SIMULATION OF MUSCLE TENSING IN PRE-IMPACT BRACING

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ABSTRACT

Reflexive muscular bracing of the lower extremities can play a significant role in occupant kinematics and kinetics. It is known that the muscles of the legs can generate substantial resisting forces that could be used to help the occupant ride down the crash. The consequences of muscular bracing on injury are investigated with numerical simulations and laboratory studies. The tests suggest that muscular bracing acts to increase the peak axial load in the tibia and the impulse absorbed by the leg; however, these values were not injurious when intrusion was not present. Numerical simulation of intrusion suggested that loading of the tensed leg could exceed the published injury threshold.

THE INCREASING USAGE of active and passive restraint systems is expected to have a continuing beneficial effect on occupant survivability in motor vehicle crashes. Paradoxically, it may also lead to an apparent increase in debilitating lower extremity injuries [Dischinger et al., 1994]. These high energy injuries have probably occurred along with the more life threatening head and torso injuries. However, prior to the introduction of airbags, the occupants frequently died from multiple head and chest injuries. As the increased use of seat belts and airbags reduces the severity of head and torso injuries, there is an increasing need to focus on the lower extremities, particularly the foot, ankle, and lower leg. Severe crashes can result in substantial shock loading or entrapment to the foot and ankle, with associated consequences for the entire leg. These effects are often associated with toepan intrusion [Reidelbach and Zeidler, 1983;States, 1984;Pattimore et al., 1991;Otte et al., 1992], which occurs as the front structure of the vehicle buckles and collapses; pushing the engine compartment components against the firewall.

Another factor which can have a profound influence on lower limb injury is the act of reflexive bracing by the occupant. In those crashes where there is sufficient time to react, it is likely that occupants will brace themselves against impact. It is known that under extreme duress, people can exhibit strength far beyond their apparent limits. Under normal laboratory conditions, it has been demonstrated that volunteers can exert substantial bracing loads against a toepan-like structure from the seated position [Armstrong et.al., 1970; Gordon et al., 1977; Begeman et al., 1980]. These loads are sufficient to significantly alter occupant kinematics and kinetics.

To study the effects of this reflexive bracing, a series of occupant computer simulations, pendulum tests, and sled tests using Hybrid III dummies and human cadavers was performed. While it is not possible to know with precision the degree of bracing and its consequences in actual car crashes, some general consequences can be obtained by comparing results from the tests and simulations in this series. It will be demonstrated in the following simulations that this bracing load, combined with associated changes in body geometry, can greatly alter the loads transmitted to the lower leg.

METHODOLOGY

NUMERICAL SIMULATION - The crash conditions used as a baseline are derived from an idealization of and actual offset frontal crash [Kuppa and Morgan, 1993]. The particular crash was a 60% offset configuration involving two vehicles of similar size each with a initial velocity of approximately 56 km/h. The vehicle pulse was the same in all simulations and is shown in Fig. 1A. For those crashes involving toepan intrusion, the toepan behavior relative to the vehicle is displayed in Figs. 1B-1D. Only horizontal intrusion (i.e., longitudinal toepan translation) was considered in this study. The toepan intrudes approximately 22 cm and reaches a peak relative velocity of 5.7 m/sec at 59 ms after impact.

Since the response of the foot is expected to play a large role in ankle injury, an advanced foot model was used in the ATB occupant model [Crandall et al., 1994]. This advanced foot model incorporates five segments representing the ankle, heel, tarsal, metatarsal, and phalangeal regions of the foot. Degrees of freedom for the joints were based on accepted values from the anatomical literature [Stiehl, 1991; Kapandji, 1987].

Careful attention was given to the effect of bracing on the occupant's geometry. From volunteer studies conducted at this laboratory, it was observed that when one braces, the legs straighten and the pelvis is pushed back into the seat and up the seat back. The entire torso rotates backward as it loads the seatback to compensate for the torque at the hips. In general, volunteers were able to generate approximately twice normal standing load on the toepan. Arm loading against the steering wheel was included for completeness since a driver is unlikely to brace with the legs alone. These initial conditions, shown in Fig. 2A, can be contrasted with those of the relaxed occupant in Fig. 2B.

The hip motion of the braced occupant tends to take up any slack in the lap belt as the pelvis rises. For this reason, the braced occupant simulations were run with no lap belt slack, while 5 cm of belt slack was used for relaxed occupants. Since slack is distributed between each side of the belt, 5 cm equates to only 2.5 cm of occupant travel before the belt begins to tighten. This is a reasonable value for standard retractors unless the occupant intentionally pulls the belt tighter.

The braced occupant's initial load against the toepan was based on the volunteer studies mentioned above. Since a 50th percentile male is modelled in the simulations, the load on each foot was chosen to be two times body weight or about 1.5 kN. The effect of muscle tension was produced by adjusting the Coulomb friction in the occupant's joints to produce constant torque levels. The knee friction produced a static resisting torque of 270 N*m. For the ankles, a torque equal to two body weights applied at the ball of the foot resulted in an equivalent resisting torque of 225 N*m. For this study, the arm joint resisting torques were estimated to be about one third of the values in the corresponding leg joints.

