increased by the application of a bracing force through the Achilles tendon. Thus while the Fz response observed in flaccid PMHS specimens is significantly different from current dummy designs. it is speculated that the theoretical response in tensed volunteers would be even further removed. One of the fundamental differences in the geometry of the human leg, compared with the design of the Hybrid III, is that the tibia is offset at an angle between the knee and ankle joints. This aspect of the dummy design will also be an important factor in the interpretation of the load cell data.

Bending Moment My - When bending moments for toe impacts are considered, the time history curves (illustrated in Figure 22 & 23) show that PMHS specimens recorded a lower bending moment (My) than the Hybrid III leg with either the Hybrid III foot or GM/FTSS foot attached. A double peak was seen with the Hybrid III foot and this was again due to the design of the ankle complex, which allows the foot to move relatively freely after the first impact until the foot hits the rubber bump stop. In the GM/FTSS foot a plateau effect is observed corresponding to the continuous resistance at the ankle joint of this dummy. These data suggest that neither dummy records bending moments, in a biofidelic fashion. This is in part due to the offset of the tibia in the dummy leg which will generate a bending moment when an applied compressive axial load (Fz) is applied. These results bring into question the reliance which can be made on tibial bending moment data for injury prediction in car crash tests with current dummy designs.

**Heel Impacts** – Figure 24 & 25 illustrate pendulum acceleration and tibial force for the heel impact tests on the PMHS and dummies. No heel impact tests were performed on volunteers nor to the PMHS specimens with imposed Achilles tendon force.

Pendulum Acceleration - Heel impacts were performed at 4m/sec on both PMHS and the two dummy designs. It can be seen that the pendulum acceleration response of the GM/FTSS foot is very similar to that of the mean PMHS response. The mean peak for both is approximately 130g. The mean peak for the Hybrid III foot is much higher at 206g and significantly different ($p<0.001$).

Tibia Fz - The second parameter analysed for heel impacts was tibia compressive force (Fz). In the dummy there are rigid connections between the foot/ankle/leg components whereas in the cadaver there is firstly the flesh and structure of the sole of the foot and then cartilage on both sides of the sub-talar and ankle joint all of which have energy absorbing properties. The tibia Fz response of the GM/FTSS foot is much closer to that of the PMHS specimens than the Hybrid III. The Hybrid III is significantly different ($p<0.001$) from the PMHS specimens, as is the difference between the two dummy types. Thus the foot design and rubber bump-stop incorporated in the GM/FTSS foot would appear to be more biofidelic in this test configuration than those used in the Hybrid III design.

**Inversion/Eversion Impacts** - There have to date been very few biomechanical tests designed specifically to assess inversion and eversion responses of the foot to impact. The simplified testing reported here was conducted as a derivation of the toe and heel impacts. The foot for all the tests was rested unconstrained on either its lateral or medial side. As such the foot was free to dorsiflex or plantar flex in each case. In the majority of these tests the foot was seen to dorsiflex in addition to either inverting or everting. This was considered to be more representative of human like behaviour, as it is known that with the human ankle fixed in a right-angle position, significantly less inversion or eversion is possible.

The graphs (Figures 26 & 27) illustrating pendulum acceleration for inversion and eversion impacts indicate a smaller magnitude of the response with PMHS specimens, compared to the two dummy designs. In inversion the difference between the mean peak response of the PMHS specimens and either of the dummies is statistically significant (PMHS v GM/FTSS $p=0.005$, PMHS v Hybrid III, $p<0.001$). In eversion, the peak responses of the two dummies are very similar but are higher than the PMHS response. This difference however, was not significant.

The mean difference between the two different dummy designs is also significant. When the impact tests are studied visually the lower centre of rotation of the GM/FTSS foot (illustrated in Figure 3) appears to give the foot a more human like movement. These data suggest that this movement is too stiff to give a similar response to the PMHS specimens. This testing has not been undertaken on volunteers and it is probable that with active muscle tension the response would be closer to that seen with the GM/FTSS foot.

The graphs illustrating the measured tibial force for inversion and eversion indicate a higher magnitude of the response recorded with the PMHS specimens when compared to the two different dummy designs. In inversion the differences between the mean peak tibia forces are not significant. The later peak seen in the PMHS specimens is possibly explained by the lack of a rigid connection between the foot and the leg and the need to move the ankle to tighten up the collateral ligaments before force can be conducted axially up the tibia. In eversion the mean peak tibia forces for the two dummies are similar however the usual double peak is seen with the Hybrid III foot corresponding to the foot hitting the bump stop in dorsiflexion. The mean peak force for the PMHS specimens (657N) is significantly higher than the dummies ($p=0.28$).

The graphs illustrating the Mx bending moment are more difficult to explain. In inversion the response of the PMHS is lower in magnitude than that of either of the dummies. This is to be expected with the sub-talar and ankle joint ligaments stretching to allow a more gradual transmission of force to bend the tibia. In eversion the PMHS appear to experience a much greater force. It is possible that this is an artefact of the way in which the eversion impacts were generated. It would be desirable to repeat these tests with a pure eversion force applied to a foot that was constrained to prevent dorsiflexion.

**Tests with an Applied Achilles Force** - Using the method described, it has been possible to generate forces up to 3300N in the Achilles tendon. Ferris et al [19], have been able to obtain the same magnitude of force within the Achilles tendon without failure using modified cryo-clamps. Other researchers have attempted to induce ankle torsion by direct application of torque through pins inserted into the calcaneus [7], but have been hampered by fractures of the calcaneus. At the University of Virginia pre-impact bracing has been induced by applying axial force through the lower leg [8]. The method described here produces bracing in a more biofidelic manner, by simulating a natural mechanism. It will allow the further evaluation of pedal induced injuries in the absence of heel to floor contact.

