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Human Ankle Impact Response in Dorsiflexion

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ABSTRACT

Although various automobile accident surveys showed between 20 to 30% of lower extremity injuries involved the foot or ankle, there is little information in the existing literature on the the injury mechanisms of ankle injuries for automobile occupants involved in frontal impacts. This study addresses the injury to ankles involving dorsiflexion caused by impact loading to the bottom of the foot. Types of injuries include malleolus fractures and ligament avulsions and ruptures.

Nine pair of cadaver and two Hybrid 3 lower limbs were impacted on the bottom of the foot with a 16 kg pneumatically propelled linear impactor. A horizontally oriented bar struck the foot 62 mm distally of the ankle joint with velocities between 3 and 8 m/s. The proximal end of the tibia/fibula was fixed to a rigid support through a triaxial load cell. Load cells on the foot and impactor along with high-speed photography provided the response data of the foot and ankle. Range of motion tests were done before and after testing.

Peak axial loads ranged between 1.1 and 5.0 kN and moments at the ankle between 52 and 300 N-m. Maximum dorsiflexion ranged between 25 and 74 degrees during impact. Inversion or eversion motions were not noticed in these tests where the foot was free to dorsiflect. There was no correlation between pre-impact range of motion values and injuries or peak forces. The best correlation for an injury pre-diction in these tests was the maximum angle of dorsiflexion; those with less than 45 deg of dorsiflexion did not sustain any injuries. Hybrid 3 responses were not comparable to cadaver responses.

INTRODUCTION AND SPECIFIC AIMS

Foot and ankle injuries are reported in various automobile accident surveys as occurring in approximately 3% to 6% of the reported injuries. These types of injuries are rarely life threatening and generally have an AIS of 1 to 2 with a maximum of 3 (AAAM,1990). Their Injury Priority Rating (a measure of economic cost) is only 0.2-0.3% of all injuries (Carsten and O'Day, 1984). The injuries are thought to be caused by upward and rearward displacement of the toe pan after the knee has impacted and become fixed in the lower dash (States, 1986, Nahum, 1968) or by the brake pedal (Backaitis, 1987), leading to high axial loading of the foot-ankle-tibia and possibly accompanied by extreme dorsiflexion.

The specific aim of this study is to investigate impact response of the foot to axial loads with dorsiflexion.

SIGNIFICANCE

Ankle injuries are one of the most common musculoskeletal afflictions and as a group, ankle fractures are the most common intra-articular fracture of a weight-bearing joint (Attarian, 1985, Diehr, 1988, Markolf, 1989). By far most of these injuries occur in sports and involve ligament tears or malleolar fractures or both. The most common mechanism of injury is thought to be supination-external rotation or inversion. However, ankle fractures from vehicle accidents can involve displaced talus fractures, tendon ruptures, as well as malleolar fractures and ligament tears. Hence, injuries resulting from traffic accidents tend to be high energy injuries and have a poorer prognosis than the low energy sports injuries or those caused by slips and falls. The rate of load also tend to be higher than sports related injuries.

Determination of the responses of the ankle to impact loading would result in the development of biofidelic dummies to better understand the mechanisms and mitigation of these type of injuries.

PREVIOUS WORK

A review of the medical literature revealed several cases of auto accident occupants with ankle injuries. For reference an illustration of the bones and major ligaments of the ankle joint is shown in Figure 1. One case involved a person with bilateral medial malleolus fractures, radial styloid fracture (a process on the radius bone of the arm), and achilles tendon rupture (Martin, 1986). These injuries were hypothesized to have occurred as a result of forced dorsiflexion. Another case involved the rupture of the posterior tibiotalar, tibio calcaneal and lateral ligaments as well as a complete rupture of the tendo Achillis and a fracture of the talus. The mechanism of injury was thought to be caused by forced dorsiflexion (Wilson-Macdonald, 1988). A third paper reports on eight cases of ankle dislocation without fractures. Six of the cases involved motor vehicle accidents and the mechanism was hypothesized to be a force applied to a plantarflexed foot (Wroble, 1988). A fourth paper reported a fractured medial malleolus and an avulsion of the deltoid ligament (O'Leary and Ward, 1989). They hypothesized the mechanism of injury to be a primary abduction-external rotation force with a secondary adduction force applied. Reidelbach and Zeidler (1983) present injury data for five cases involving foot and ankle injuries due to high speed offset frontal collisions. These cases contained malleolar and bimalleolar fractures, displaced joints, ruptured ligaments, and fractured talus, calcaneus, cuboid and metatarsal bones. They attribute the increased incidence of lower extremity injuries among belted occupants to an impact shock syndrome caused by intrusions of the footwell (Zeidler et al, 1981).

A number of researchers have studied the properties of the ankle and its ligaments as well as determined the function of ligaments by serial sectioning (Siegler et al, 1988 and 1989; Attarian et al, 1985; Rasmussen et al, 1982; Markolf et al, 1989; Hirsch and Lewis, 1965; Stormont et al, 1985; Lassiter et al, 1989; Allinger et al, 1990; Lauge-Hansen, 1950). All of these studies are valuable contributions to the understanding of the kinematics of the ankle and the properties and function of the ligaments. However, none of them considered injuries under high axial loads or high rates of loading. Culver (1984) loaded some cadaveric lower legs in an Instron materials testing machine, pseudo-statically at a fixed dorsiflexion, till ankle joint failure at approximately 5.5 kN

at which point the foot had rotated medially approximately 35 degrees relative to the tibia resulting in fracture of the anterior portion of the calcaneus. Fractures of the talus and calcaneus have been known to occur with high axial loads on the tibia such as jumping or in aircraft crashes, presumably with little dorsiflexion (Pozo, 1984).

Agarwal and Gottlieb (1977) have shown that the ankle can be modelled reasonably well as a second-order linear differential equation with constant coefficients: $A/T = 1/(Jss + Bs + K)$, where J is the moment of inertia of the foot and plate, B is the angular viscosity, K is the angular stiffness, and s is the Laplace transform complex frequency. They found that above 10 Hz the inertial term dominates the torque and the effect of bias torque (muscle tone), becomes less effective. Although their work involved relatively small angular displacements, a study by Kearney and Hunter (1982) found that ankle stiffness decreases with larger angular displacement.