PENDULUM TESTS - To simulate the effect of muscle tension on test dummies and cadaveric specimens, a parametric study was conducted [Crandall et al., 1993] to develop external hardware for test subject legs. The goal of this study was to produce loading in the ankle and foot analogous to that produced by active muscular bracing and to produce similar occupant kinematics.

In order to study the effects of muscle tensing in a simple dynamic environment, a custom ankle test cell driven by a compound pendulum was used to generate impulsive loads than model toepan intrusion pulses. In this device, the pendulum provides the impulse via a transfer piston to an instrumented plate configured to act like a vehicle toepan. The actual transmitted pulse is moderated by a contact pad which is designed to shape the toepan pulse and was customized for the test series. Muscle tension in the leg was simulated by the use of a spring-loaded knee harness which acted to extend the knee. It also draws the knee together to prevent dislocation of this joint during the impact. The leg specimen (either dummy or human above knee amputation) was secured to a pin-mounted bar.

The H-point location was controlled by an adjustable bracket. This configuration avoids constraining the knee and ankle to better approximate the natural (i.e. biofidelic) response of the ankle. A review of the anatomical literature suggests that ankle response is influenced by the action of the proximal tibio-fibular joint as well as the knee. The test fixture was designed to allow the leg and foot to respond naturally to absorb the loads imposed during impact.

Prior to impact, the leg specimen was mounted into the test fixture and the position of the H-point and footplate were adjusted until the desired initial position of the leg was achieved (i.e. femur and tibia angle and initial ankle flexion and xversion angles). Sensor bridges were balanced and the knee harness was tightened until the desired preload was achieved. Typically, the preload was specified as a multiple of normal standing load as measured in the distal axial force load cell for the dummy. The human specimen uses an in-situ load cell mounted via intermedullary posts to the midshaft of the tibia. In this case, the midshaft axial force was used to indicate preload level. A final check of leg position was then taken prior to impact by the pendulum.

Test data were recorded using a DSP Technology Traq-P Digital Data Acquisition System at 10,000 samples per second with an SAE J211 channel class 1000 filter. Impactor contact velocity was measured via a 15.2 cm speed trap.

SLED TESTS - Since the actual crash environment is more complex than that simulated by the pendulum, a Via Systems HITS-713 deceleration sled was used to simulate vehicle impacts. This testing included the study of interactions between the occupant and the knee bolster which could not be simulated on the pendulum. This testing also allowed a direct verification of the validity of the pendulum boundary conditions. A knee-harness and tether system similar to that used on the pendulum tests was employed. The test subject (either Hybrid III dummy or human cadaver) was initially positioned in the seat and the restraints were attached. The knee harness was attached and tightened until the desired preload was obtained, again using the tibia axial load cell information. After preloading, final position measurements were made.

The sled used a buck adapted from a midsize passenger vehicle. In this case, the interior geometry of the 1990 Ford Tempo sedan was used. The energy absorbing steering column and lower dash panel were custom test devices developed at this laboratory and were installed to conform to the Tempo interior geometry. The tests were conducted at 50 km/h with peak decelerations of 20 g. No intrusion was generated during the tests. Both cadavers and test dummies used comparable instrumentation by which is meant that accelerations and loads were measured in anatomically similar locations.

RESULTS

NUMERICAL SIMULATION - Figure 3A shows graphical results of the axial loads transmitted to the tibia for the four simulation runs. The case of a braced occupant with toepan intrusion produced the highest loads. Its double peaks of 6.4 kN and 5.6 kN exceed the maximum of all other cases. The first peak occurs during initial toepan intrusion before significant occupant motion has occurred and is due to the toepan striking the occupant. During the second peak, the toepan is slowing down, but the occupant is still moving forward rapidly. In this case, the occupant's momentum drives the feet into the toepan. Both of these peaks are near the injury threshold for tibial fracture. The next worst case is the braced occupant without intrusion. The tibia load in this simulation achieves a peak of 5.4 kN at 25 ms and then slowly declines over the next 75 ms. The two relaxed occupant cases show substantially lower loading with peak values of 2.2 kN and 1.2 kN for intrusion and no-intrusion respectively.

A simple conjecture is the effect of bracing would be to raise the tibia loads by the preload amount over the relaxed situation. It appears, however, that there is an additional, probably geometrical, effect which is significant. Comparing the braced and unbraced cases above, it is seen that the difference in peak tibia load is 4.2 kN, while the preload on each leg is only 1.5 kN. This nearly threefold difference is likely to be caused by the much straighter legs of the braced occupant, which efficiently transmit axial load at the beginning of the crash event.

Figure 3B shows the effect of leg bracing on pelvis motion. Only the intrusion cases are plotted since the no-intrusion curves are nearly identical. The graph shows that bracing has reduced the final horizontal position of the occupant's pelvis by 5.9 cm. This is due primarily to the difference in initial position (3.5 cm) and the tightening of the lap belt as a consequence of bracing. If the relaxed occupant simulation is rerun with no belt slack, the pelvic displacement is reduced by 1.8 cm which leaves about 0.6 cm that can be attributed solely to the restraining force of the legs.