Ferris et al. [19] have reported that in studies on live subjects, the Achilles tendon contributes only 66 percent of the plantar flexing force at heel rise, the remainder being attributed to, five other muscles in the lower leg. In the PMHS test methodology described here, the whole plantar flexing force was generated through just the Achilles tendon, this may account for the high incidence of tendon rupture seen in this study. It is recognised that Achilles tendon strength decreases with age [20]. The specimens were taken from an elderly population and it is acknowledged that this will have had an effect.

This study reports a statistically (paired-t test) significant increase in axial loading and decrease in movement of the foot when a plantar flexing force is generated. The results support those of Petit et al. [7] who observed, in testing, an increased injury threshold for a dorsi flexing impact with an applied Achilles force.($47 \pm 17$Nm with applied force, $33 \pm 17$Nm without). Our results are also in agreement with Klopp et al. [8] who report an increase in axial loading three times the magnitude of the applied bracing force. By comparison, in this study the mean difference in the peak axial load was approximately twice the applied Achilles load. This supports the view of other researchers that a plantar flexing force in the Achilles tendon has more than a simple additive effect [8].

The significant increases in mean impactor force with an applied plantar flexing force, reported in this study, suggest that, in future biofidelity assessments of dummies, some account of bracing should be considered.

The finding that axial loading is increased by a magnitude significantly higher than the applied Achilles tension implies that, in bracing, car occupants could be at higher risk of these disabling fractures.

In this work specimens amputated through the knee were used with both heads of the gastrocnemius being detached. It has, however, been reported that the gastrocnemius does not contribute significant resistance when the knee is bent, as is usually the case when driving a car [9].

The method developed to apply an active plantar flexing force in PMHS specimens is being further developed. It is anticipated that the use of an active plantar flexing force will be required in future work particularly to generate pilon fractures.

## ACCIDENT ANALYSIS

There have been many recent analyses addressing lower leg injuries in frontal car trauma. The following accident analysis is unusual in that an emphasis has been placed on the review of the clinical notes by an orthopaedic surgeon, in order to attempt deduce more accurately the mechanism of injury. Injuries have been ranked according to their frequency, severity and impairment in order to determine which injuries are a priority for prevention.

A retrospective, case by case accident analysis has been conducted to determine the incidence and mechanisms of lower extremity injuries to front seat occupants, and to determine which injuries are a priority for prevention. This has been based on their potential for causing long term impairment and disability. This study is the largest in-depth analysis of lower extremity injuries to be conducted in the UK. The accident data were taken from the CCIS database with in addition; the hospital medical notes and x-rays being obtained. This enabled a detailed examination of the injuries and associated vehicle damage for each case.

The specific objectives of this study were to:

1. Identify the lower limb fractures and serious soft tissue injuries sustained by front seat occupants in frontal crashes.
2. Identify the source of injury from each vehicle.
3. Detail the precise mechanism of injury (i.e. the load path and the direction of forces transmitted to the lower limb).
4. Judge which injuries are priorities for prevention with respect to their high incidence or their potential for causing severe and long-term disability.
5. Recommend methods of reproducing the injuries observed in 'real world situations in a laboratory.

Data for 114 front seat occupants, with a total of 194 lower leg injuries were obtained for this analysis.

### Injury Severity

Most accident investigators use the Abbreviated Injury Scale (AIS) to code injuries to car occupants.

However, this scale categorises injures primarily on the basis of 'threat to life. It does not indicate the mechanism of injury or the likelihood of that injury being a cause of long term disability. An injury scale developed by the American Orthopaedic Foot and Ankle Society (AOFAS) was considered more useful in this respect and has been used in this study. An outline of the severity and impairment scoring system is shown in Table 5.

**Table 5.**
**The AOFAS Severity and Impairment Codes for Foot and Ankle Injuries**

| Score | Severity of Injury | Expected Functional Outcome |
|---|---|---|
| 0 | No injury | No Impairment<br>No residual signs or symptoms associated with the injury |
| 1 | Minimal Injury | Minimal Impairment<br>Able to do all desired activities, may be slightly limited at impact activities, occasional discomfort |
| 2 | Mild Injury | Mild Impairment<br>Unable to do impact activities, some limitations at work. Can't do a job requiring constant standing, walking or climbing |
| 3 | Moderate Injury | Moderate Impairment<br>Walking is limited. Can do most activities but unable to walk for long periods. May use cane for support occasionally |
| 4 | Severe Injury | Severe Impairment<br>Able to walk about living quarters. Usually can weight-bear but often needs walking aid (cane). Needs to sit most of time at work. Regularly uses medication to control pain. |
| 5 | Very Severe Injury | Very Severe Impairment<br>Can barely get around living quarters with out walking aids. Must use walking aids or wheel chair outside of house. Only able to work in limited jobs requiring no standing, walking or climbing. |
| 6 | Currently Un-treatable | Total Impairment<br>Unable to weight bear must use walking aid or wheel chair at all times. Unable to perform any type of work activities and/or household chores. Pain very poorly controlled |

**Lower Limb Injury Accident Analysis - Methods**

The cases for the retrospective study were taken from the existing Co-operative Crash Injury Study (CCIS) database. This database contains data collected by professional accident investigators, who study accidents that occur in specific sampling areas within the UK. A detailed examination of vehicle damage is made, and compared with the occupants' medical data from hospital records, occupant questionnaires and post-mortem reports, as appropriate. The accidents were sampled such that fatal and serious crashes were investigated, where possible. Thus the database was biased towards accidents that are more serious.

The data contained within the CCIS database lacked the necessary detail to determine the mechanisms and source of injuries to the lower extremities of front seat occupants. In order to determine this information, a more in-depth systematic case-by-case analysis of the CCIS cases was undertaken which included reference to the original hospital medical notes and X-rays.