While reviewing the above mentioned literature it was noticed that the available field accident injury descriptions did not match particularly well with those produced in laboratory testing of cadaver lower legs. It is hypothesized that for malleolar fractures to occur there must be hyperdorsiflexion (and axial load) and that there may be some dynamic factors involved, particularly since it is known that for some ligaments there is a significant increase in strength during high rates of loading (Begeman et al, 1987). Gloyns et al (1979) found that 75% of the fracture-dislocations of the foot and ankle were accompanied by footwell intrusion, indicating that these injuries are produced in high speed collisions.

METHODS

The setup used for the impact testing is shown in Figure 2. An impact load was applied to the bottom of the foot using a pneumatically propelled linear impactor with a impacting mass of approximately 16.3 kg. Velocities ranged from 3.0 to 8.1 m/s. Twenty tests were run on 9 pairs of thawed fresh frozen human cadaver lower legs and a Hybrid 3 lower limb (2 of the 20 tests). Both male and female specimens were used and ranged in ages from 38 to 71. Pertinent cadaver information is presented in Table 1.

The striking end of the impactor was a 25 mm diameter bar, positioned horizontally, and covered with 25 mm of Ensolite to dampen the contact force. The bar was attached to the impactor mass through a biaxial load cell which measures the impact forces in the horizontal direction along the axis of motion (x-axis), and the vertical direction (z-

axis). A uniaxial accelerometer, mounted behind the bar in the x-direction, was used to correct the load cell data for the mass in front the load cell. Impactor velocity was determined from a magnetic pickup speed trap. The analog data was digitized at 8000 samples/second and processed as channel class 600 data.

A 12 mm aluminum plate with two uniaxial load cells was strapped onto the bottom of the foot. (These load cells were not used in tests 1 through 8). One of load cells was centered on the ball of the foot and the other on the heel. The proximal end of the tibia was held captive in a metal box with 25 mm of ensolite between it and the bottom of the box. This restraint box was attached to an adjustable height table through a triaxial load cell (x, z, and My axes) which, in turn, was bolted to the floor. This triaxial load cell was used to measure the reaction forces through the leg. A block was placed under the tibia near the ankle so that the tibia would maintain an approximate level position during the impact.

The leg was initially positioned horizontally and the foot was positioned at 90 degrees to the tibia. The centerline of the impactor was positioned at 62.5 mm above the ankle joint. This position was chosen in order to generate significant axial force on the tibia as well as dorsiflexion. Two Steinman pins with photographic targets on them were inserted in the tibia and two more in the foot. These were used to photographically measure the dorsiflexion angle of the foot relative to the tibia. One 16 mm high-speed camera, running at 500 fps, was used to capture the photographic targets. A contact switch was used to synchronize the analog data with a flash on film.

Before and after each test, range of motion measurements of the foot were made. These included pronation (inversion), supination (eversion), dorsiflexion and plantarflexion. X-rays were also taken before and after each test. Each ankle was autopsied after testing by an orthopaedic resident.

The reaction forces and moments (Fx, Fz, and My) generated at the reaction plate load cells were used to calculate the moment generated at the ankle joint by using free-body diagrams of the experimental setup.

RESULTS

The pre- and post-impact range of motion values are shown in Table 3. The pool of specimens showed a wide range of motions although the mean dorsiflexion and plantarflexion values were close to normal. An average of 6.6 degrees increase in dorsiflexion was observed post-impact in the specimens. For the specimens experiencing less than 38 degrees of dynamic dorsiflexion,

the post-impact dorsiflexion increased approximately 5 degrees. For the specimens experiencing more than 38 degrees of dynamic dorsiflexion, the post-impact dorsiflexion had increased by an average of 8.5 degrees.

Examination of high speed movies did not reveal any signs of eversion or inversion during loading or failure. The injuries observed were malleolar and bimalleolar fractures, torn muscles, and stretched and torn ligaments. The details of these injuries are shown in Table 2. All the injuries incurred by the specimens in these tests were classified as AIS 2 (AAAM, 1990). These injuries may be considered to be typical of hyper-dorsiflexion of the ankle joint and are dissimilar to those produced by other investigators who have produced talus and calcaneus fractures while simulating foot entrapment.

The peak forces and moments developed in the tests are given in Table 4. A considerable specimen to specimen variation of these forces and moments were observed and may be explained by the wide variance in sizes of the specimens. Peak axial reaction forces ranged between 1350 N to 3900 N for all cadaveric subjects and between 1850 N to 3160 N for subjects that sustained injury. Similarly the calculated ankle moments ranged between 52 to 210 N-m in all the tests and 70 to 210 N-m in tests where injuries were observed. The axial forces developed in this series of tests were much lower than the 5 kN force reported by Culver to produce crushing or compression type fractures.

The ball and heel load cells data show that the heel carried approximately twice the load of the ball.

A scatter plot of peak reaction axial force and dorsiflexion angle is shown in Figure 3. The correlation coefficient is 0.39. A correlation coefficient of 0.32 was found between peak ankle moment and peak dorsiflexion angle shown in Figure 4. As shown in Figure 5, a correlation coefficient of 0.66 was found between pre-test range of motion and peak dynamic dorsiflexion angles.

Figures 6 to 9 show the comparison of results of 5.5 m/s tests on a cadaveric specimen and the Hybrid 3 dummy. The Hybrid 3 dummy was tested at a joint stop angle of 20 degrees. Figure 6 shows that the reaction forces generated in the Hybrid 3 dummy were much higher than those generated in the cadaveric specimen. Also, the ankle moments generated in the Hybrid 3 dummy were much higher than those in the cadaveric specimen (Figure 7). Figure 8 shows the moment-time history of angular deflection for the two tests. The moment-angular displacement relationship from the two tests are shown in Figure 9. It can be seen from Figure 9 that the Hybrid 3 dummy has a high inertial loading followed by a drop in loads, and then another high loading as the ankle joint hits the joint stop. In contrast, the cadaver

ankle joint maintains loads during angular deflection.

From the moment-angle curve a compliance value (rad/N-m) was calculated. For the cadaver tests these values ranged from 0.0014 to 0.003 rad/N-m. From the angular deflection-time curves, the half period of an equivalent cyclic loading has been calculated to be between 18 to 35 Hz. These values of ankle joint compliance and natural frequency agree with those reported by Agarwal and Gottlieb (1977).