Fig. 4 shows the occupant positions at the time of maximum upper torso displacement. The effect of arm bracing , in conjunction with the shoulder belt, has reduced head displacement significantly. The occupant leg positions for the corresponding braced and relaxed cases are very similar, but the load histories by which they arrive at these positions are different. Also of interest is the evidence that intrusion, by pushing the legs away from the firewall, may reduce the chance of lower limb entrapment below the dashboard if there is sufficient initial clearance. On the other hand, the knees of the occupants experiencing toepan intrusion rise much higher than they do with no intrusion. Thus, if the knee-to-dashboard distance is initially very small the knees may get caught underneath the dashboard and then get crushed into it as the toepan intrudes.

PENDULUM TESTS - Figure 5 shows the general effect of muscular preload on the response of a dummy leg. The static preload was equivalent to normal standing load and was measured at the distal tibia load cell. The effect of the preloading was to increase the total load and impulse absorbed by the leg. In this example, the peak axial force increased from 1897 N to 3472 N. Considering just the increase over the preload level, this represents a 64% increase in axial force due to impact. Similarly, the impulse delivered axially to the leg increased by 26% (20.3 N*s vs. 25.5 N*s). There was no appreciable increase in pulse duration. Similarly, peak anterior/posterior bending moment increased by 129%

SLED TESTS - For the Hybrid III standard leg (Figure 6), the effect of preloading was to increase the total load and impulse absorbed by the leg. The peak axial force increased from 2109 N to 2997 N. Considering just the increase over preload level, this represents a 24% increase in axial force. Similarly, the impulse delivered to the leg increased by 130% (85.4 N*s vs. 196.4 N*s). The loading occurs some 20 msec earlier for the preloaded leg and the duration is correspondingly longer. A double axial load peak was observed for the relaxed leg but not for the tensed leg.

For the cadavers (Figure 7), similar behavior was observed. Peak axial force increased from 768 N to 2239 N which is 110% over the preload level. Total impulse delivered to the leg increased by 237% (40.9 N*s vs. 138 N*s).

Knee contact with the knee bolster was observed in all tests, however less knee travel into the bolsters was observed for the tensed legs. The difference was more pronounced for the dummy tests (approx. 10 cm vs. 2 cm) than for the cadavers. The tensed cadaver suffered a tibia fracture later in the crash which made it difficult to assess the difference due to pretension on knee bolster stroke. However, dummy results were consistent with the simulation results where the tensed leg acted as an additional restraint.

CONCLUSIONS

In general, the effect of muscular preloading was to increase the efficiency of load transmission to the leg. The preloaded legs acted as additional restraints and helped the occupant ride down the crash pulse. Tibia loading levels for the sled tests did not approach the accepted threshold for injury. Horizontal pelvic motion was also reduced for the braced occupant. Post-test necropsy and radiographs did not indicate injury for the human specimens. Since intrusion was not generated in the sled tests, no definitive conclusion can be reached at present. The pendulum tests generated intrusion (13.6 cm), however the H point remained static. It is conjectured that these loads will increase with intrusion and sled tests incorporating an intruding toepan will be conducted soon to investigate this.

The result of this and previous work has shown that the legs can be used as part of the restraint system. The question to resolve is how to take advantage of this extra restraint by designing it into the system without increasing the risk of injury.