For this study, the cases selected were frontal impacts, i.e. the principal direction of force was between 11 and 1 o'clock (any accidents involving a rollover or multiple impact were excluded). The front seat occupant had to have sustained an Abbreviated Injury Score (AIS) of two or more to the lower extremity.

**Retrospective Study Injury Types**

The injuries found from the retrospective investigation were first broken down according to frequency of occurrence (Table 6). The most frequently injured regions were the ankle (18.6%), femoral shaft (18.6%), and fractures of the talus (8%), tibial shaft (7%) and forefoot (7%). Here, 'ankle' refers to the articulation between the talus, tibia and fibula. Therefore fractures of the ankle include malleolar fractures and fractures of the distal tibia weight bearing area (pilon fractures).

**Table 6.**
**Frequency of AIS 2+ Injuries to the Lower Extremity by Anatomical Site for Subjects in the Retrospective Study**

| **Injury Region/type** | **Frequency** | **%** |
|---|---|---|
| Ankle Fractures | 36 | 18.6 |
| Femoral shaft | 36 | 18.6 |
| Patella | 20 | 10.3 |
| Talus | 16 | 8.2 |
| Forefoot | 13 | 6.7 |
| Tibial shaft | 13 | 6.7 |
| Tibial plateau | 11 | 5.7 |
| Supra-condylar | 7 | 3.6 |
| Hip dislocation | 6 | 3.1 |
| Acetabulum | 6 | 3.1 |
| Phalanges | 6 | 3.1 |
| Midfoot | 6 | 3.1 |
| Calcaneus | 6 | 3.1 |
| Lisfranc's joint | 5 | 2.6 |
| Femoral neck | 3 | 1.5 |
| Knee ligament | 2 | 1.0 |
| Pelvis | 2 | 1.0 |
| Total | 194 | 100 |

**Below Knee Injuries**

This study focused on below knee injuries, which constitute over one half of the total lower limb injuries shown in Table 6. These were defined as AIS 2+ injuries to the tibial plateau and below. For ease of reference, these will be referred to as below knee injuries, although the tibial plateau constitutes part of the knee. 112 'below knee' injuries to 78 occupants (65 drivers and 13 front seat passengers) were analysed. The ankle was the most frequently injured site. One third of these ankle injuries were pilon fractures (Table 7). By frequency alone, injuries to the ankle (32%), talus (14%), tibial shaft and forefoot (both 12%) and tibial plateau (10%) would appear to be the most important injuries. However, this does not take into account that many of these injuries are of low severity and are associated with a good outcome, and also that several injuries that occur relatively infrequently account for a large proportion of the most disabling injuries.

**Table 7.**
**Injury Frequencies by Site for All Below Knee Injuries.**

| **Injury** | **Frequency** | **%** |
|---|---|---|
| Ankle Malleolus | 23 | 20.5 |
| Talus | 16 | 14.2 |
| Ankle Pilon # | 13 | 11.6 |
| Tibial Shaft | 13 | 11.6 |
| Forefoot | 13 | 11.6 |
| Tibial Plateau | 11 | 9.8 |
| Midfoot | 6 | 5.4 |
| Phalanges | 6 | 5.4 |
| Calcaneus | 6 | 5.4 |
| Lisfranc's Joint | 5 | 4.5 |
| Total | 112 | 100 |

The primary injury mechanism for each of these injuries was deduced from analysis of the X-rays of each injury. The breakdown of mechanism of injury for each site is contained in Tables 8-11.

**Table 8.**
**Primary Mechanism of Fracture for Injuries to the Tibia and Ankle**

| Injury Site | Mechanism | Frequency |
|---|---|---|
| Tibial Plateau # | Axial Load + Varus/Valgus | 7 |
| | Axial Load + Rotation | 1 |
| | Axial Load + Combination | 3 |
| Tibial Shaft # | Bending | 5 |
| | Bending + Compression | 6 |
| | 4-Point Bending | 2 |
| Ankle Pilon # | Combined (Axial Load) | 13 |
| Ankle Malleolus # | Abduction | 11 |
| | Adduction | 8 |
| | External Rotation | 4 |

The majority of tibial plateau fractures were attributed to either a varus or a valgus force applied to the tibia (i.e. a force tending to rotate the tibia about the knee either towards [varus] or away from [valgus] the midline). This resulted in a unilateral tibial plateau fracture pattern. More complex loading mechanisms, implying greater axial compression and rotation were responsible for the bilateral plateau fractures.

The majority of tibial shaft fractures were attributable to a bending mechanism. A greater degree of axial compression of the tibia was implied by fragmentation at the fracture site.

Fractures of the ankle that involved only the malleoli were caused by rotation of the talus within the ankle mortise, as the foot was rotated about the tibia. Although common, these injuries are relatively straightforward to treat and are associated with a good outcome.

Fractures to the main weight bearing area of the distal tibia (pilon fractures) were caused principally by axial loads (i.e. loads in the direction of the long axis of the tibia) with secondary rotations of the foot ankle complex. In contrast to malleolar fractures, pilon fractures are very difficult to treat and often lead to poor outcome for the patient (Table 8).

The most important fractures of the hind-foot, fractures of the talar neck and intra-articular calcaneal fractures, are caused by axial loads (i.e. loads parallel to the long-axis of the tibia). Hind-foot injuries caused by inversion or eversion of the foot would not be expected to cause long-term disability, unlike fractures of the talar neck and calcaneus (caused by axial loading) which often leave the patient severely disabled (Table 9).

