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The Influence of Muscle Tension on Lower Extremity Response

Jeff R. Crandall
Edwin S. Sieveka
Walter D. Pilkey
University of Virginia
Deparment of Mechanical Engineering
Charlottesville, Va. 22902

Richard Morgan
DOT/NHTSA
Washington, D.C. 20590

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Introduction

A major goal of lower extremity research is to understand the effect that muscle tension and "bracing" have on the occupant's kinematics and loading during the crash event. Volunteer sled tests (Armstrong et al., 1968, Gordon et al., 1977, Begeman et al., 1980) and postulated injury mechanisms from NASS studies (Morgan et al., 1991) suggest that muscle tensing of the lower extremities significantly changes the response of the vehicle occupant. In order to simulate the response of living humans using cadavers and anthropometric test dummies, tensing of the lower extremities must be incorporated into sled testing of human surrogates. The University of Virginia has developed a simulation model of the lower extremities in order to examine the influence of lower extremity pre-tensing on occupant response. Using the simulation model, a simple mechanical system for human surrogates has been developed to reproduce the effects of occupant bracing.

Previous Work

Armstrong, Waters, and Stapp (1968) conducted sled tests with human volunteers to determine the role of lower extremity muscle forces during impact. They determined that the restraining power of the tensed legs transferred significant amounts of the occupant's kinetic energy to the test sled and decreased the loads transmitted to the other restraint systems (i.e., belts). For the average male volunteer in the study, the maximum preload by bracing was estimated to be 1800 N (400 lb) per leg . During the sled tests, the force exerted by the lower extremities remained nearly constant for the entire deceleration event. Increased contact forces between the foot and toepan with increasing severity of impact were attributed solely to increased inertial loading. No rate or angle dependence in the flexure of the lower limb was identified.

Gordon (1977) evaluated the lower extremity response of human volunteers using a hydraulic impact device that represented an intruding toepan/floorpan structure. The apparatus delivered a maximum stroke of 20 cm (8 in.) to the lower extremities over a minimum duration of 200 ms. Contrary to the findings of Armstrong et al., Gordon identified a positive correlation between the rate of leg flexure and the resisting forces and torques. Although dynamic peak values were up to twice as high as static results, the load-time histories displayed a similar form for all tests.

Begeman et al. (1980) subjected human volunteers to low-level sled decelerations in order to investigate the effects of muscular restraint during impact. The electromyographic (EMG) activity in the major muscle groups of the lower extremities was recorded. The test results indicated that the reflex response of human volunteers was too slow to effect the occupant's response during impact. However, pre-impact contractions of the lower extremity musculature that were maintained throughout the impact event transmitted a significant amount of the occupant's energy to the test sled.

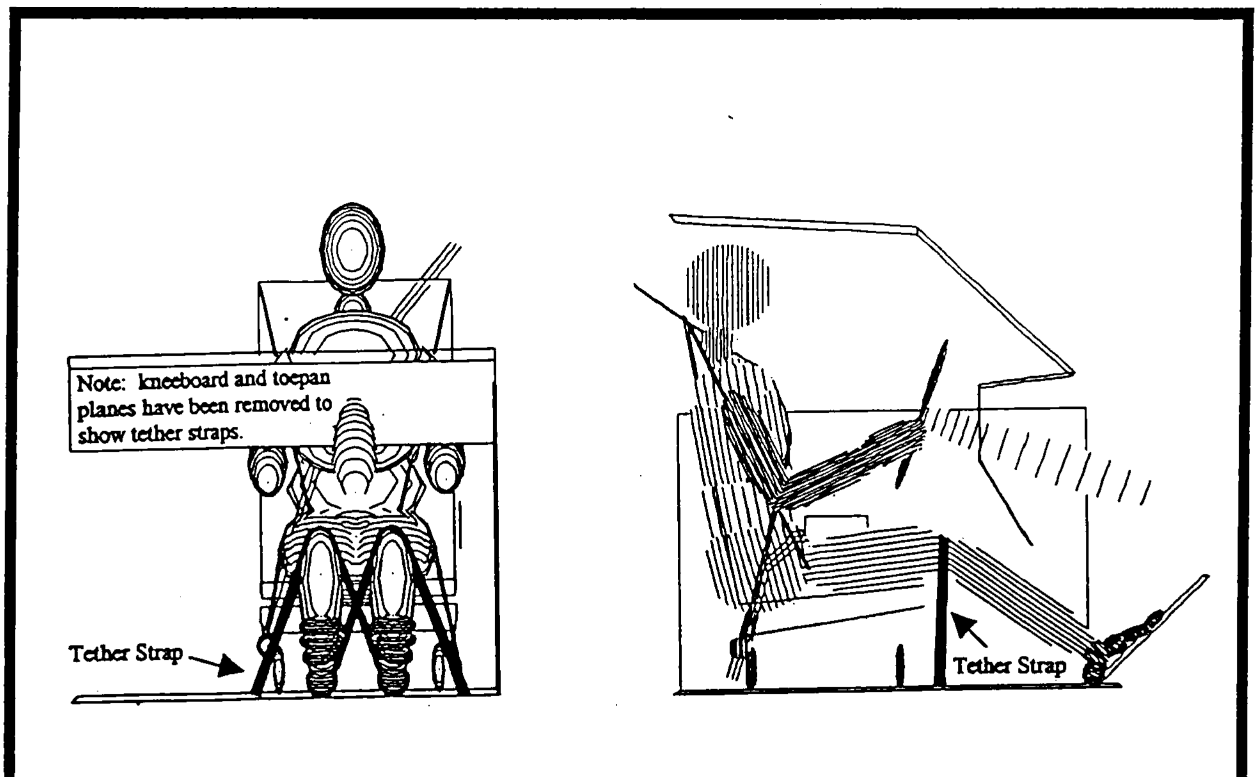
Several studies have attempted to simulate the effects of muscle tensing using human surrogates. The most common approach has been to attach external mechanical apparatus to the lower extremities of cadavers and dummies. Levine et al. (1978) affixed a friction type brace to the knees of human cadavers in order to prevent submarining during sled tests. Similarly, Begeman (1980) generated joint torques about the knee using rotational springs. Neither system was particularly effective in reproducing the response of human volunteers subjected to the same impact environment.