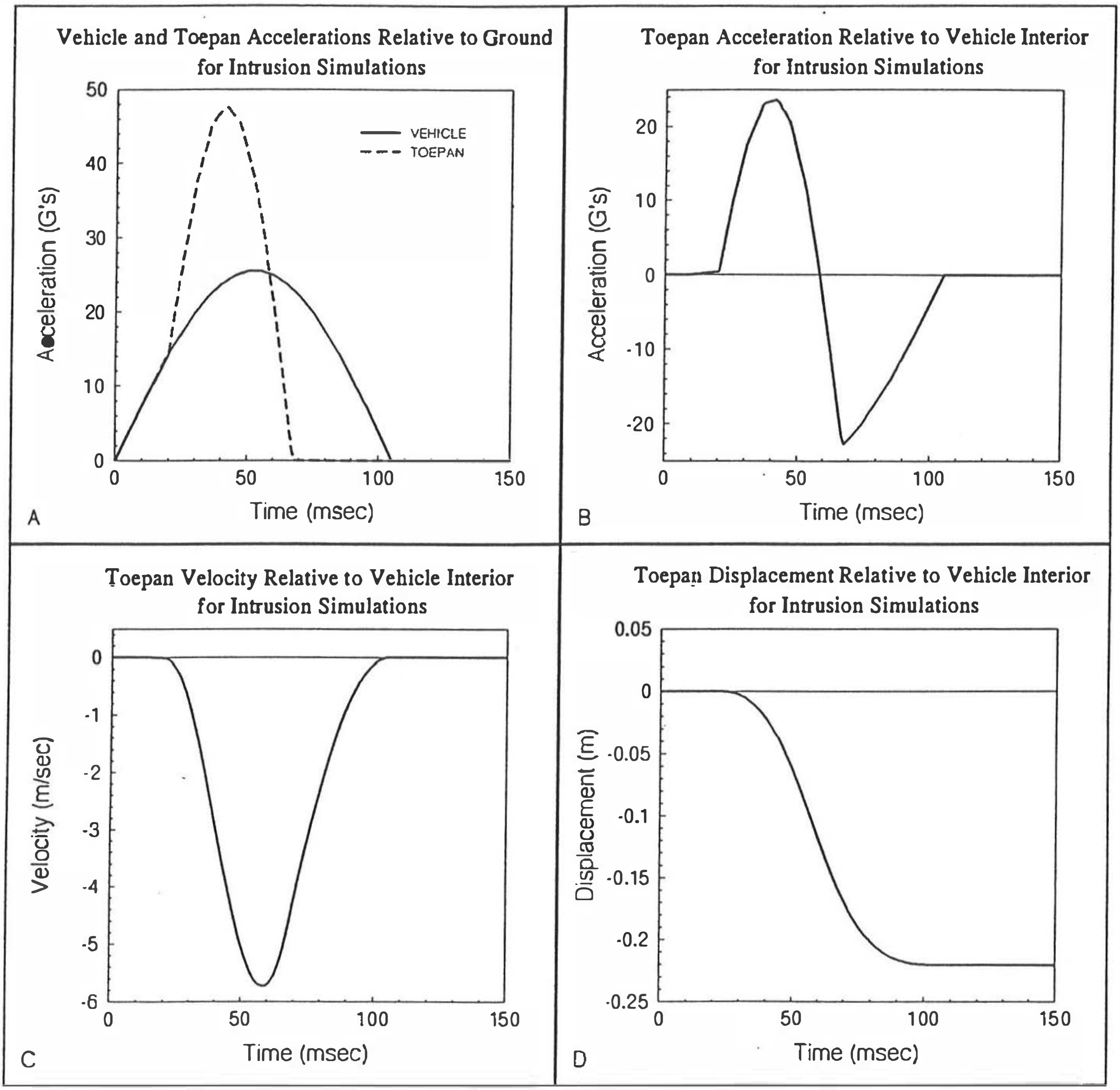


Figure 1 - Model Vehicle Pulse [from Kuppa and Morgan, 1993]