**Table 9.**
**Principal Mechanisms of Injury to the Talus and Calcaneus**

| Injury Site | Mechanism | Frequency |
|---|---|---|
| Talar Neck # | Axial load and Dorsiflexion | 9 |
| Talar Avulsion # | Inversion/Eversion of Foot | 3 |
| | Dorsiflexion and Inversion | 4 |
| Intra-articular Calcaneus # | Axial Load | 5 |
| Calcaneus Avulsion # | Inversion/Eversion of Foot | 1 |

Fractures to the mid-foot and the Lisfranc's joint were caused by indirect loading to these structures and excessive bending in abduction, adduction and plantar flexion (Table 10).

**Table 10.**
**Principal Mechanisms of Injury to the Mid-foot and *Lisfranc's Joint***

| Injury Site | Mechanism | Frequency |
|---|---|---|
| Mid-foot # | Medial Stress | 1 |
| | Lateral Stress | 5 |
| Lisfranc's Joint # | Plantar-flexion / Abduction/Adduction | 5 |

In contrast, injuries to the forefoot region are usually a result of direct trauma to the fractured area (Table 11). Often damage is localised to a small area (e.g. one or two metatarsal necks only) suggesting a small area of concentrated load.

**Table 11.**
**Principle Mechanisms of Injury to the Forefoot (Metatarsals and Phalanges)**

| Injury Site | Mechanism | Frequency |
|---|---|---|
| Forefoot # | Direct Blow | 11 |
| | Bending | 2 |
| Phalanges # | Crush/Direct Blow | 4 |
| | Axial Load + Varus or Valgus | 2 |

From this analysis, it would appear that the most important mechanism of loading of the foot and ankle is from forces whose principal vector is along the axis of the tibia. Although injuries caused by rotations of the foot and ankle account for a large number of injuries, these are not likely to cause the more severe injuries seen to front seat car occupants.

**Below-Knee Primary Injury Sources**

For each injury detailed above, a primary contact source was deduced from inspection of the vehicle and the type and mechanism of injury. This was defined as the Primary Injury Source. Table 12 details the different primary injury source for each injury type. Several categories have been amalgamated for ease of analysis (e.g. firewall and wheel-well intrusion are coded separately on the database but combined in Table 12.

Intrusion was correlated with the primary injury mechanism in almost half of injuries below the knee. Intrusion also played a significant role in the 19 cases where entrapment between floor and facia occurred. In these cases, it was judged that, because the leg had become trapped, as evident by contacts identified on the facia, the injury was not simply correlated with intrusion. The entrapment would be likely to increase the risk of injury or to increase the severity of injuries that may have occurred in the presence of intrusion alone.

**Table 12.**
**Primary Injury Source for Below-Knee Injuries**

| | A | B | C | D | E | F |
|---|---|---|---|---|---|---|
| Tibial Plateau # | 5 | | | | 3 | 3 |
| Tibial Shaft # | 8 | | | | 3 | 2 |
| Pilon # | 3 | 1 | | | 9 | |
| Malleolus # | | 2 | 8 | | 10 | 3 |
| Talus # | 2 | 2 | 1 | | 11 | |
| Calcaneus # | 1 | | | | 5 | |
| Midfoot # | | | | 1 | 5 | |
| Lisfranc's # | | 2 | | 1 | 2 | |
| Forefoot # | | 10 | | | 3 | |
| Phalanges # | | 2 | | 2 | 2 | |
| Total | 19 | 19 | 9 | 4 | 53 | 8 |

A = Entrapment between floor and facia
B = Foot-pedal contact
C = Rolled off pedal
D = Foot trapped under pedal
E = Intrusion of footwell
F = Floor contact

For pilon fractures, malleolus fractures, talus fractures, fractures of the calcaneus and midfoot region, intrusion was again correlated with primary injury mechanism. Intrusion was also implicated in fractures of the tibial shaft and tibial plateau but in addition, there was evidence of the knee becoming trapped by the facia in over 50 percent of these fractures.

Injuries attributable to contact with a foot pedal or the foot rolling off a pedal accounted for 25 percent of below knee injuries. The 'foot roll-off pedal' injuries were almost exclusively ankle-malleolus fractures and this mechanism accounted for 8 of the 23 (34%) malleolar fractures. It was concluded that for these injuries, the foot was on the pedal (usually the brake pedal) at the time of impact and then rolled off one side of the pedal due to the crash pulse, leading to ankle fractures..

The majority of forefoot injuries attributed to foot pedal contact were fractures of the distal metatarsals. From careful consideration of the vehicle inspection notes and accident circumstances, it was concluded that the foot was on the pedal at the time of impact and that the fractures seen in these cases were consistent with concentrated loading over a small area of the foot (e.g. fracture of two metatarsals only).

**Injury Severity and Impairment**

When only the frequency of the different below knee injuries is considered, fractures of the ankle (malleolar and pilon fracture types) are the found to be the most important fractures to prevent, followed by fractures of the talus (all types), forefoot and tibia (shaft and plateau). In order to prioritise the injuries, the severity of each injury and its potential for causing long term disability needs to be evaluated. This analysis was performed using the injury severity and impairment scales defined by the American Orthopaedic Foot and Ankle Society Trauma Committee (AOFAS). Table 13 shows the AOFAS injury severity score by anatomical site for all below knee injuries.