Simulation Model

A goal of the research at the University of Virginia is to develop a muscle tensing system for human surrogates that is simple but still duplicates the critical loading and kinematics of the occupant. To aid in the design process, a simulation model was created to investigate the influence of lower extremity muscle tensing on the occupant's response and to study the feasibility of simulating this muscle tension with a mechanical tether system. Using the Articulated Total Body (ATB) program, the anatomy, response, and geometry of the foot/ankle complex was simulated with a foot model consisting of five distinct segments: the ankle, the heel, the tarsal region, the metatarsal region, and the phalangeal regions. The heel attachment was represented by a stiff ball joint whereas the other joints contained two orthogonal axes which represent rotations about the anterior-posterior axis (i.e., inversion/eversion) and the medial-lateral axis (i.e., plantar flexion/dorsiflexion). To duplicate the plantar surface of the foot, an arch was included in the model. Obtaining representative response data for the model is one goal of the lower extremity research effort at the University of Virginia. At present, the joint flexion and stop data are approximate.

Tether System

Parametric studies with the simulation model determined that a configuration of two tether systems was necessary to duplicate the response of an occupant tensing the back and lower limb muscles. Figure 1 depicts the side and frontal views of the tether systems used in the model; the primary pair of tethers wrap around the distal thigh and attach to the floor on either side of each leg. To promote lateral stability and to prevent twisting of the lower extremities, the anchor points of the tethers were spaced approximately 30 cm. (12 in.) apart. These knee tethers applied a downward force on the leg and resulted in a reaction at the foot and toepan interface. The second set of tethers was attached to the proximal thigh near the hips and was anchored at points near the top of the seat. Since the primary knee tethers force the pelvis into the seat, the pelvic tethers were added to compensate for the substantial downward force generated by the knee tethers and to reproduce the upward component that exists for tensed volunteers. In sled tests with human volunteers, the braced occupant tends to push the torso back and upward. The pelvic tethers approximate this lift by exerting an upward force on the upper body through the hip joint.



a). Frontal view of tether system b). Side views of tether system

Figure 1.

Muscle Tensing/Tethering Conditions

The joint properties of the dummy in the ATB model were adjusted to simulate a wide range of muscle conditions and responses. Initially, the preloads in the ankle, knee, hip, pelvis, and waist joints were increased until the contact forces between each foot and the toepan were established at 1800 N (i.e., the pre-load level determined by Armstrong et al). Deviations from the initial conditions were governed by joint torque functions in the waist, hips, and knees. Moments about the joints were allowed with the angle of deflection but, despite the volunteer testing that has been performed, the relationship between the degree of leg flexion and the resistive moment is still unknown. Therefore, four cases of active and passive muscle action during impact behavior were initially studied