Figure 10 shows a scatter plot of injury or no injury as a function of peak dorsiflexion angle. The data indicate that approximately 45 degrees dorsiflexion angle is the threshold for ankle injuries in this series of tests. It is not clear whether this is valid for living humans with muscle tone. Additionally, peak axial forces and moments do not appear to be good indicators of ankle injuries in this test series, since the correlation coefficients between axial forces or moments and dorsiflexion range between 0.32 to 0.39.

DISCUSSION

It would appear that dynamics and dorsiflexion are very important in generating field-like injuries. The foot cannot be held at a fixed angle relative to the tibia for these injuries to occur. Other ankle tests (with the foot fixed from dorsiflexing) noted foot eversion during failure (Culver, 1984). The tests conducted in this study did not show any signs of eversion (or inversion) during loading or failure. Entrapment can lead to injuries, but of a different type than those observed in this series.

Since axial loading in these tests did not come close to the failure load level found by Culver, eversion and calcaneus fracture injuries were neither expected nor observed. However, under large dorsiflexion angles the foot may not be prone to eversion.

In this series of 18 cadaver tests, 5 of the 6 female subjects sustained injuries whereas only one of twelve male subjects was injured. This suggests that there may be different tolerance limits for males and females. Data on the strength of human long-bones show females to be weaker (McElhaney, 1976), which supports these results. However, there were not enough tests in this series to establish separate tolerance limits.

In order to improve the Hybrid 3 ankle joint, these tests indicate that the joint stop should be changed to 45 degrees and the joint should supply some resisting moment. All the non-injured cadaver moment-angle results are plotted in Figure 11. In this figure a preliminary ankle stiffness corridor for the Hybrid 3 dummy is suggested. In addition it is suggested that some damping,

such as friction, be maintained and some rubber bump stops be incorporated to avoid an abrupt loading at the end of its range of motion.

CONCLUSIONS

1. The forced high-speed dorsiflexion of the foot relative to the tibia in this test series has produced some of the ankle injuries observed in motor vehicle accidents.

2. Peak axial forces and bending moments did not show any correlation to ankle injuries.

3. At first glance it appears that 45 degrees dorsiflexion angle is the threshold for ankle injuries for cadaveric specimens. However, it is not clear whether this angle is related to the rate of angular deflection. Further tests are required in which 45 degrees dorsiflexion is produced at varying rates of angular deflection.

4. Five of the six female specimens tested sustained injuries while only one of twelve male specimens had an injury. This suggests possibly different injury threshold for male and female subjects. There is insufficient data from this series to establish the above hypothesis.

5. The data indicate that the Hybrid 3 dummy ankle joint should be modified to accommodate at least 45 degrees of dorsiflexion. A damping element and rubber bump stops should also be included to reduce ankle loadings to more realistic values and to simulate muscle tone.

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TABLE 1
Cadaver Physical Data

Test No.	Subject	Age	Sex	Weight (kg)	Height (cm)
1,2	846	59	F	76	169
3,4	914	62	M	53	140
5,6	690	71	M	49	169
7,8	807	38	F	65	168
9,10	469	69	M	64	171
11,12	646	37	M	49	181
13,14	805	58	F	55	167
15,16	786	60	M	79	180
17,18	828	67	M	78	170

TABLE 2
Ankle Injuries

Test	Subject	Description
1	846R	Avulsion of posterior talo-fibular libament from fibula (several small pieces). Complete avulsion of deltoid ligament - anterior, medial and posterior. Torn ligaments at origin: flexor hallicis, tibialis posterior, and flexor digitorum longus.
2	846L	Medial and posterior deltoid avulsion fractures.
3,4	914	No injury.
5,6	690	No injury.
7	807R	Torn ligaments: tibialis posterior and flexor digitorum longus. Torn muscle belly of peroneus longus and peroneus brevis.
8	807L	All segments of deltoid ligament ruptured at origin at medial malleolus - small segment of anterior portion was spared.
9,10	469	No injury.
11,12	646	No injury.
13	805R	No injury.
14	805L	Bimalleolar fractures - AO type C. Medial malleolus transverse fracture at level of mortice.
15,16	786	No injury.
17	828R	No injury.
18	828L	Medial malleolar fracture.

TABLE 3
Range of Motion Data
(degrees)

Subject	Test No.	Dorsiflexion		Plantarflexion		Pronation#		Supination#	
		Pre	Post	Pre	Post	Pre	Post	Pre	Post
914L	3	20	20	50	50	30	30	30	40
914R	4	20	30	40	45	30	30	25	30
690L	5	32	33	47	42	24	24	26	25
690R	6	25	30	38	38	18	25	29	26
807R	7	40	48	46	47	40	38	45	45
807L	8	40	49	48	48	35	51	34	39
469L	9	0	15	25	25	30	28	50	48
469R	10	0	5	20	30	30	30	55	60
646R	11	1	3	29	30	15	25	45	50
646L	12	0	11	25	22	25	25	40	60
805R	13	28	30	42	45	32	45	34	33
805L	14	26	29	46	51	20	16	36	38
786L	15	15	26	35	40	19	21	24	22
786R	16	19	28	37	34	18	11	40	29
828R	17	22	26	38	36	5	6	24	30
828L	18	23	33	36	40	7	8	14	10
Mean		19	26	38	36	24	26	34	37
Normal*		20		50					

Note: Measurements were not done on 846 (tests 1 and 2).

- Includes forefoot rotation

* - (Hoppenfeld, 1976)

TABLE 4

Summary of Peak Test Data

Test No.	Velocity m/s	Impact Fx N	Impact Fz N	React Fx N	React Fz N	Calc My N-m	Ball Forc N	Heel Forc N	Dorsi Ang Deg	Injury
1	5.3	2470	880	1990	750	210			70	mal fx, del av
2	4.2	2490	950	2090	800	180			57	del av fx
3	3.0	1400	1280	1610	810	110			37	none
4	3.2	1470	1250	1650	940	110			43	none
5	4.4	1800	1875	1820	1500	70			36	none
6	4.4	2290	2630	3900	2300	140			45	none
7	5.4	1540	1990	3160	2100	104			74	tend
8	5.3	1070	1730	2900	1650	130			61	del lig rup
9	4.1	1800	2100	1350	1500	95	800	1420	38	none
10	5.5	2250	2800	2200	2200	142	1160	2500	38	none
11	4.1	2500	1650	1400	1240	102	950	1500	25	none
12	5.8	2250	3400	2250	2600	182	1240	2850	40	none
13	5.5	1900	2200	1900	2050	52	900	1840	33	none
14	6.9	1500	2400	1850	2300	70	1100	1750	57	bimal fx
15	5.5	2850	2660	2370	2200	138	1200	2550	31	none
16	7.3	3600	3750	3150	3200	160	1730	3530	44	none
17	6.2	2400	3970	2200	3900	135	1390	2700	36	none
18	8.1	3600	4950	2850	3900	135	2060	3150	49	mal fx
19	4.6	2400	2400	2800	1730	170	1100	2400	20	Hyb 3
20	5.6	3600	3750	5000	3300	300	3750	4000	20	Hyb 3