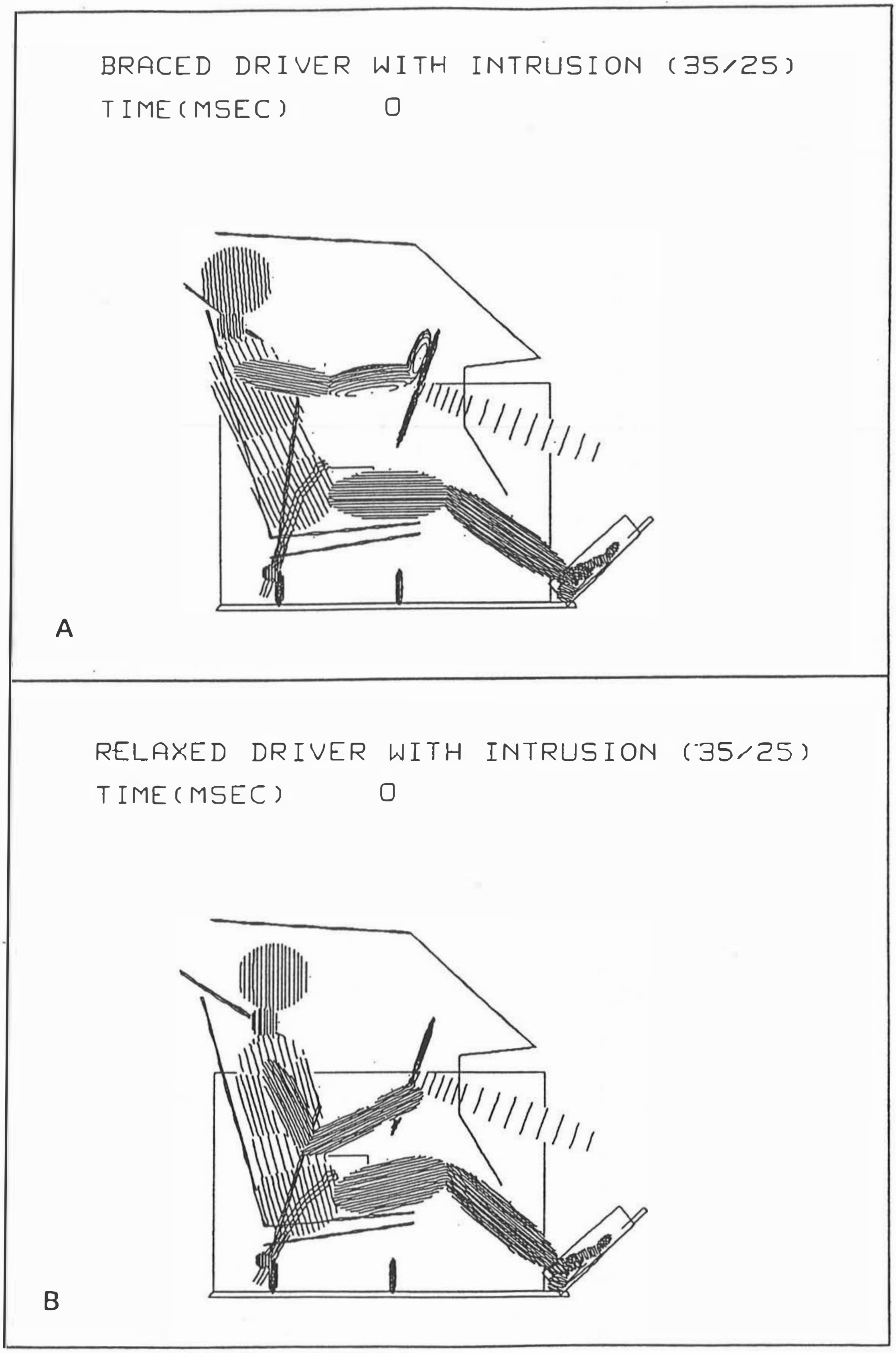
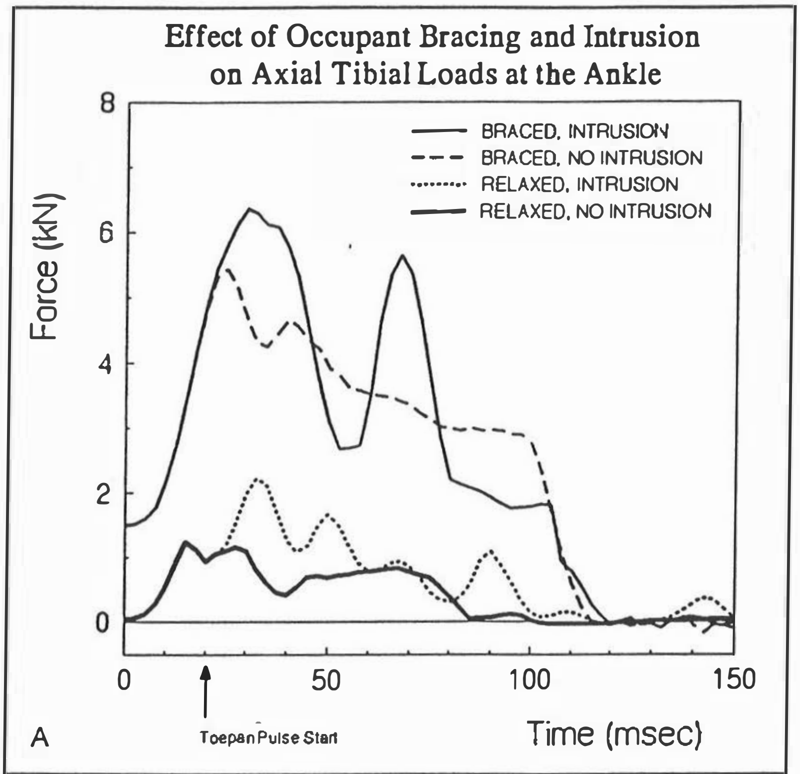


