MECHANISMS OF INJURY AND INJURY CRITERIA FOR THE HUMAN FOOT AND ANKLE IN DYNAMIC AXIAL IMPACTS TO THE FOOT

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ABSTRACT

A series of 50 amputated human lower limbs were tested to determine the injury tolerance of the ankle and foot structure to kinematic and kinetic parameters that could be used to describe the impact environment. The test apparatus consisted of a pendulum-driven plate constrained to move longitudinally while simulating the motion of the toepan structure in an automobile. The leg specimens were amputated at the midshaft of the femur and attached to a device simulating the hip joint. The legs were mounted to the plate in a position approximating the geometry typical for drivers and were constrained with a spring loaded tether and knee harness which simulated the action of the leg muscles. Pendulum speed, foot plate acceleration and loads, leg and foot angular velocities and accelerations, and tibia internal loads were measured directly for each test. The lower limb trauma included fractures of the calcaneus, talus, malleoli, and ligamentous tears identified from radiography and detailed necropsy. Logistic regression analysis was used to interpret the data. The peak plantar contact load and its rate of onset, as well as resultant heel acceleration were good predictors of injury ($p<0.05$). The fifty percent probability of injury level using these single factor models is at 9.3 kN peak contact force, 5 kN/msec peak contact force onset rate, and 216 G's peak resultant heel acceleration. Initial position of the foot was found to influence injury outcome with the dorsiflexed foot being more resistant to injury than the neutral or plantarflexed foot. Finally, injury risk models using a linear combination of contact force or its onset rate, initial flexion angle, and resultant ankle angular velocity were found to be very good predictors of injury ($p=0.0001$).

HIGH ENERGY TRAUMA to the ankle and foot region is a debilitating and expensive consequence of motor vehicle accidents. Morgan et al. (1991) estimated their incidence at 10% of all AIS≥2 injuries. Pattimore et al. (1991) reviewed the UK CCIS accident database and noted that 68% of all skeletal injuries to front seat occupants in frontal impacts occur below the knee. These injuries have a maximum possible AIS rating of 2

(AIS, 1990) defined as a moderate injury, unless exsanguination or uncontrolled sepsis occurs. However, excepting neurological cases, lower extremity injuries tend to drive the costs of subsequent outpatient costs due to the long-term disability and impairment associated with these injuries (Crandall et al., 1994).

Footwell intrusion and interaction with pedal controls appear to be primary sources of lower limb injuries. States (1986) identified leg entrapment followed by intrusion as a likely mechanism. Morgan et al. (1991) reviewed accident data and postulated mechanisms to explain the injuries. They found that contact with the foot pedals or the floor were the main sources of injuries. Pattimore et al. (1991) also surveyed accident data and found footwell contact to be the main cause of foot and ankle trauma. Lestina et al. (1992) reviewed accident data in Virginia and noted that intrusion was associated with most of the injuries. They also systematically reviewed the patient radiographs and postulated that inversion or eversion played a role in 65% of foot and ankle injuries and 92% of malleolar injuries.

Portier et al. (1993) noted a correlation between the level of intrusion and injury. However, this result was confounded by the strong correlation between the level of intrusion and the vehicle delta-V. Thomas et al. (1995) used UK crash data and found that the risk of lower extremity injury correlated better with the level of footwell intrusion than with vehicle delta-V. However, he noted that 30% of foot and ankle fractures occurred without measurable footwell intrusion.

Dischinger et al. (1994a, 1994b) used epidemiological studies to determine that footwell intrusion was not a requirement for severe foot and ankle injury. They noted that calcaneus and pilon fractures often occurred with minimal intrusion levels. Crandall et al. (1995, 1996) reviewed accident investigations, crash tests, and computer simulations. They found that injury correlated with the vehicle delta-V, the timing of the intrusion event, the level of intrusion, and the interaction with pedal controls.

In addition to the epidemiological studies, controlled laboratory tests have been conducted to determine injury mechanisms and thresholds. Different protocols have been used which can be categorized by the location of the excision to isolate the foot and ankle specimens. In this context, above-the-knee means excision at mid-thigh where the knee and leg musculature are intact and functional, at-the-knee means excision at the knee joint capsule where the femur is removed and some leg musculature is detached, and below-the-knee means excision at mid-shank where the ankle and distal tibia-fibula syndesmosis are intact.

Begeman (1990, 1993a, 1993b, 1994) conducted a series of tests designed to produce dynamic rotations about isolated anatomical axes of the ankle using impulsive loads to the foot. Below-the-knee leg specimens were used where the proximal tibia/fibula were potted to a load cell. The injuries obtained were ligament tears and avulsion fractures which suggested the injury thresholds of 45 degrees forced dorsiflexion (Begeman and Prasad, 1990), 60 degrees forced inversion or eversion (Begeman et al., 1993a), and 50 degrees forced internal or external rotation (Begeman et al., 1994) of the ankle. No correlations with peak loads, moments, or loading rates were observed.