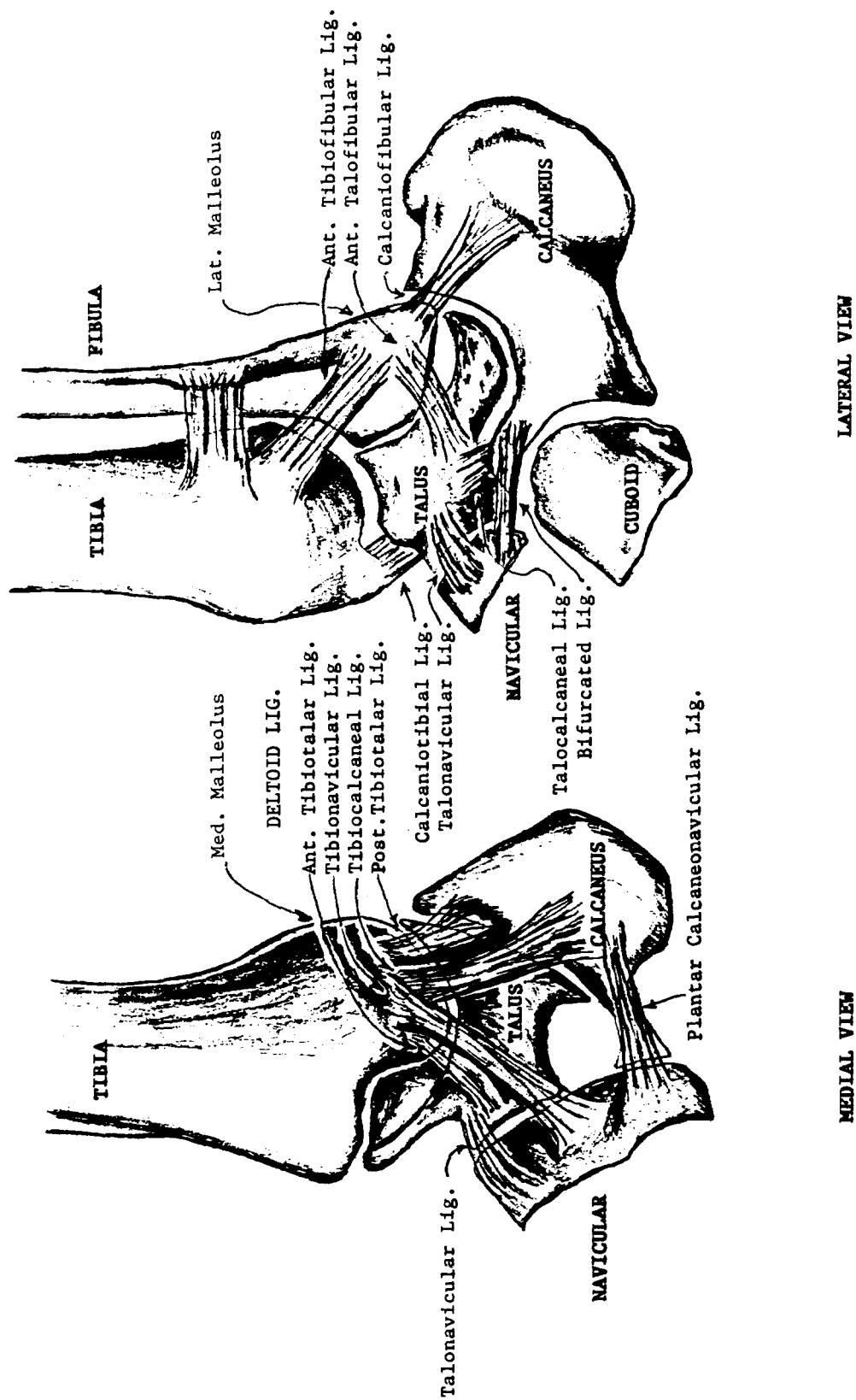


Figure 1 - BONES AND LIGAMENTS OF THE ANKLE JOINT

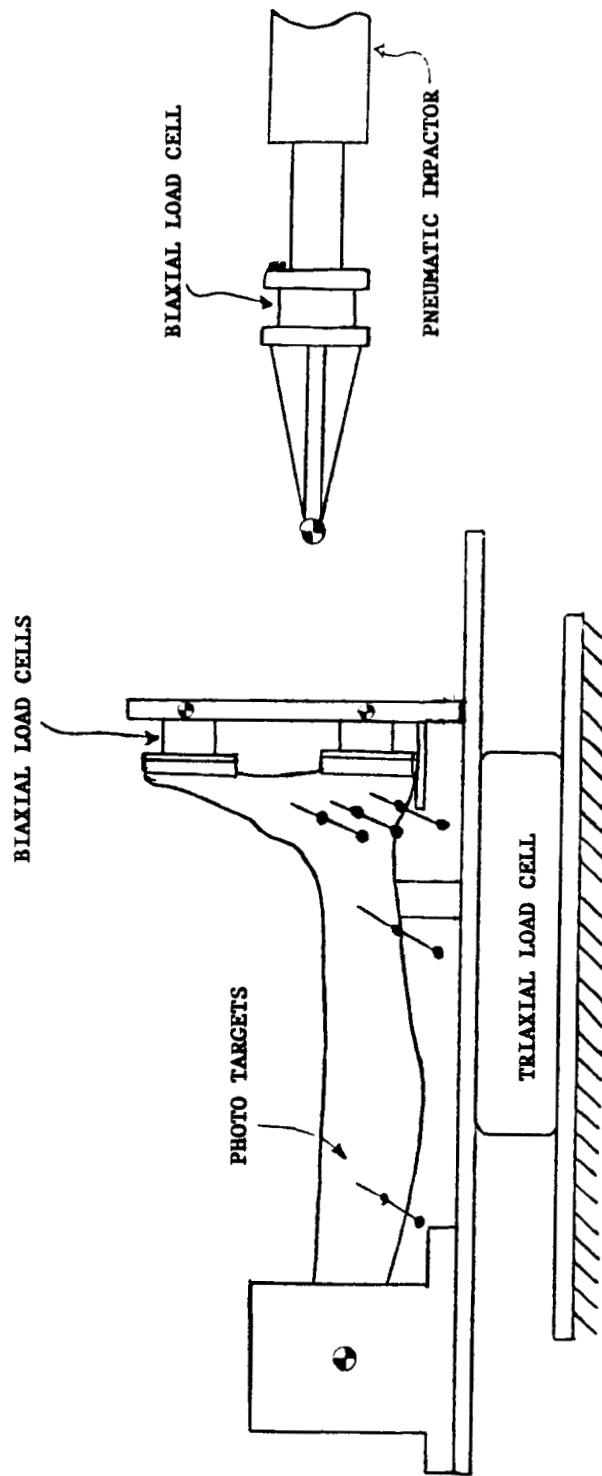


Figure 2 - IMPACT TESTING ARRANGEMENT