Figure 2 - Occupant Model - Initial Position



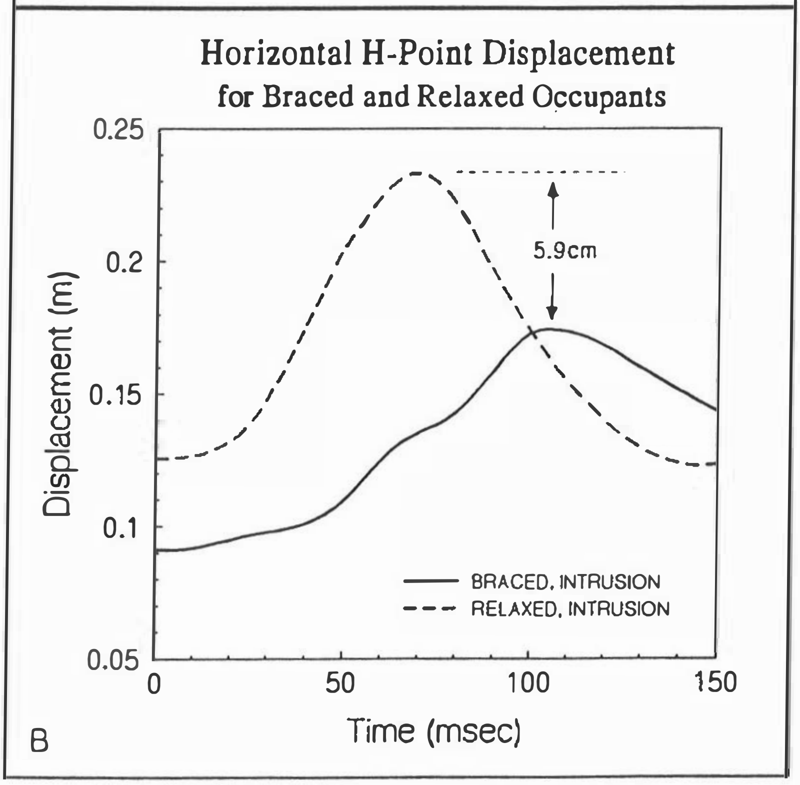


Figure 3 - Results for Occupant Model

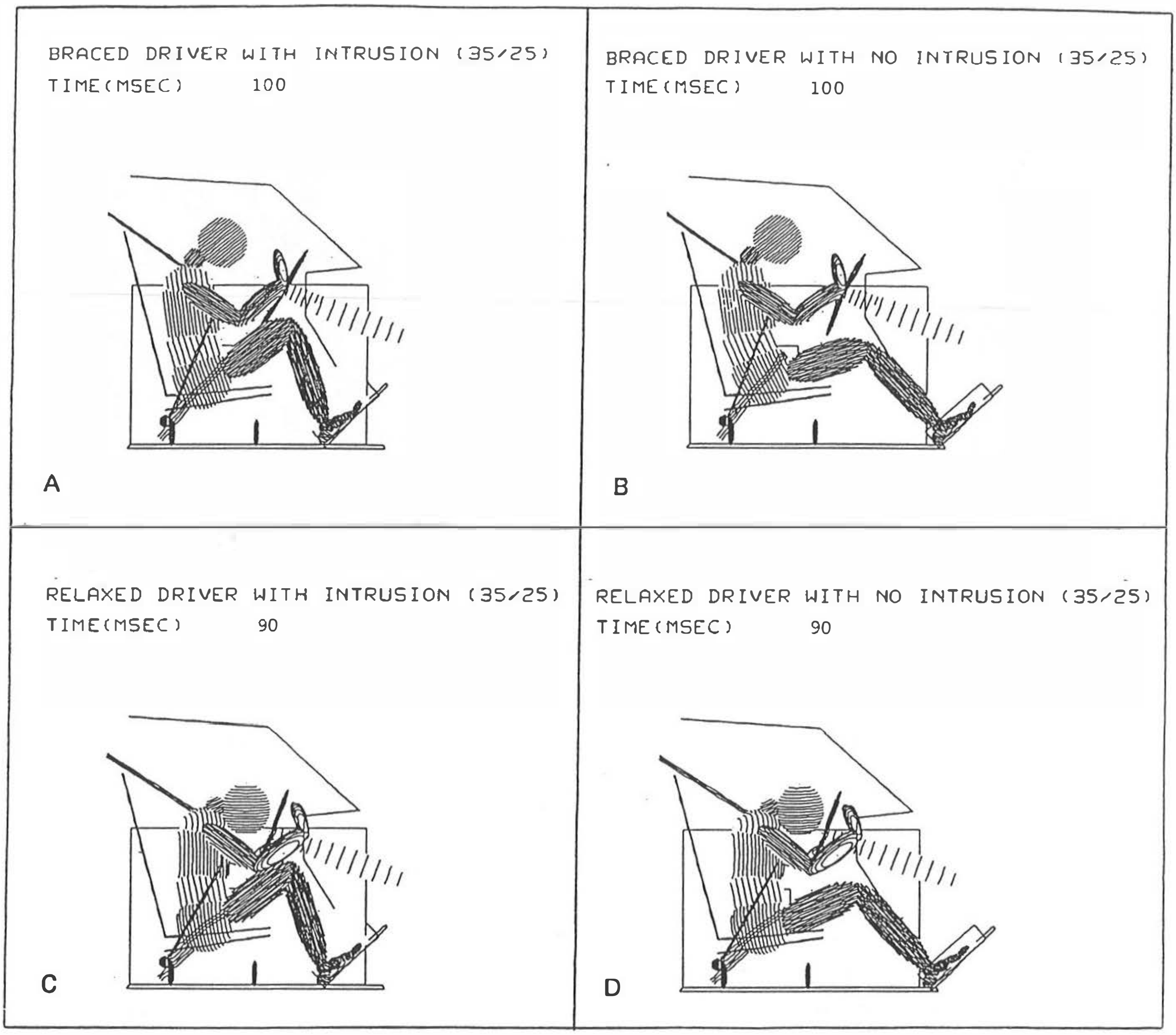


Figure 4 - Occupant Kinematics Obtained from Numerical Simulation