Roberts et al. (1992) subjected above-the-knee specimens to compressive loads using a constant velocity device. They found that the threshold level for dynamic loads was approximately twice that for static loads. Yoganandan et al. (1996) applied dynamic loads to the plantar surface of the foot of at-the-knee specimens using a pendulum device. Fractures to the calcaneus and distal tibia were observed. They combined this work with the previous work of Begeman and Roberts to obtain an axial load injury

threshold of 6.7 kN measured at the mid-shank of the leg.

Paranteau et al. (1995) used below-the-knee specimens which were excised just above the distal tibia-fibula syndesmosis. Average quasi-static failure loads at the ankle were determined to be 33 Nm in dorsiflexon, 34 Nm in inversion, and 48 Nm in eversion with average joint rotations of 44 degrees in dorsiflexion, 34 degrees in inversion, and 32 degrees in eversion. However, these results did not consider the effect of the leg musculature applied through the Achilles tendon. Petit et al. (1996) used the same apparatus but included the effects of the Achilles tendon by applying a constant traction to the tendon. They obtained average ankle failure loads of 47 Nm in dorsiflexion, 40 Nm in inversion, and 35 Nm in eversion, with average rotations of 49 degrees in dorsiflexion, 34 degrees in inversion, and 32 degrees in eversion.

Crandall et al. (1996) reported results from quasi-static ankle rotation tests using volunteers and cadavers. Moment response curves at the ankle were obtained for a range of knee flexion angles. The location of the excision was found to be important where legs excised below the knee had significantly different responses than those excised above the knee. Knee flexion angle also had a significant effect. Leg musculature was determined to have a significant effect on the moment response and range of motion of the ankle. The ankle extensors were particularly significant and they concluded that biofidelity of response depended on the extent to which the leg musculature was still intact and the location of the excision.

Concurrent with the laboratory work, the test dummies (Hybrid III) have been upgraded to improve their biofidelity, based on the tests of Begeman and Prasad (1990). The Hybrid III ankle range of motion in dorsiflexion has been increased to 45 degrees. Using the Hybrid III lower limb, Mertz (1993) proposed an injury criterion for combined loads to the lower leg based on the static strength of the tibia. He used a cumulative damage model normalized to 225 Nm for bending moment and 36 kN for axial loads below 8 kN. This criterion, known as the Tibia Index, is intended for use on the data obtained from the leg load cells installed on the Hybrid III test dummy.

There is no general agreement on the mechanisms of injury for the foot and ankle complex due to the use of significantly different test protocols and to the fact that there are many different injury mechanisms of the foot and ankle. In particular, the differences arise from:

1. Excision site for the specimen: Some protocols have excised the specimens below the knee and so detached the leg musculature which has been found to significantly affect ankle response. Furthermore, the joints of the foot and knee are considered to act in concert to protect the ankle joint.
2. Boundary conditions: Some protocols have attached the feet to the test fixture with screws or straps. This would prevent the arching of the foot which influences ankle motion considerably. Other protocols have fixed the mid-shank to a load cell mounted to a rigid fixture. This prevents relative motion at the proximal tibia-fibular joint which affects the dorsiflexion motion at the ankle.
3. Imposed motions: Some protocols imposed rotations about one of the anatomic axes of the ankle. Inman (1991) and others have shown that these anatomic axes are only for description and do not correspond to the kinematic axes of the ankle.
4. Different injury mechanisms: The injury response for the ankle is likely a combination of mechanical risk factors. The allowable levels for individual factors depends on the concurrent levels of the remaining factors.

In an attempt to incorporate biofidelic boundary conditions and anatomically intact

foot and ankle specimens, a test protocol, custom instrumentation, and test apparatus (Figure 1) were developed at the University of Virginia. This paper presents a series of cadaver tests conducted with this apparatus in developing injury criteria of the foot and ankle complex.

METHODOLOGY

Human lower leg specimens were obtained from medical cadavers pursuant to the ethical guidelines and research protocol approved by the Human Usage Review Panel, National Highway Traffic Safety Administration and an institutional review board. Each specimen was obtained by excision at the midshaft of the femur. All knee ligaments and leg musculature were intact and knee and proximal tibia-fibula joint motions were normal. An initial series of radiographs was used to determine evidence of existing pathologies that would influence the outcome of the tests. Suitable fresh specimens were at -70° C and held in this condition for up to six months.

Prior to testing, the specimens were thawed at room temperature for 36 to 48 hours. A segment of the tibia diaphysis, approximately 9 cm long, was removed and an in situ five axis load cell was installed (torsion was not measured). Concurrently, a 3 cm incision was made over the medial aspect of the calcaneus and a mounting plate was screwed into the medial cortex of the calcaneus or through both cortices in later tests. This plate was used to attach a triaxial angular rate sensor and accelerometer assembly. A similar installation was performed on the tibia.