**Table 13.**
**AOFAS Injury Severity Score by Site for Below Knee AIS 2+ Injuries**

| Injury Site | Injury Severity Score (AOFAS Scale) | | | | | |
|---|---|---|---|---|---|---|
| | 1 | 2 | 3 | 4 | 5 | 6 |
| Tibial Shaft # | | | | 4 | 7 | 2 |
| Malleolus # | 10 | 5 | 5 | 3 | | |
| Pilon # | | | 1 | 5 | | 7 |
| Talus # | 1 | 1 | 8 | 1 | 4 | 1 |
| Calcaneus # | | | 1 | | 5 | |
| Mid-foot # | | | 6 | | | |
| Lisfranc's # | | | | | 5 | |
| Forefoot # | | 2 | 7 | 4 | | |
| Phalanges # | | 2 | 4 | | | |
| Totals | 11 | 10 | 32 | 17 | 21 | 10 |

Of the most severe below knee injuries (i.e. those that are most difficult to treat), the most important injuries in descending rank were:

1 Pilon fractures
2 Tibial shaft fractures
3 Fractures of the Calcaneus
4 Fractures of the Talus (Talar Neck fractures)
5 Lisfranc's Joint injuries

However, this analysis fails to take into account the likely outcome from such injuries. An analysis can be carried out in which the impairment scores, as determined by the AOFAS scale, can examined by injury site (Table 14).

**Table 14.**
**AOFAS Injury Impairment Score by Site for Below Knee AIS 2+ Injuries**

| Injury Site | Impairment Score (AOFAS Scale) | | | | | |
|---|---|---|---|---|---|---|
| | **0** | **1** | **2** | **3** | **4** | **5** |
| Tibial Shaft # | | 3 | 8 | 2 | | |
| Malleolus # | 10 | 6 | 7 | | | |
| Pilon # | | | 6 | | 7 | |
| Talus # | | 8 | 2 | 5 | 1 | |
| Calcaneus # | | | | 5 | | 1 |
| Mid-foot # | | 2 | 2 | 2 | | |
| Lisfranc's # | | | | 5 | | |
| Forefoot # | | 8 | 5 | | | |
| Phalanges # | | 6 | | | | |

If the likely long-term impairment from each injury is considered then the most important injuries become:

1. Pilon Fractures
2. Fractures of the Talar Neck
3. Fractures of the Calcaneus
4. Lisfranc's Joint Injuries
5. Fractures of the Tibia
6. Malleolar Fractures

### Comparison of Driver and Passenger Injuries

There were 93 individual injuries to drivers and 19 injuries to front seat passengers. With such a small sample of front seat passenger injuries, it is difficult to draw any firm conclusions. An interesting point to note is that for both talus fractures and pilon fractures, the driver's right leg is seen to be at risk of injury (although in neither case was this relationship seen to be statistically significant). This may reflect that intrusion of the foot-well region is usually highest on the vehicle outboard because of the locality of the wheel-well. However, it should also be remembered that the right leg is used for braking.

Apart from these general observations, there is insufficient data to allow any firm conclusions to be drawn regarding the mechanisms of injury to front seat passenger legs, feet and ankles. The spectrum of injuries is broad with only small numbers in each category of primary injury source and so it was not possible to make any further observations.

### Accident Analysis Discussion

The primary focus of future impact biomechanics experiments should be:

1. Pilon fractures
2. Fractures of the Calcaneus
3. Fractures of the Neck of the Talus

For these three injuries, the principal mechanism of is axial loading. Injuries related to foot pedal roll-off were common but would not be expected to cause significant long-term disability or impairment. The application of a protection criterion, relating to this phenomenon, would be somewhat difficult in full-scale impact testing due to the unpredictable nature of the roll-off process.

It was concluded that intrusion was the most important method of loading the tibia, foot and ankle. However, it was not possible to demonstrate the relationship between increasing levels of intrusion and increasing risk of severe and impairing injuries.

It is recommended that the loading mechanisms in the laboratory-based experiments should be a simplification of the loading mechanisms that occur in 'real world' car crashes in which occupants sustain lower limb injuries, in order that the mechanisms can be studied in a more defined manner.

## SUMMARY

In the driving simulator trials, the mean brake pedal force in an emergency event was 630N. At the same time a mean plantar flexion of the foot of 15° was recorded. Half of the drivers lifted their heel from the floor pan during the braking action.

In live humans, it is known that the braking force will be generated by a combination of extension at the knee joint and tension applied through all the plantar flexing muscles (Achilles, soleus, flexor hallucis longus etc). In the biomechanical tests reported here, it was only possible to use the Achilles tendon to regenerate the plantar flexing force. Anatomical measurements, made during the study, were used and a mean required force in the Achilles tendon of 1.5kN was calculated. The results reported here and previous reports [6-8], have shown that the application of such a force will significantly enhance peak axial loading.

The accident analysis, summarised at the end of this paper and previous reports [18, 21], have indicated the importance of axial loading in the three most disabling injuries at the ankle (pilon, calcaneal and talar neck). It seems probable that, while bracing hard may help an occupant remain in the seat during an impact, it will at the same time significantly increase the risk of a disabling lower leg injury.

It is not known at what point foot and ankle injuries occur during an impact. If drivers are usually braking prior to the accident, as suggested by previous work, [18] and the simulator trial, it would appear desirable to conduct biomechanical tests with the foot in an initial plantar flexed position with a loading force applied. The indicated 1.5kN of active lower leg muscle force calculated here is larger than the passive forces used in previous biomechanical studies (450N [22] 960N [6]). This figure is a good indicator of the forces that should be considered for future biomechanical studies to reproduce a realistic situation. It also seems essential that biomechanical evaluations of fracture mechanisms should encompass both initial heel floor contact and no contact between the heel and the floor in the experimental set up.

Currently the Hybrid III is used almost universally for legislative car crash testing, and the test procedure specifies that the dummy foot should be placed on the vehicle accelerator at the start of the test. This study indicated that, in frontal collisions, most drivers were braking at the time of the impact. The position of the feet in future car crash testing needs careful consideration, if the next generation of dummies are to be used to accurately measure the risk of lower leg injury.