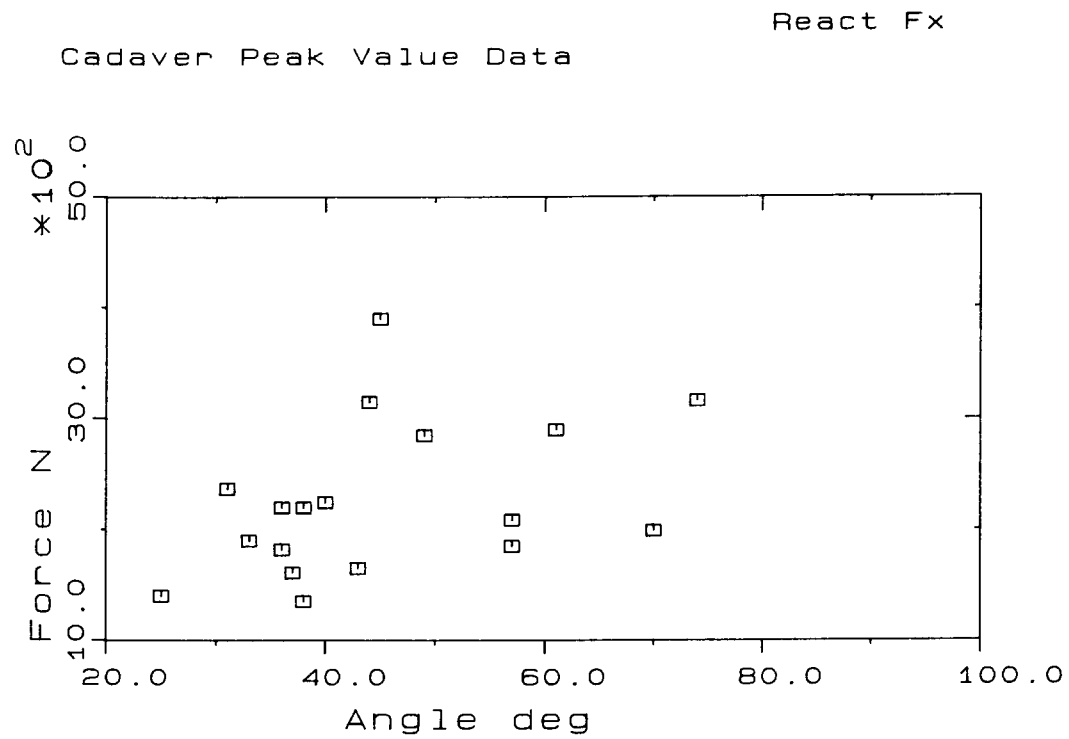


Figure 3. Peak Axial Force versus Maximum Dorsiflexion Angle.

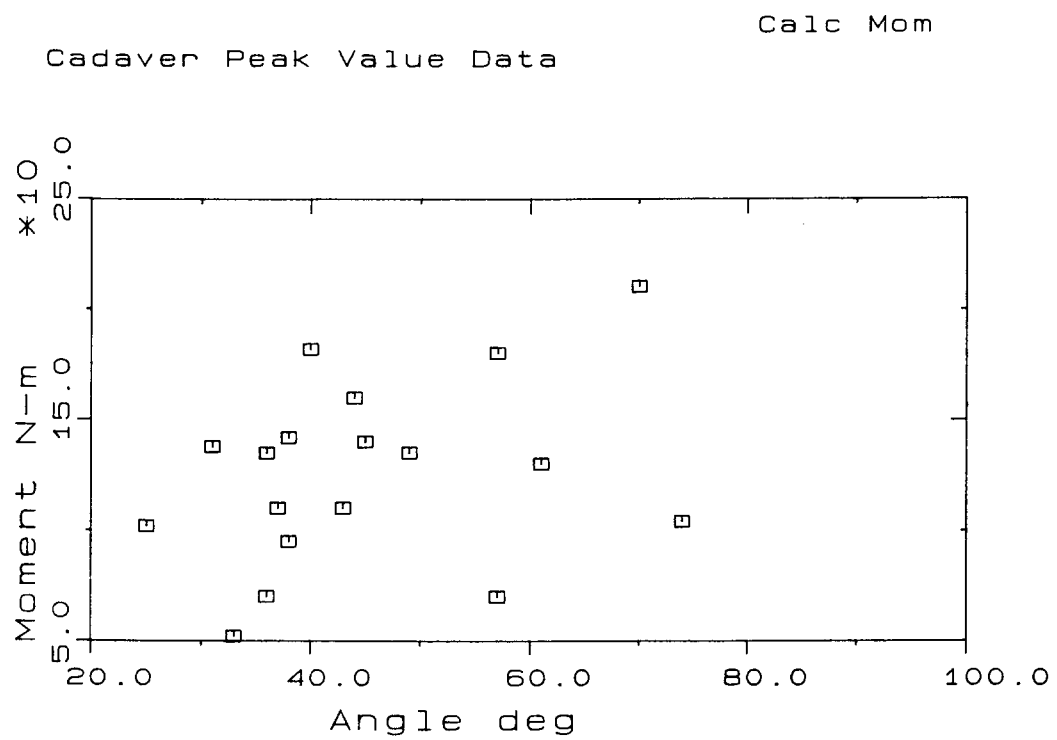


Figure 4. Peak Ankle Moment versus Maximum Dorsiflexion Angle.