The test apparatus consisted of a pendulum-driven ankle test cell (Figure 1). The device was intended to simulate the interaction of the intruding floor pan with the lower extremity. The ankle test cell consisted of a contact pad, transfer piston, positioning brackets, and a foot plate which was constrained to move longitudinally. This pulse shape was controlled by varying the drop height of the pendulum, its effective mass, and the composition of the contact pad which acted as a pulse moderator. The positioning brackets allowed initial positioning of the foot and ankle complex about the flexion and inversion/eversion axes. The foot plate was instrumented with a five axis load cell and a triaxial accelerometer assembly.

A femur bar was attached to the femur to reproduce the original hip to knee length of the specimen. The locations of the ankle and knee centers were determined by palpation and marked on the skin. This was done to locate the photo targets and to improve repeatability for the measurements of joint locations during the tests. The leg was installed in the test fixture by placing the foot on the foot plate and attaching the femur bar to the hip clevis with a pin.

The hip center location was adjusted along with the angle of inclination of the foot plate to obtain the femur and tibia angles corresponding to the desired initial ankle flexion angle for the test. The initial inversion/eversion angle was obtained by rotating the foot plate about its length which corresponded to rotating the foot about its long axis. The sign convention assigned dorsiflexion and eversion as positive values while plantarflexion and inversion were assigned negative values. Neutral position for the foot was defined as that configuration where the normal to the plane of contact for the sole of the foot was parallel to the mechanical axis of the tibia.

Once the initial leg configuration was established, a harness was placed over the knee. This harness was attached to a spring via a block and tackle assembly which could be adjusted to provide an initial preload. Using the data obtained from the tibia load cell,

the preload was adjusted to give an increase in tibia axial load corresponding to the normal standing load (i.e. one half body weight). This preload was imposed to set the ankle and arch and obtain reasonable initial conformities for these joints.

Testing was conducted as per a factorial design based on the initial dorsiflexion and eversion angles, peak contact load, and the onset rate for the contact load. All electronic data were obtained using a DSP TRAQ-P data analysis system with a 3300 Hz 8 pole butterworth low pass prefilter for anti-aliasing. The data were then digitized at 10,000 samples per second and digitally filtered to SAE J211 channel class 180. A video record was also made of each test using a Kodak Ekta-Pro high speed (1000 fps) monochrome video camera and motion analysis workstation.

RESULTS AND DISCUSSION

TEST DATA - The contact force between the foot plate and the plantar surface of the foot was obtained by mass compensating the foot plate load cell data. The onset rate for the contact force was calculated as the average slope between the points corresponding to 10% and 90% of the total rise. The angular velocity data were integrated to obtain angular displacement. Ankle rotations were obtained by transforming the rotation of the foot into the reference frame of the leg using Euler angle transformations (Hall, 1996). The ankle angular velocity was obtained by differentiating the ankle rotations. Resultant heel acceleration was computed from the three axis array of accelerometers attached to the heel.

The fifty pendulum tests were classified into four categories based on the contact force and its onset rate. A representative test in each category is summarized in Table 1 and Figures 2 and 3. For these tests, peak contact loads (Figure 2), peak heel acceleration, peak axial tibia force, and peak ankle angular velocity occurred before 25 ms with ankle motion in dorsiflexion less than 15° (Figure 3). This motion was well below the dorsiflexion thresholds defined by Begeman (1990), Parenteau (1995), and Petit (1996). Xversion motion (not shown) was less than 5°, also well within the limit proposed by Begeman (1993).

Table 1. Tests of Different and Loads and Rate of Loading

	Low Rate of loading (<0.8 kN/msec)	High Rate of loading (> 1.0 kN/msec)
Low Force (<4900 N)	25G	26D
High Force (≥4900 N)	26F	25D

The test results and the injury outcome for each of these tests is presented in Appendix A and B. Among the fifty human subjects tested, one sustained a pilon fracture, five sustained calcaneus fractures, three sustained malleolar fractures, two sustained talar fractures, and two sustained ligament tears. A review of axial impact test data (Yoganandan, 1996) suggested that calcaneus fractures, common in automobile crashes (Crandall, 1995), can be attributed to axial loading of the foot. Ligament tears, malleolar fractures, and talar fractures observed in these tests are often attributed to large rotations of the ankle (Lestina, 1992). However, the ankle rotations were small in these tests while the corresponding rate of ankle rotation was high. These results suggest that foot and ankle injuries common in automobile crashes can occur due to a combination of

axial loading of the foot and high rate of ankle rotation and may not require large ankle rotations.