In a similar way the application of a braking force needs careful consideration, although there would clearly be difficulties in achieving this controllably under crash test conditions. The proposed design for ALEx II incorporates a passive Achilles tension that will apply a consistent force to the ankle joint without a large increase in bending moment at the end of the tibia. This will allow more biofidelic movement at the ankle joint and at the same time, a partial addition to axial loading. A higher, active force would have to be applied to the simulated tendon if plantar flexing forces, such as those recorded in this work, were to be generated.

The biofidelity tests carried out for this work have examined the response of the foot and ankle to sub-injury simple pendulum impact tests, based around the EEVC foot certification procedure. Tests were performed on both PMHS specimens as well as live human volunteers. For live volunteers the foot response with the lower leg muscles both tensed and relaxed was tested. The response of the foot to both heel and toe impacts was studied. For the PMHS specimens and for the dummy, attempts to assess the dynamic response of the foot in inversion and eversion were made. An attempt was also made to normalise the results of both the PMHS specimen tests and the volunteer tests in order to align the results with that of the fiftieth percentile. The normalisation procedure reduced the scatter for some of the test results, but was not sufficiently well developed nor validated at this stage to allow a comparison of normalised results to be made. The reason that this failed was probably due to the fact that it was not possible to measure all potentially influential factors prior to testing. For future work, extra measurements will be made to improve the normalisation procedure e.g. ankle joint stiffness. The results presented are thus the unnormalised data from the PMHS specimens and volunteers compared with the dummy results.

Two designs dummy feet and ankle were tested, the Hybrid III with the 'soft-stop' 45° ankle and the GM/FTSS foot, attached to the Hybrid III leg. The GM/FTSS foot and ankle responses were closer to those of the PMHS specimens and human volunteers (where measured) for the mean peak dorsiflexion angle, pendulum acceleration and tibial bending moment than those obtained for the Hybrid III. Conversely for the mean peak tibia force, the Hybrid III results were closer to those of the PMHS specimens than the GM/FTSS foot, although the difference were statistically significant for both. The tests using heel impacts have shown that the data recorded using the GM/FTSS foot is very similar to the data obtained with PMHS specimens. The data obtained with the Hybrid III foot differs significantly. Neither dummy foot appears to be superior, regarding the biofidelity to inversion or eversion. Both generate a higher acceleration response on the pendulum but result in a lower peak tibial axial force when compared with the PMHS test results.

Most people involved in frontal impacts will be aware of the impending accident. Therefore, it could be considered that the most appropriate condition that the dummy response should simulate, would be that of the aware volunteers.

Due to the reduction in fatalities of occupants in car accidents, injuries to the lower leg are being increasingly recognised as a source of both severe and long term impairing injuries. Dummy design has been aimed at addressing injury assessment for body regions where there is a high risk of fatality if injury occurs. The shift in emphasis to other body regions calls for the refinement in both the design and injury criteria of the current dummy used for assessment of vehicle performance in frontal impact, the Hybrid III.

The accident analysis of lower leg injuries in the UK population showed that, if frequency alone is considered, the ankle is the most common fracture site (32%), followed by fractures of the talus (14%), forefoot and tibia (both 12%) and tibial plateau (10%). However, if severity and impairment are taken into consideration this ranking changes and the order for prevention is pilon, calcaneal and talar neck fractures. The primary source of injury was attributed to intrusion of the footwell.

**Future Work** - The next phase of this work will focus on recreating the lower leg injuries, seen in car accidents in the laboratory. The aim will be to recreate the loading patterns and intrusion seen in the foot well and use cadaver legs and dummy legs to compare the results. This research should lead to a clearer understanding of the mechanism of injury and aid the development of future dummies such that new injury criteria can be developed.

## CONCLUSIONS

1. In simulated frontal impacts, drivers were found to apply a mean peak force of 630N to the brake pedal with the ball of the foot.

2. The equivalent force that would be needed in the Achilles tendon to generate this applied braking force was calculated to be 1.5kN

3. In just under half of the emergency braking events, the subject's heel was not in contact with the floor pan at the point of applied peak brake force.

4. The mean ankle articulation at the time of peak brake force application was found to be 15° of plantar flexion

5. Dynamic responses of PMHS and volunteers' lower legs to sub-injury pendulum foot impacts have been obtained to provide biomechanical reference data for dummy response evaluation.

6. A technique has been developed to generate forces of up to 3.3kN in the Achilles tendon of the PMHS prior to impact.

7. The application of Achilles tendon force significantly increases the tibia axial force and decreases the dorsiflexion angular response to impacts to the PMHS toe area.

8. For the aware volunteers, the dorsiflexion angular displacement was reduced and the impactor response increased in comparison with unaware volunteers.

9. Neither of the two dummy feet and ankles evaluated gave the same responses as the PMHS or volunteers for all the parameters measured. The GM/FTSS foot was closest for most conditions with good biofidelity for heel impacts. The Hybrid III foot was nearer to the PMHS for tibia axial force for toe impacts but the differences were significant.

10. The Hybrid III dummy leg would have to be redesigned to align the axis of the tibia with the ankle and knee joints if an artificial generation of tibia bending moments from an applied tibia compressive force is to be avoided.

11. The three most important lower leg injuries found in accidents, taking into account frequency, severity and outcome, are pilon, calcaneal and talar neck fractures.

12. The most likely cause of these most important lower leg injuries is considered to be axial force.

13. For future legislative crash testing, the initial position of the dummy lower limb and the representation of active muscle tension in the lower leg should be considered, particularly if the risk of lower leg injury is to be assessed in a biofidelic manner.