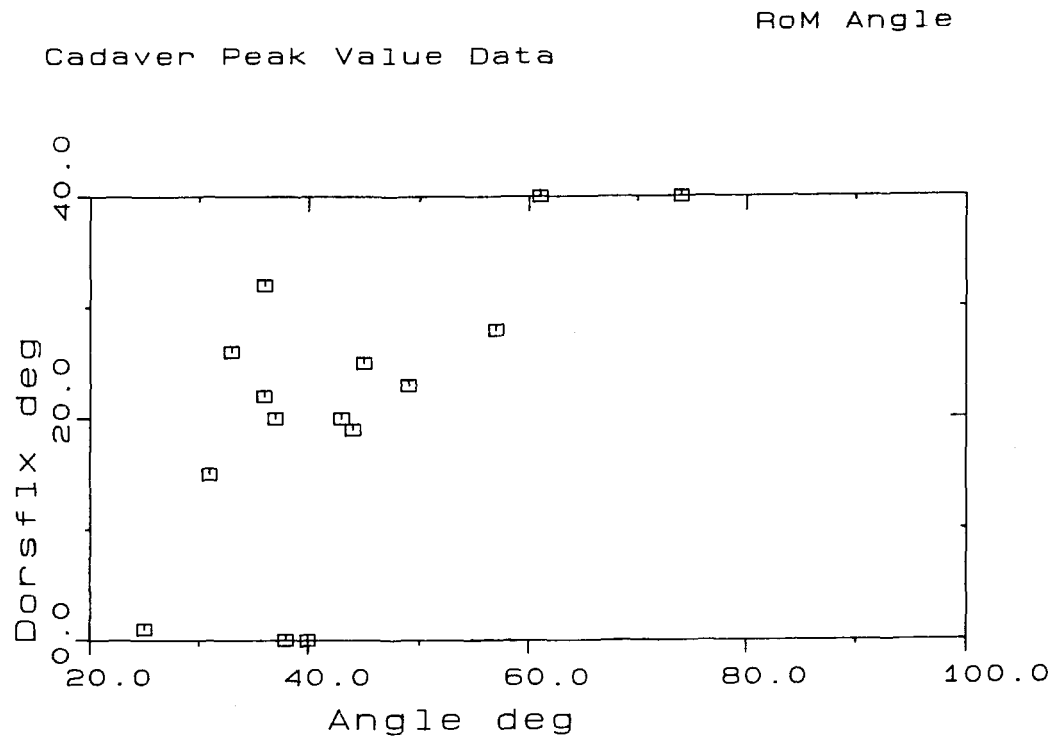


Figure 5. Pre-test Range of Motion versus Maximum Dorsiflexion Angle.

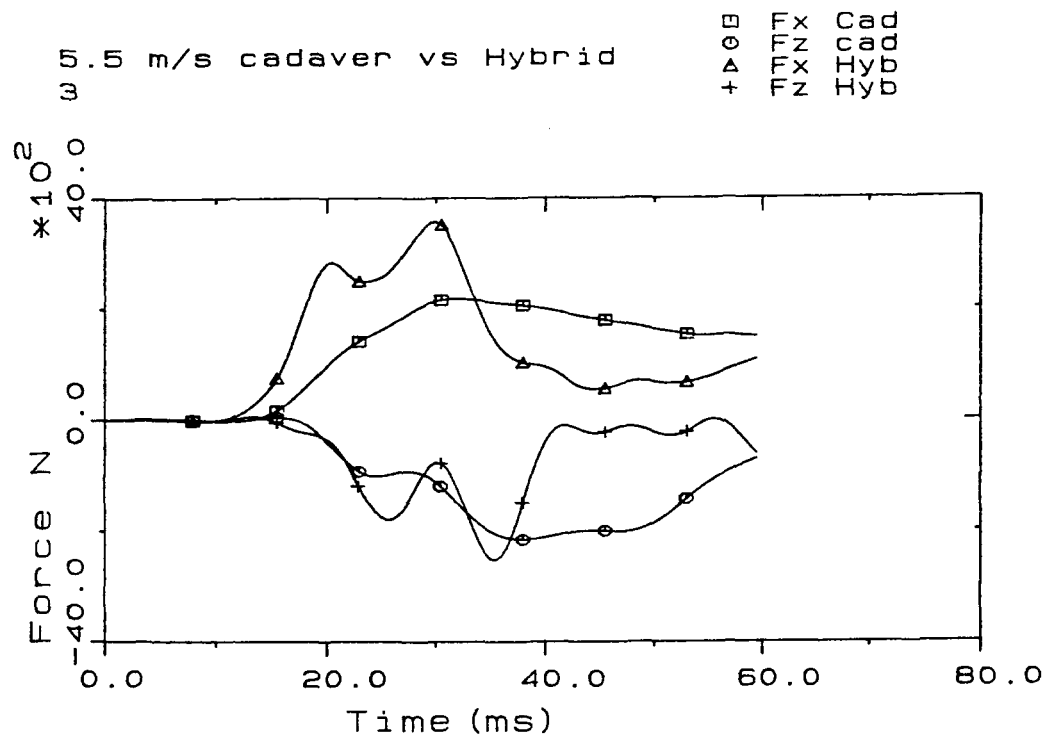


Figure 6. Reaction Forces on Cadaver and Hybrid 3 Ankle Joint.