STATISTICAL ANALYSIS - Statistical analysis was conducted to develop lower extremity injury criteria based on measured physical parameters. Linear logistic regression (Hosmer and Lemeshow, 1989) was used to associate independent physical parameters with the observed presence of injury. The analyses were conducted using the SAS software package (SAS Institute, 1990).

The linear logistic model for the probability of injury has the form

$$p = \frac{1}{1+e^{-(\alpha+\sum_{1}^{n}\beta_i x_i)}}$$

where α is the intercept, x_i are the n independent variables used in the model and β_i are the corresponding coefficients associated with each independent variable. The independent variables used in the analyses were the measured forces, accelerations, and moment at the tibia and the ankle and computed parameters such as resultants of accelerations, forces, and angular velocities, rate of loading, impulse, and energy. The data were also tested for bias due to sex, age, bone mineral content, and bone size at the tibia mid-diaphysis. Although no bias was found, two older females (11D, 22D) were excluded based on significantly lower bone mineral content.

Table 2 presents the logistic regression models using peak contact force (*Fpltfz*), peak contact force onset rate (*Fprate*), and peak resultant heel acceleration, (*HeelAr*), which were all significant at the 95 percent confidence level. The table presents the p-value of the models, the α and β values of the model, and the value of the independent variable at twenty and fifty percent probability of injury. The p-value of a model refers to the probability of rejecting the null hypothesis that there is no association between the independent variable of the model and the injury outcome. At a 95 percent confidence level, only models with p value less than 0.05 were accepted. Figure 4 presents the risk curves along with the corresponding ±1 standard deviation using *Fpltfz* as the injury predictor function.

Table 2. Logistic Regression Models

Logistic Model	p-value	α	β	Value at X% probability of injury	
				X=20%	X=50%
Fpltfz (kN)	0.0196	-3.381±1.100	0.3633±0.1630	5.5 kN	9.3 kN
Fprate (kN/msec)	0.0126	-1.9886±0.5149	0.3988±0.1659	1.5 kN/msec	5.0 kN/msec
HeelAr (G's)	0.0222	-2.4401±0.8067	0.0113±0.0052	93 G's	216 G's

The peak axial force, *TibFz*, (p=0.6) and peak resultant moment, *TibMr*, (p=0.24) measured at the tibia load cell were not good predictors of injury. A possible explanation is that the measurements at the tibia load cell are confounded with the effects of parallel load paths (i.e., the fibula structure and leg musculature) and the inertial effects of the leg

between the ankle and the load cell.

The peak ankle rotations along the three anatomical axes were not good predictors of injury (dorsiflexion: p=0.4, eversion: p=0.56, internal rotation: p=0.61). As was noted earlier, the ankle and foot injuries were not associated with the amount of rotation but with the loading conditions and the rate of ankle rotation. The Tibia Index computed from the peak axial force and sagittal plane bending moment measured by the tibia load cell was also a poor predictor of injury (p=0.3).

Analysis of the data suggested that the resultant ankle angular velocity, *AnklVr*, and initial dorsiflexion angle, *DflexI*, also influence the injury outcome. Figure 5 presents a plot of peak contact force, *Fpltfz*, versus peak resultant ankle angular velocity, *AnklVr*. Each data point represents a test with initial flexion angle and the injury outcome noted. There are no foot and ankle injuries below *Fpltfz* of 4.2 kN in the 'non-failure zone' while there are only injuries in the 'failure zone'. There are both injury and non-injury cases in the 'intermediate zone'. All the non-injury cases for forces above 8 kN occur when the foot is initially dorsiflected ($10<DflexI<15$ degrees), while there are injury cases with the foot initially plantarflexed ($-15<DflexI<-10$) even at low values of contact force and resultant angular velocity.

The ankle resultant angular velocity maybe associated with the rate dependent response of the ligaments and soft tissue of the foot and ankle structure. Preliminary analysis of the data suggested that the resultant ankle angular velocity correlates with the velocity and energy of the intruding foot plate. A foot initially dorsiflexed is less likely to be injured than a foot initially neutral or plantarflexed since the contact area at the subtalar joint is greater and the joint is better packed to withstand greater loads in the dorsiflexed state than in the neutral or plantarflexed state. The initial Xversion angle, *EversI*, was found to have no influence on the injury outcome.

Since there was more than one parameter influencing injury outcome, multifactor logistic regression models were examined. A risk function using a linear combination of *Fpltfz* and *Fprate* could not be achieved because the two were highly correlated. Risk functions using a linear combination of *Fpltfz* or *Fprate* along with *DflexI* and *AnklVr* were found to be the best predictors of ankle and foot injuries (p=0.0001) as shown in Table 3. Figure 6 presents the probability of foot and ankle injuries versus the linear combination of *Fpltfz*, *AnklVr*, and *DflexI* along with the corresponding standard error in the probability. The linear combination separates the injury cases from the non-injury cases better and the standard error bounds are smaller than in the single factor models. Figure 7 presents iso-injury lines for 20% and 50% probability of injury for varying values of *AnklVr* and *Fpltfz*. The iso-injury lines are drawn for the foot in initial plantarflexion, neutral, and dorsiflexion. For *AnklVr*=50 rad/sec, the fifty percent probability of injury level is at 2 kN for the foot in initial plantarflexion, while it is at 11 kN for the foot in initial dorsiflexion.