## ACKNOWLEDGEMENTS

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## APPENDIX 1 – TIBIAL LOADCELL INSERTION

The method of insertion of the tibial load cell was developed specifically for this project, a 12x5cm area of skin and superficial tissue were removed from the anterior surface of the leg at a point defined by the subcutaneous surface of the tibia 6cm above the ankle joint. The area of tibia thus exposed was then stripped of the periosteum by a periosteal elevator. The 12-cm section of the tibia was then freed from all surrounding tissues by gentle sharp dissection around its circumference. Particular attention was paid to preserving the soft tissue integrity around the posterior aspect of the tibia. The fibula shaft was vulnerable to fracture and great care was taken to preserve its integrity. The intra-osseous membrane was only divided for the 12cms and care is taken to preserve the structure both proximally and distally at the syndesmosis.

The PMHS specimen was held with the leg in its anatomical position such that the foot rests at 90° to the tibial shaft with the 2nd toe pointing vertically upwards. The two posterior end sections of the drilling jig (Figure A1) were passed around the tibia along with securing ties. The anterior section of the jig was positioned to join with the two posterior sections and secured. The whole drilling jig was then manipulated and adjusted around the tibia using four adjusting screws to ensure that the orientation was correct in all planes, both radially and axially. The drilling jig was marked with the central longitudinal axis, to assist with this task. Locating holes were drilled through the tibia through which four fixing pins were placed, two at right angles to each other at each end (Figure A2). Care was exercised when drilling through the tibia to minimise the risk of mal-alignment, as the hole was drilled obliquely into the surface of the tibia. The tibia was also marked for cutting through the jig. The tibia was divided and the removed section kept for further physical property analysis. A fine cutting ring was then placed over the ends of the tibia and fixing pins reinserted to maintain its position. The sawn end of the tibia was then ground down to exactly the right length (Figure A3).

The cut tibial ends were prepared, over a 25mm length from the cut edge, with a degreasing agent. The potting cups were then placed on the tibial ends and realigned with the fixing pins to maintain their position and a dummy tibial load cell was inserted to maintain the alignment of the leg (and stop any relative movement due to dehydration or degradation) (Figure A4). Potting media is pressure injected into the cups via two 6mm holes (Figures A5) drills to maintain position. The completed tibial load cell assembly is shown in Figure A6.

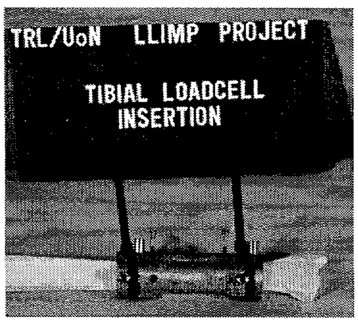


**Figure A1 - Cutting Jig In-Situ on Tibia**

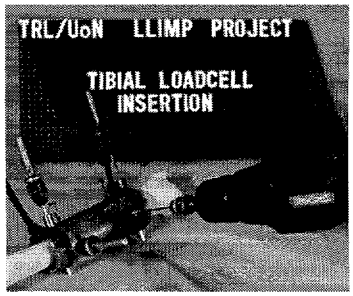


**Figure A2 - Guide Wire Drilling**

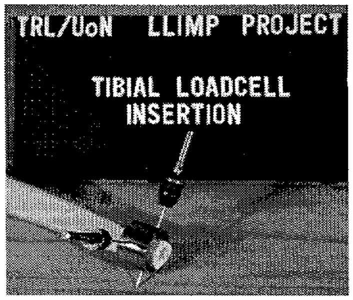


**Figure A3 - Cutting Jig and Guide Wires**

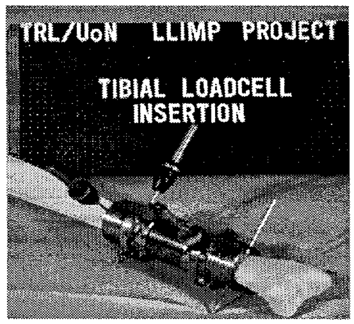


**Figure A4 - Dummy Load Cell In-Situ**

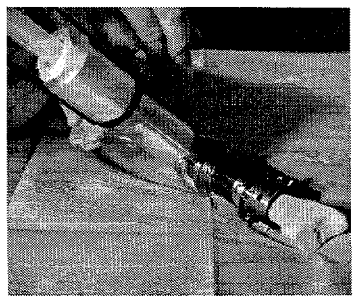

**Figure A5 - Pressure Injection**

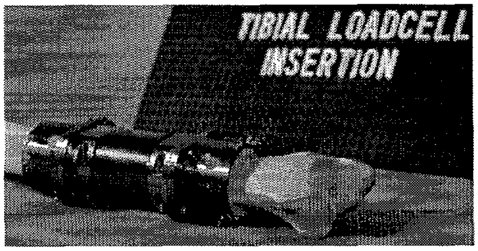


**Figure A6 - Completed Assembly**

Acoustic Transducers - It was hypothesised that a failure, whether bone fracture or ligament rupture could produce an acoustic 'event' i.e. a high frequency signature. These data could be used to determine the environment and forces that exist at the time of any defined injury. For the purpose of these tests a simple 'acoustic' guitar pick up, one located on the surface of the second metatarsal and the other on the tibia was used to measure any acoustic signals which might be generated due to an injury event. In the tests reported in this paper no failures were detected