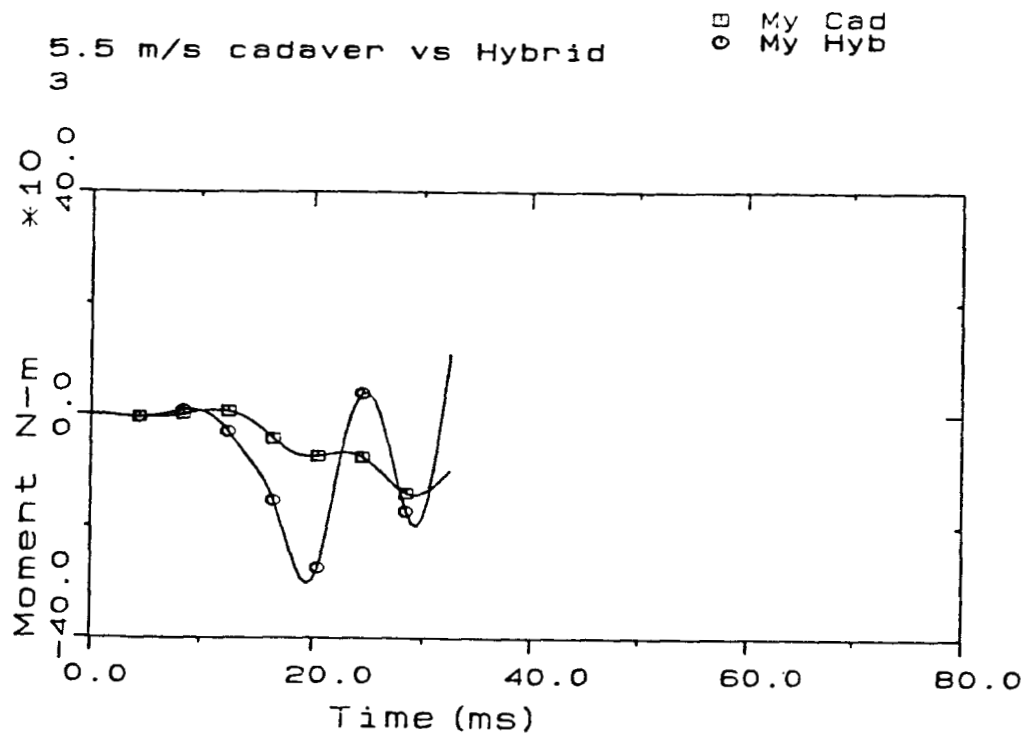


Figure 7. Calculated Ankle Moment for Cadaver and Hybrid 3.

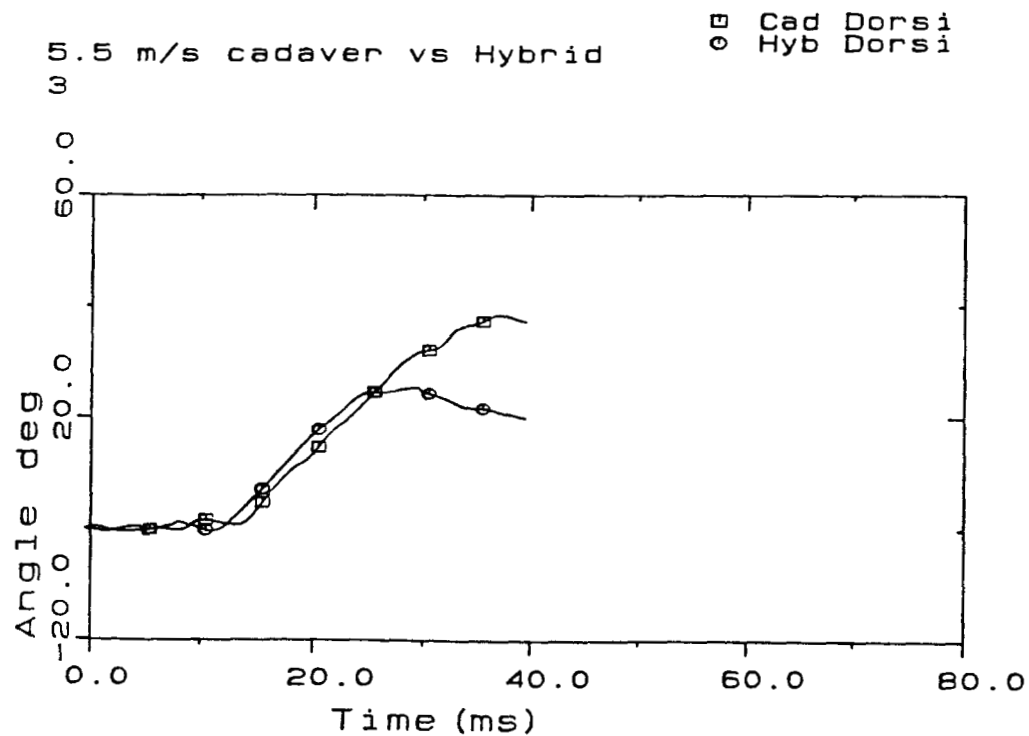


Figure 8. Dorsiflexion Angle Response of Cadaver and Hybrid 3.

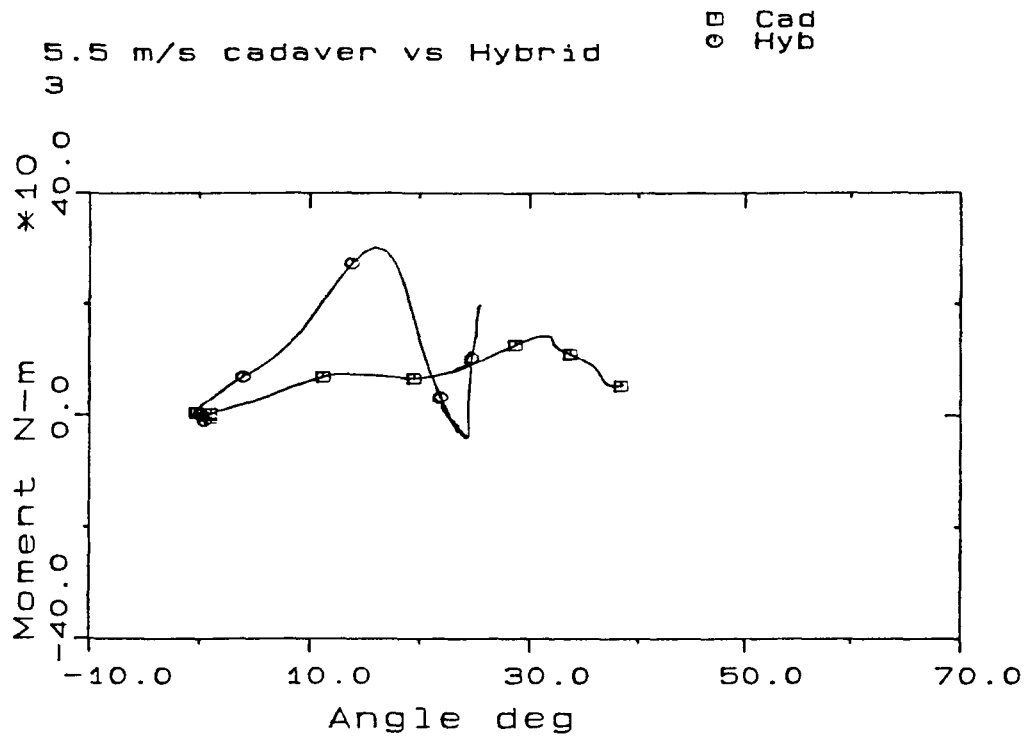


Figure 9. Moment-Angle Relationship of Cadaver and Hybrid 3.