CONCLUSIONS

Pendulum impact tests were conducted on 50 amputated human lower limbs to understand better the injury mechanisms and injury tolerances of the foot and ankle structure. The impacts were designed to represent the interaction of the lower extremities with the intruding floor pan during a vehicle crash. The injuries sustained by the subjects were representative of those noted in real world crashes. However, only a few severe foot and ankle injuries were generated.

Table 3. Multi-factor Regression Models

Logistic Model	p-value	α	β
Fpltfz (kN) *AnklVr* (rad/sec) *DflexI* (degrees)	0.0001	-9.7534±3.4812	0.4672±0.3149 0.1340±0.0465 -0.1436±0.0813
Fprate (kN/msec) *AnklVr* (rad/sec) *DflexI* (degrees)	0.0001	-11.3019±4.1977	1.0935±0.5473 0.1786±0.0675 -0.3278±0.1601

Foot contact force, contact force onset rate, and heel acceleration, were good predictors of injury with $p<0.05$. The fifty percent probability of injury level for the single factor models are at 9.3 kN of plantar contact force, 5 kN/msec of contact force onset rate, and 216 G's of resultant heel acceleration.

The resultant ankle angular velocity and the initial position of the foot had considerable influence on the injury outcome. Risk functions using a linear combination of the foot contact force or contact force onset rate, initial dorsiflexion angle, and the resultant ankle angular velocity were found to be the best predictors of injury ($p=0.0001$). The resultant ankle angular velocity maybe associated with the viscous response of the ligaments and soft tissue in the ankle. Preliminary analysis of the data suggested that the resultant ankle angular velocity correlates with the velocity and energy of the intruding foot plate. Under similar impact conditions, the initially dorsiflexed foot was less likely to be injured than the initially neutral or plantarflexed foot. This is because the ankle joint is better packed and has increased contact area for force transmission in the dorsiflexed state.

Initial eversion angle had no effect on injury outcome. Forces and moments measured at the tibia and peak ankle rotations were poor predictors of injury. The Tibia Index, as defined by Mertz (1993), was also a poor predictor of injury.

The results suggested that under dynamic loading conditions, as in a vehicle crash, foot and ankle injuries commonly attributed to large rotations of the ankle, can occur even when ankle rotations are small. Analsyis of the test data suggested that injury occurred early in the impact event when forces, accelerations, and moments were high, but the rotation of the foot was small.

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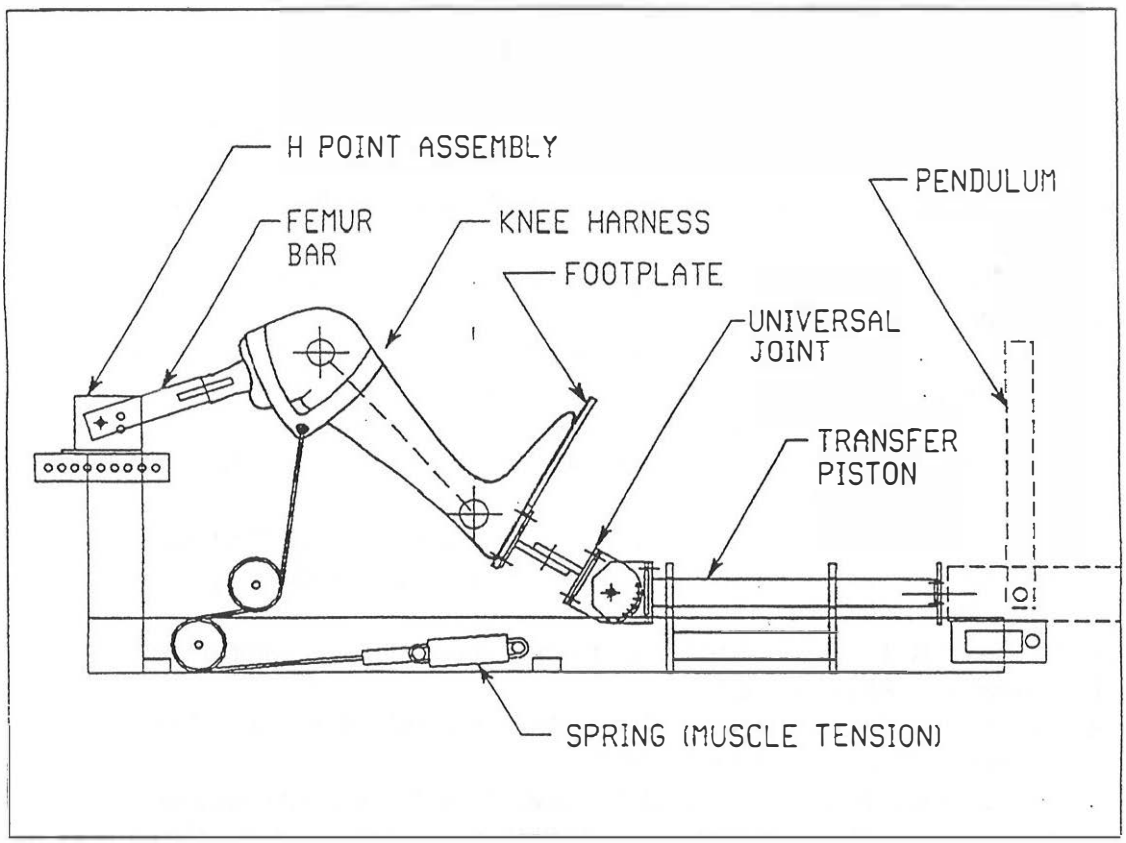