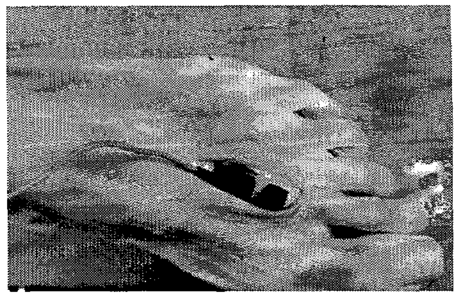

**Figure A7 - Acoustic Transducer on Second Metatarsal**

## APPENDIX 2 – PEAK RESPONSES FOR INDIVIDUAL TESTS FOR PMHS, DUMMIES AND VOLUNTEERS

| | Peak Pendulum Acceleration (g) | | | | |
|---|---|---|---|---|---|
| Specimen | Heel Impact 4m/sec | Toe Impact 4m/sec | Toe impact 6m/sec | Eversion Impact 4m/sec | Inversion Impact 4m/sec |
| TRL001R | Not tested | 21.3 | 43.75 | Not tested | Not tested |
| TRL001L | 132.78 | 21.7 | 41.37 | 28.17 | 21.59 |
| TRL004R | 117.83 | 34.82 | 62.25 | 31.51 | 26.79 |
| TRL004L | 146.28 | 33.94 | 51.01 | 25.89 | 35.95 |
| TRL007R | 130.74 | 27.96 | 46.57 | 52.57 | 31.71 |
| TRL007L | 122.11 | 26.41 | 44.91 | 32.02 | 21.17 |
| TRL005L | 146.22 | 26.92 | 53.04 | 29.77 | 27.38 |
| *TRL005R | 140.09 | 22.88 | 41.48 | 33.14 | 26.89 |
| GM/FTSS – Test 1 | 132.56 | 51 | 87.69 | 58.02 | 39.86 |
| GM/FTSS – Test 2 | 133.77 | 54.37 | 86.2 | 36.43 | 42.56 |
| GM/FTSS – Test 3 | 127.91 | 54.46 | 87.96 | 53.36 | 41.34 |
| Hybrid III – Test 1 | 213.29 | 86.06 | 139.41 | 43.15 | 54.87 |
| Hybrid III – Test 2 | 206.43 | 85.36 | 141.11 | 47.59 | 55.08 |
| Hybrid III – Test 3 | 201.18 | No data | 136.13 | 38.64 | 53.43 |
| Volunteer 1 - Unaware | | 26.05 | 47.42 | | |
| Volunteer 2 - Unaware | | 32.64 | 76.19 | | |
| Volunteer 3- Unaware | | 23.96 | 64.35 | | |
| Volunteer 4 - Unaware | | 46.81 | 61.97 | | |
| Volunteer 5 - Unaware | Not tested | 36.83 | 78.48 | Not tested | Not tested |
| Volunteer 1 - Tensed | | No data | 57.64 | | |
| Volunteer 2 - Tensed | | 29.17 | 94.97 | | |
| Volunteer 3 - Tensed | | 29.49 | No data | | |
| Volunteer 4 - Tensed | | 56.03 | 95.69 | | |
| Volunteer 5 - Tensed | | 40.46 | 88.44 | | |

*Above knee Amputation

| | Peak Tibial Force Fz (N) | | | | |
|---|---|---|---|---|---|
| **Specimen** | **Heel Impact 4m/sec** | **Toe Impact 4m/sec** | **Toe impact 6m/sec** | **Eversion Impact 4m/sec** | **Inversion Impact 4m/sec** |
| TRL001R | Not tested | -861 | -1676 | Not tested | Not tested |
| TRL001L | -2329 | -1112 | -1839 | -764 | -265 |
| TRL004R | -2179 | -532 | -1181 | -832 | -404 |
| TRL004L | -2640 | -600 | -1109 | -455 | -391 |
| TRL007R | -1834 | -428 | -1003 | -916 | -335 |
| TRL007L | -2047 | -541 | -1009 | -326 | -331 |
| TRL005L | -2199 | -709 | -1304 | -651 | -577 |
| *TRL005R | -2219 | -851 | -1575 | -543 | -645 |
| GM/FTSS – Test 1 | -1861 | -153 | -198 | -284 | -293 |
| GM/FTSS – Test 2 | -1894 | -158 | -189 | -221 | -308 |
| GM/FTSS – Test 3 | -1819 | -182 | -233 | -323 | -255 |
| Hybrid III – Test 1 | -2999 | -495 | -860 | -328 | -256 |
| Hybrid III – Test 2 | -3094 | -572 | -821 | -361 | -246 |
| Hybrid III – Test 3 | -2878 | No data | -785 | -269 | -258 |

*** Above knee Amputation**

| | Peak Tibia Bending Moment, My (Nm) | | Peak Tibia Bending Moment, Mx (Nm) | |
|---|---|---|---|---|
| **Specimen** | **Toe Impact 4m/sec** | **Toe Impact 6m/sec** | **Eversion Impact 4m/sec** | **Inversion Impact 4m/sec** |
| TRL001R | 5.86 | 9.85 | Not tested | Not tested |
| TRL001L | 6.95 | -19.2 | -9.91 | -7.71 |
| TRL004R | 9.44 | 14.83 | -15.86 | 6.08 |
| TRL004L | -7.23 | -14.27 | 6.7 | 6.08 |
| TRL007R | -6.3 | -17.95 | -26.2 | 5.23 |
| TRL007L | -9.43 | -20.57 | -8.38 | -5.39 |
| TRL005L | -11.9 | -21.99 | 13.01 | 3.83 |
| *TRL005R | 5.82 | -10.74 | -5.91 | -9.45 |
| GM/FTSS – Test 1 | 19.14 | 39.22 | -7.07 | 6.07 |
| GM/FTSS – Test 2 | 19.98 | 38.19 | -9.92 | -4.68 |
| GM/FTSS – Test 3 | 17.99 | 39.35 | 8.3 | -4.7 |
| Hybrid III – Test 1 | 51.11 | 115.85 | 10.81 | -9.25 |
| Hybrid III – Test 2 | 51.73 | 114.01 | -8.8 | -8.54 |
| Hybrid III – Test 3 | No data | 111.79 | 10.79 | -7.98 |

*** Above knee Amputation**