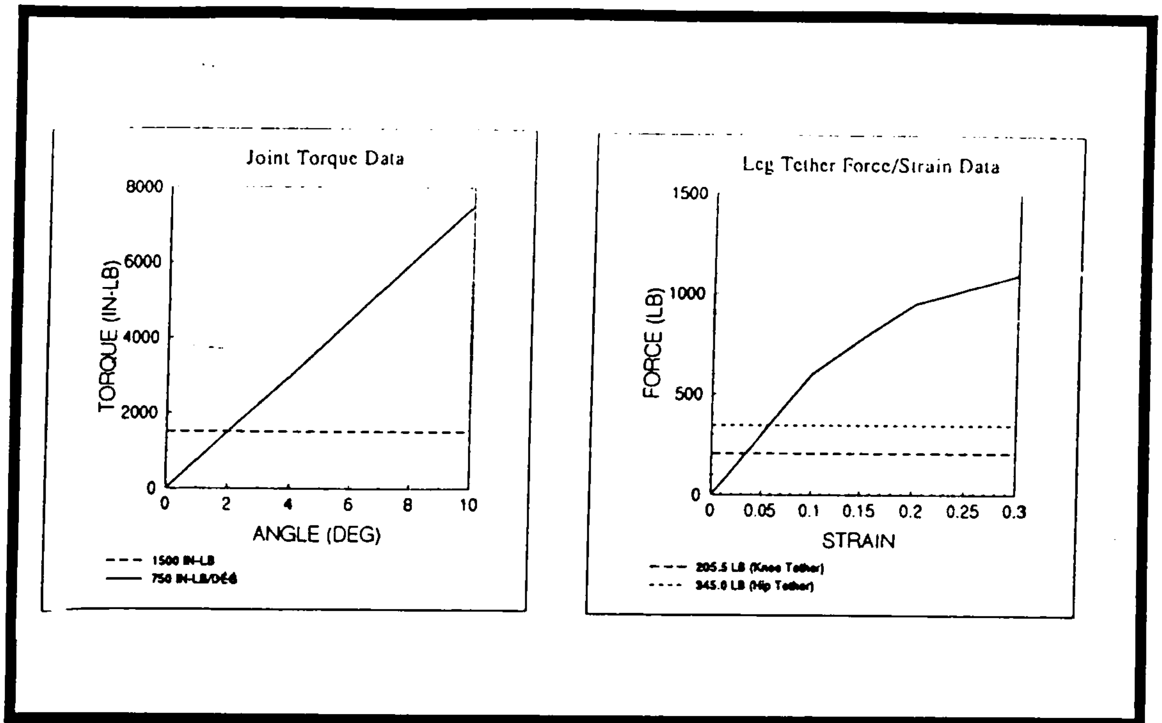
1). Relaxed legs with no preload
2). Constant joint torque with 100% elastic unloading
3). Linearly increasing joint stiffness with nearly elastic unloading (90% Energy Return)
4). Linearly increasing joint stiffness with nearly inelastic unloading (2% Energy Return).

These cases were considered to cover the range of available responses for a tensed occupant during impact. The joint torque-angle relations are depicted in Figure 2a.

The force-displacement properties of the tether systems were changed such that the occupant's response matched that of the tensed condition and as with the joint torque conditions, the relaxed state and the three cases of active and passive muscle tensing were simulated. The four conditions were

1). Relaxed legs with no tether
2). Tether system with constant force-strain function
3). Tether system with piece-wise linear increasing function and nearly elastic unloading (90% Energy Return)
4). Tether system with piece-wise linear increasing function and nearly inelastic unloading (2% Energy Return)

The piece-wise linear force-displacement functions for the tether systems are illustrated in Figure 2b.



a). Joint Torque Data b). Tether Force-Strain Data

Figure 2.

Test Environment

The simulation tests with the tethers and joint tensions used a three-point seat belt as the primary restraint. No pedal effects were included in the model and no response differences between the left and right lower extremities were anticipated. To produce potentially injurious loads and moments in the lower extremities, the vehicle characteristics of a frontal-offset collision were incorporated into the model. In particular, intrusion of the toepan and floorpan were simulated.

Intrusion data for the toepan were obtained from vehicle-to-vehicle frontal offset tests performed with late model full-size vehicles. The vehicle test were conducted with an initial velocity of 58 km/h (36 mph) and exhibited a peak vehicle deceleration of 25 G. Combining accelerometer time-histories and post-crash profiles of the toepan/floorpan structures, Kuppa and Morgan (1993) determined the characteristics of a toepan intrusion pulse. The toepan traveled 21.5 cm (8.5 inches) in the longitudinal direction and nearly 12 cm. (4.5 in.) in the vertical direction relative to the vehicle. The maximum toepan velocity with respect to the vehicle was 22 km/h (20 f/s). The toepan pulse relative to the vehicle began 40 ms. into the impact event and lasted approximately 40 ms. For the purposes of the simulation model, both the vehicle and toepan acceleration pulses were approximated as half-sine waves. In addition, only the longitudinal component of the toepan intrusion was incorporated into the model.