Figure 1. Schematic of test apparatus

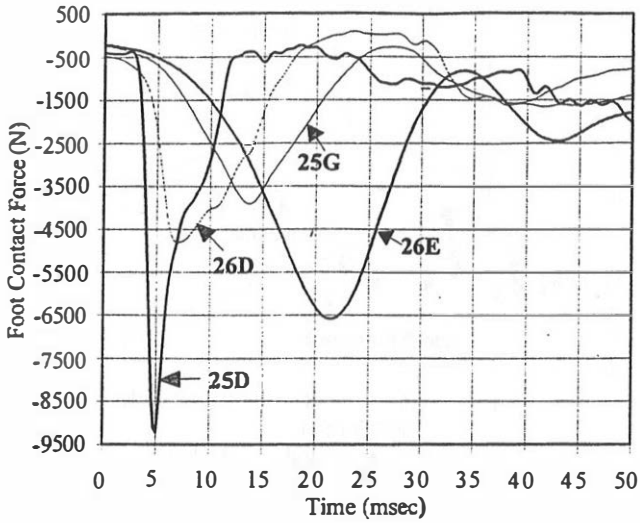

Figure 2. Contact force for representative tests.

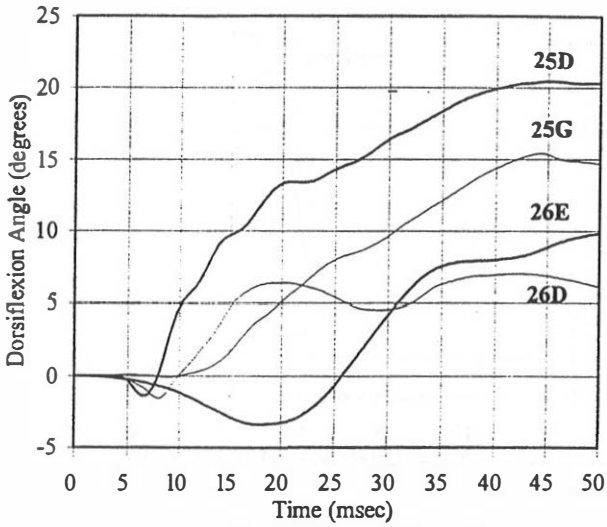

Figure 3. Ankle rotation in dorsiflexion for representative tests.

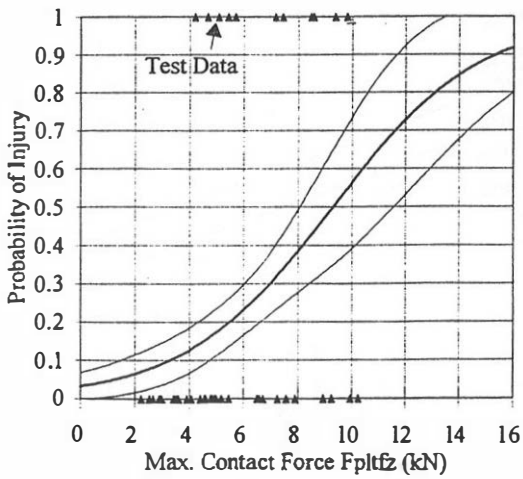

Figure 4. Probability of injury versus *Fpltfz*.