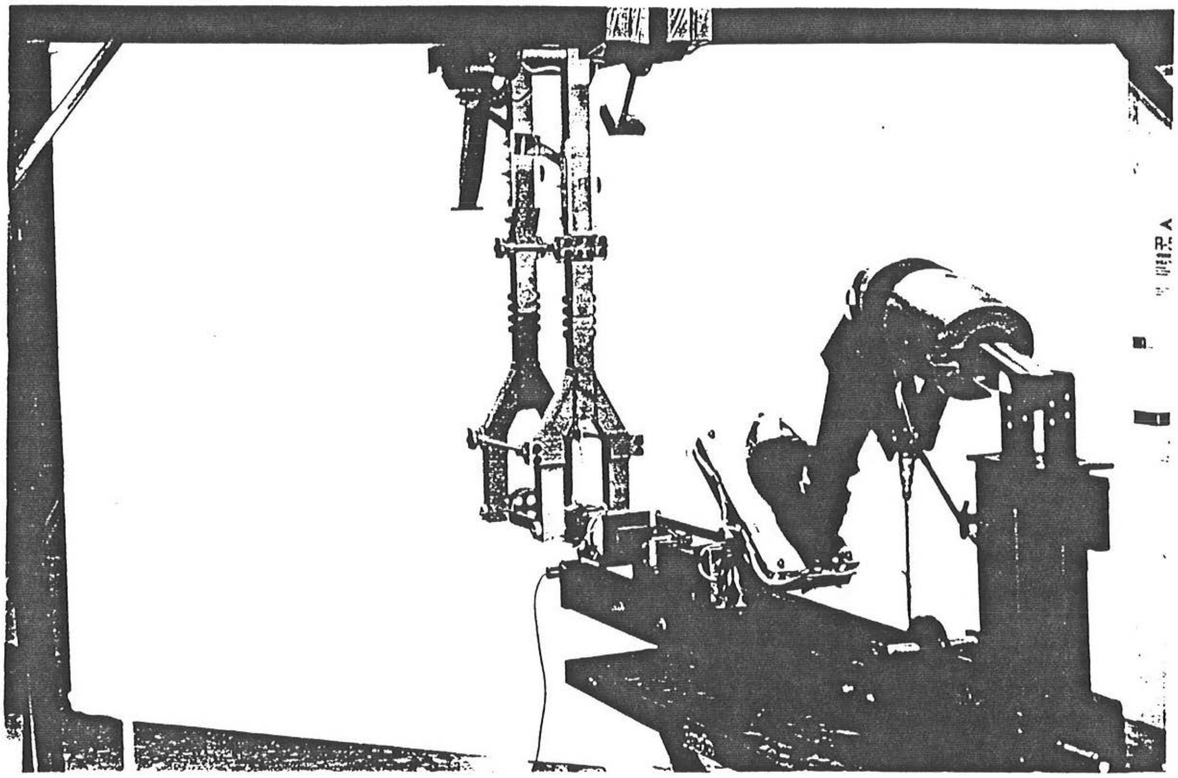


Figure 5 - Pendulum Test Fixture

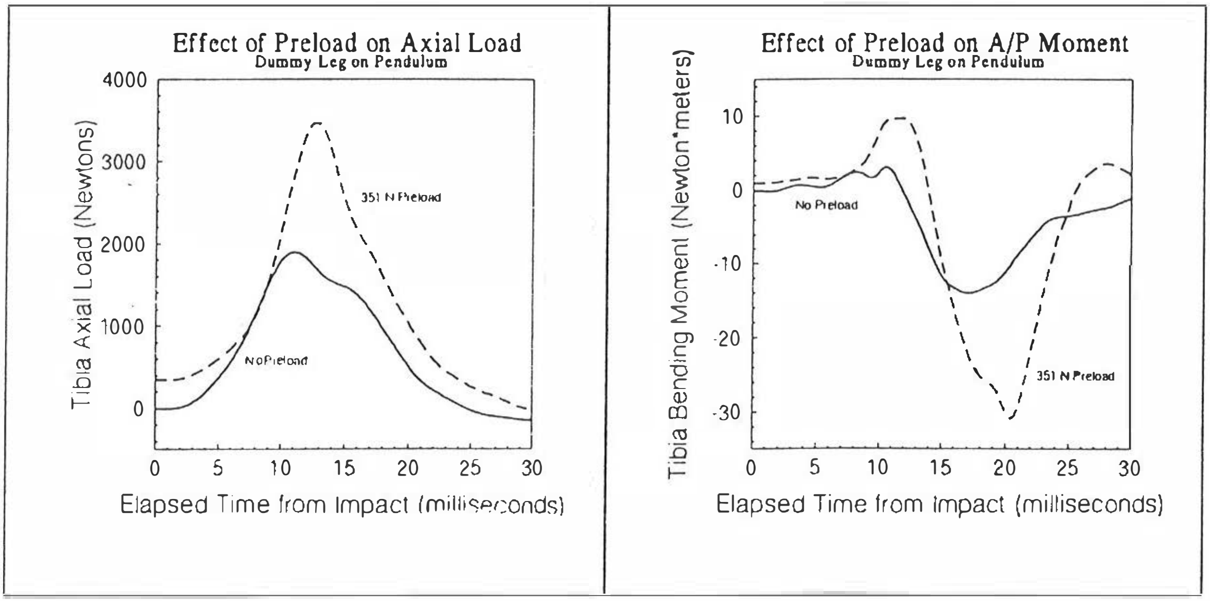


Figure 6 - Results from Pendulum for Dummy Leg