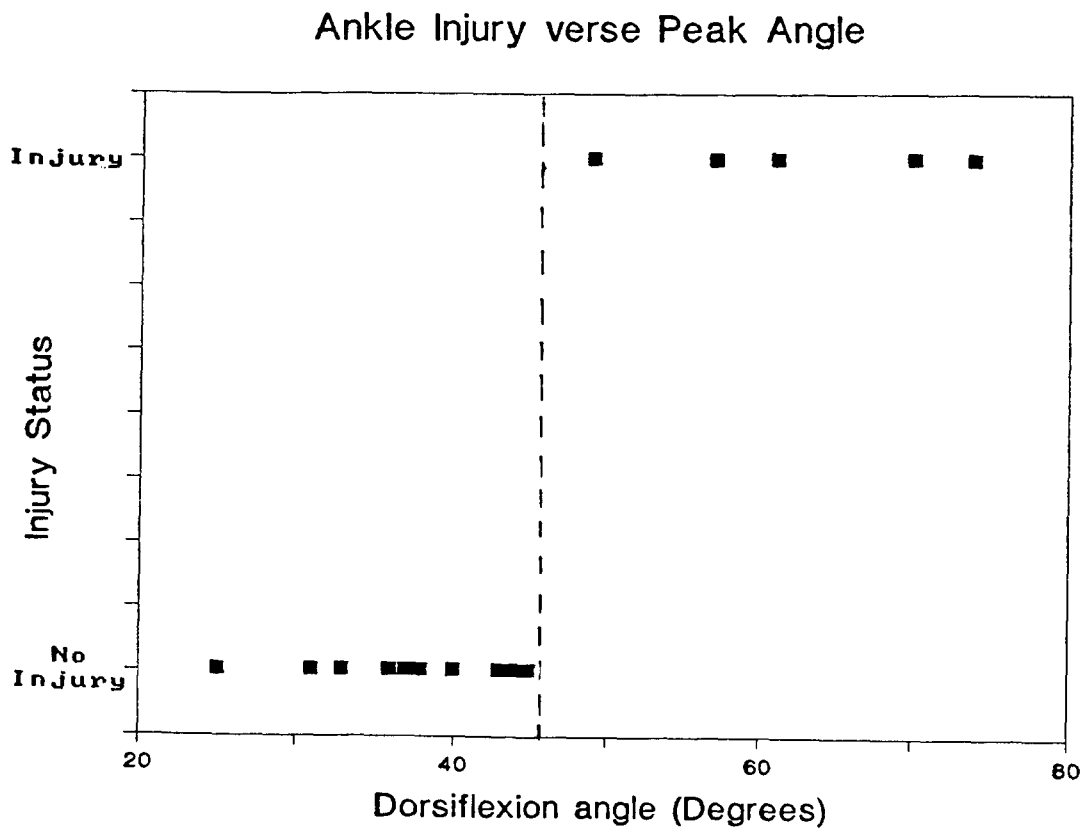


Figure 10. Injury - No Injury versus Maximum Dorsiflexion Angle for Cadavers.

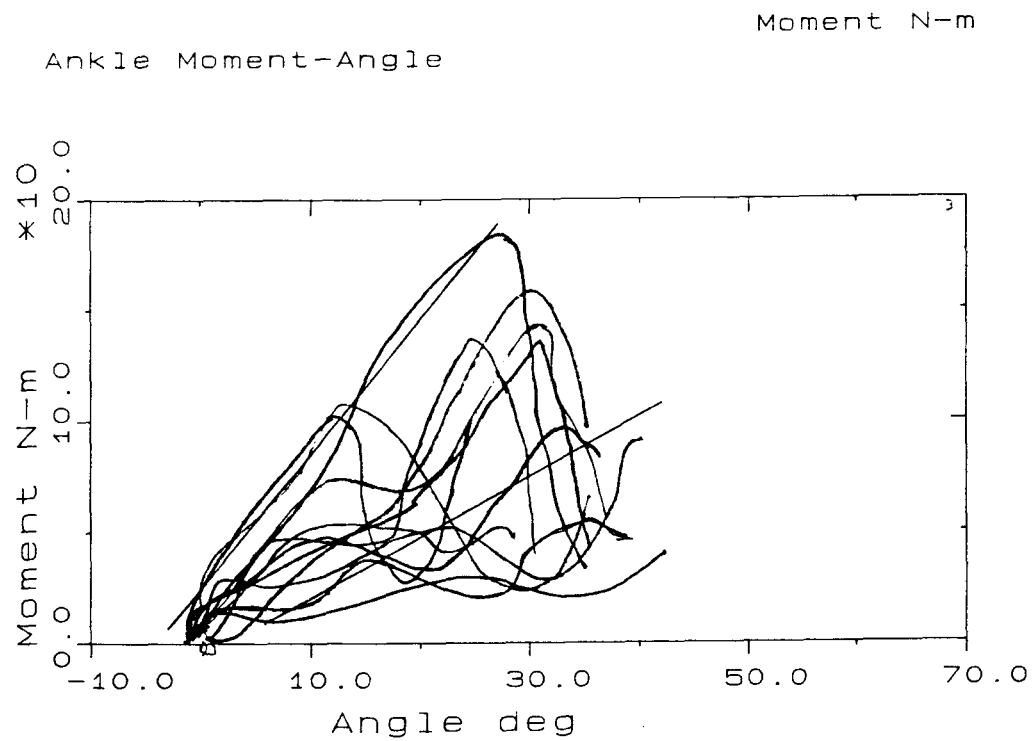


Figure 11. Moment-Angle Relation for All Non-Injured Cadavers.