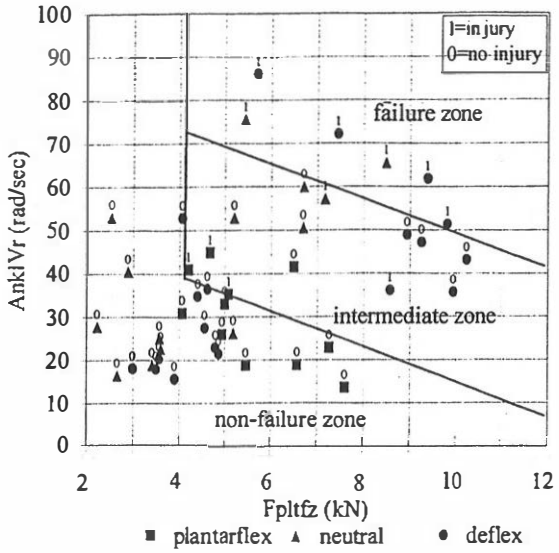

Figure 5. *AnklVr* versus *Fpltfz* for the test data

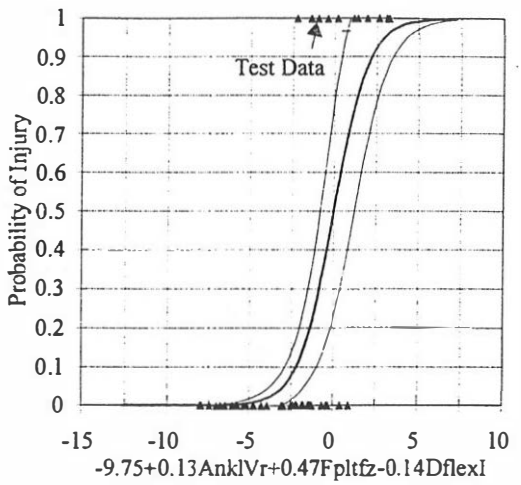

Figure 6. Probability of injury versus linear combination of *Fpltfz*, *DflexI*, and *AnklVr*.

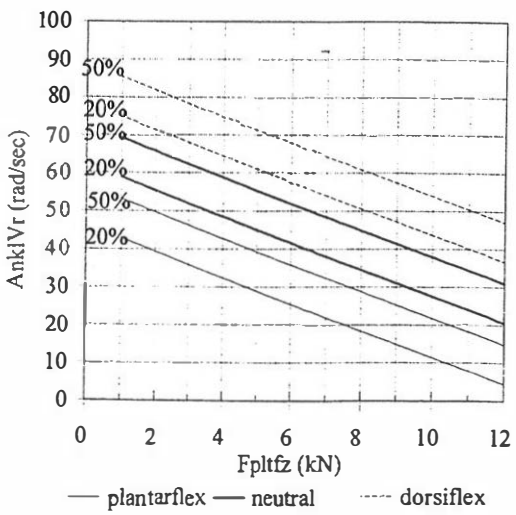

Figure 7. Iso-injury lines for different values of *AnklVr*, *Fpltfz*, and initial flexion angle.