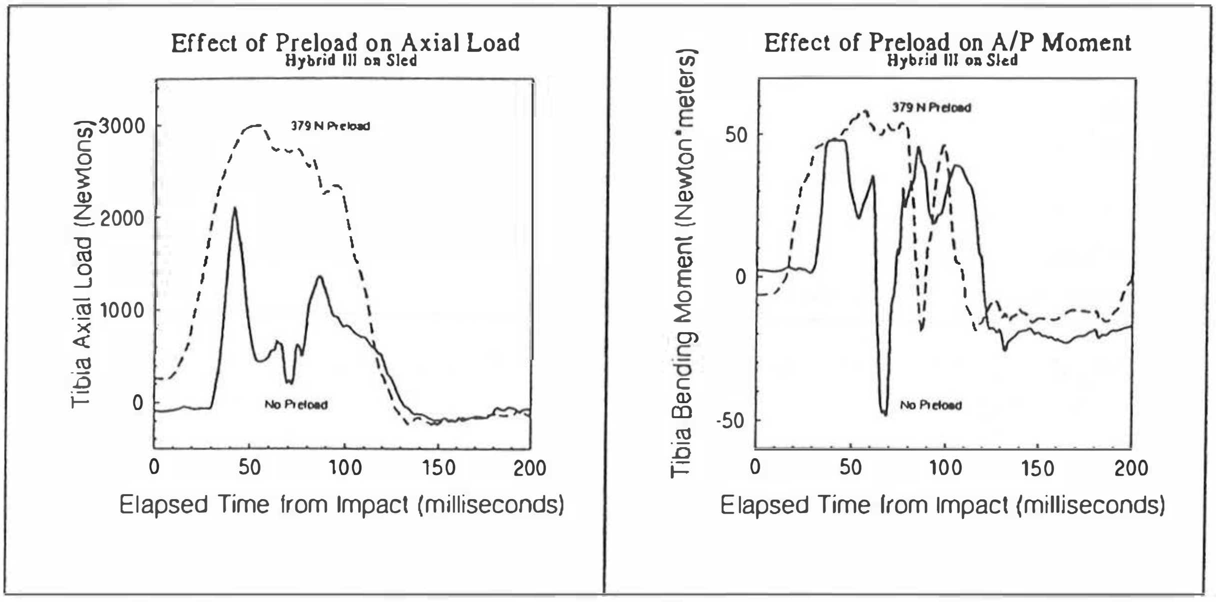


Figure 7 - Results from Sled for Dummy

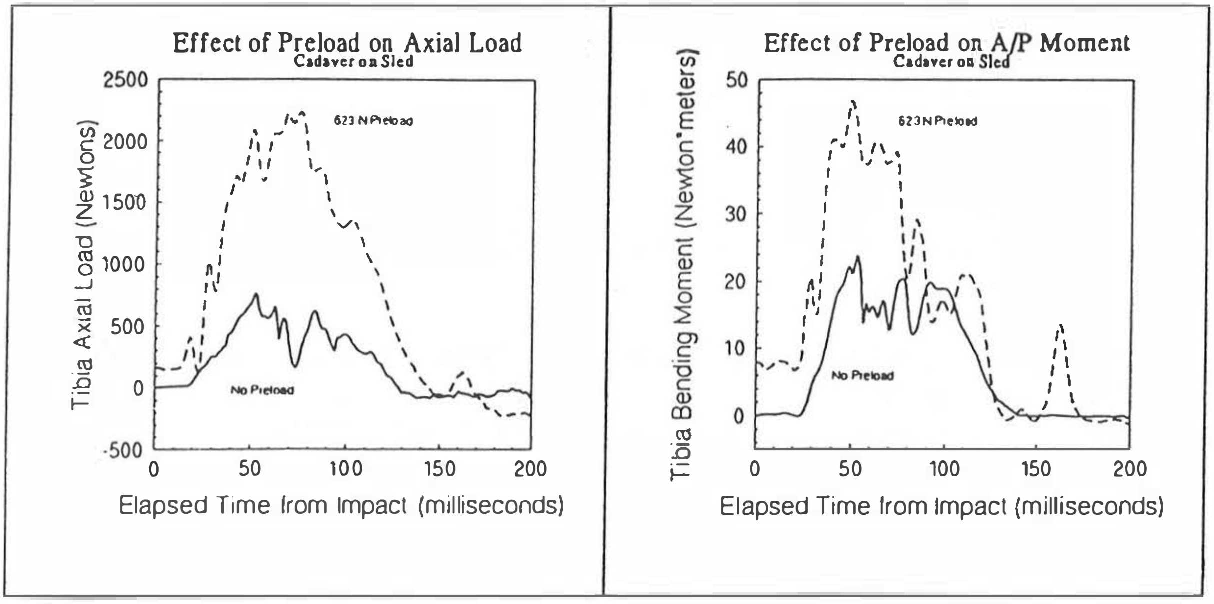


Figure 8 - Results from Sled for Cadavers

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