Appendix A - Data Summary

Test	Age	Gender	Injury	*DflexI*	*EversI*	*Fpltfz*	*FpRate*	*Tibfz*	*TibMr*	*HeelAr*	*Vres*
			yes/no	deg	deg	*kN*	kN/ms	kN	Nm	g's	rad/s
11A	89	F	no	12.0	0.0	3.545	0.337	2.015	8.1		
11B	89	F	no	3.0	0.0	2.950	0.174	2.151	17.6		
11C		F	no	1.0	-10.0	2.237	0.131	1.783	22.0		27.8
11D		F	drop	0.0	-10.0	3.145	0.309	1.997	27.6		40.1
11E		F	no	0.0	10.0	2.532	0.235	1.620	22.8		53.0
11F		F	no	1.0	10.0	3.412	0.215	2.386	27.8		18.9
16A		M	no	4.0	0.0	3.566	0.243	2.420	12.9	55.2	25.1
16B		F	no	5.0	0.0	2.664	0.154	1.895	17.0		
19A	76	F	no	3.0	-10.0	2.658	0.160	1.862	10.6	32.1	16.5
19B	89	F	no	3.0	-10.0	5.206	0.453	4.111	33.1	49.8	52.9
20A	68	F	no	0.0	10.0	2.890	0.268	2.127	26.7	72.0	40.5
20B	68	M	no	-2.0	10.0	3.596	0.217	2.950	13.4	30.8	22.7
20C	72	F	no	12.0	0.0	4.562	0.413	3.225	38.8	82.1	27.5
20D	72	F	no	-11.0	0.0	6.499	0.772	5.414	50.4	88.8	41.6
20E	95	F	yes	-9.0	0.0	4.200	0.500	2.707	16.6	103.5	41.0
20F	95	F	no	10.0	0.0	4.063	0.477	2.790	25.1	112.6	52.9
20G	50	M	no	14.0	-10.0	3.000	0.184	1.783	14.0	35.9	18.1
20H	50	M	no	8.5	10.0	4.407	0.466	2.580	36.7	66.2	34.8
20I	80	M	yes	-8.5	-10.0	5.089	0.506	3.367	56.7	75.6	35.3
20J	80	M	yes	-12.0	10.0	4.678	0.481	2.778	36.4	63.9	44.9
22A	69	M	no	9.0	10.0	3.493	0.273	2.020	40.5	47.6	18.0
22B	69	M	no	14.5	-10.0	3.562	0.305	2.191	34.4	46.6	20.3
22C	76	M	no	-11.0	-10.0	5.465	0.475	4.033	46.7	47.9	18.8
22D	40	F	drop	2.0	0.0	4.663	1.841	2.147	20.5	189.0	43.9
22E	77	F	no	4.0	0.0	6.705	1.836	5.140	61.1	237.6	50.6
22F	77	F	no	3.5	-10.0	6.713	2.842	5.243	60.0	262.0	60.0
22G	73	M	yes	1.0	-10.0	7.166	2.163	5.473	70.9	144.9	57.2
22H	85	M	no	1.0	10.0	5.192	2.040	3.731	59.1	107.2	26.3
22I	68	M	yes	-2.0	10.0	8.481	2.404	6.241	58.6	177.0	65.6
22J	64	F	yes	-1.5	10.0	5.437	2.228	2.808	29.9	139.8	75.6
25A	70	M	yes	15.0	0.0	9.810	6.713	4.022	63.6	183.3	51.4
25B	70	M	no	15.0	0.0	10.247	7.170	4.155	57.2	191.7	43.2
25C	60	M	no	15.0	0.0	7.913	3.333	4.837	62.1		
25D	60	M	no	15.0	0.0	9.247	6.349	4.903	47.4	203.9	47.2
25E	57	M	no	9.0	0.0	8.940	4.422	6.742	70.2	263.5	49.0
25F	57	M	yes	11.0	0.0	9.375	6.439	6.511	78.4	214.5	61.9
25G	60	F	no	15.5	0.0	3.903	0.379	3.180	14.9	50.5	15.7
25H	60	F	no	13.0	0.0	4.876	0.559	3.655	31.3	89.4	21.5
25I	77	M	yes	10.5	0.0	7.445	4.764	3.438	42.5	207.4	72.2
25J	77	M	yes	9.5	0.0	8.577	6.030	4.328	47.4	278.8	36.1
26A	61	F	yes	13.0	0.0	5.705	0.886	4.863	37.5	156.2	86.3
26B		M	no	10.0	0.0	9.937	1.620	10.866	100.5	104.4	35.7
26C	75	M	no	14.0	0.0	4.622	1.391	3.131	37.3	175.0	36.5
26D		M	no	11.0	0.0	4.802	1.355	4.377	31.1	135.7	23.0
26E	64	M	no	-14.0	0.0	6.578	0.432	6.226	90.6	22.1	18.9
26F	48	M	no	-13.0	0.0	7.260	0.453	7.737	45.1	30.7	22.9
26G	64	M	no	-9.0	0.0	4.061	0.995	3.391	53.3	104.5	30.9
26H	48	M	no	-10.0	0.0	4.952	1.230	4.404	27.5	100.7	26.0
26I	73	M	no	-8.0	0.0	5.007	1.234	4.788	31.2	123.9	33.1
26J	73	M	no	-14.0	0.0	7.593	0.481	7.685	79.7	19.8	13.6

Appendix B - Injury Summary

Test	Age	gender	Injury Description
11D	89	F	Closed complete fracture of talar neck. Mild subtalar dislocation
20E	95	F	Lateral ankle subluxation, bimalleolar fracture with displacement.
20I	80	M	Closed complete oblique fracture of calcaneus from posterior side into subtalar joint.
20J	80	M	Closed complete comminuted fracture of calcaneus from posterior cortex into subtalar joint.
22D	40	F	Transverse comminuted fractures of lateral malleolus and tibia plafond. Anterior subluxation of talus. Comminuted fracture of calcaneus into subtalar joint.
22G	73	M	Closed complete transverse fracture of lateral malleolus. Total transverse tear of middle section of anterior tibio-fibular ligament.
22I	68	M	Longitudinal split of mid-substance of peroneus brevis tendon, retinaculum intact.
22J	40	F	pilon fracture. Severe ankle disruption.
25A	70	M	Incomplete fracture of lateral process of talus.
25F	57	M	4 cm longitudinal tear of middle of peroneus brevis tendon approximately over lateral malleolus. Longitudinal tear of middle tibiocalcaneal leaflet of deltoid.
25I	77	M	Fracture of calcaneus from posterior cortex into subtalar joint.
25J	77	M	Horizontal fracture of calcaneus.
26A	61	F	Multiple comminuted fractures of calcaneus.