The relaxed occupant condition (i.e., no tethers and no joint torques) represented the baseline test case. This configuration resulted in considerable forward excursion of the dummy and flexion of the lower extremities. During the crash event, the body's inertia forced the hip and pelvis to slide forward toward the feet which were situated on the toepan. The peak contact force exerted by each foot on the toepan reached nearly 3500 N (Figure 4). The contact force, however, dropped to zero at approximately 65 ms. into the event due to the foot losing contact with the toepan (Figure 3). Although the toepan continued to intrude into the occupant compartment for another 15 ms, the inertia of the lower limbs caused the foot to loose contact with the intruding component.

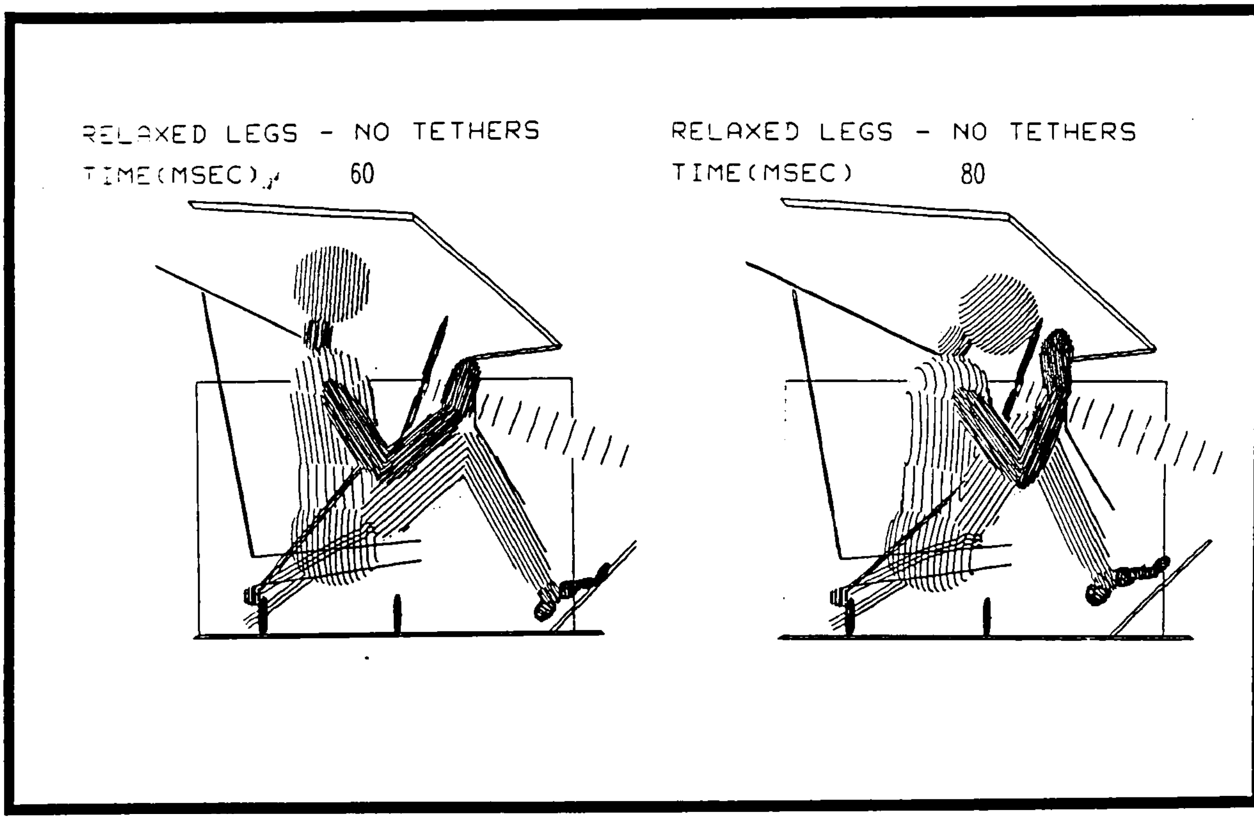


Figure 3. Relaxed legs with no tethers and no joint-torque functions

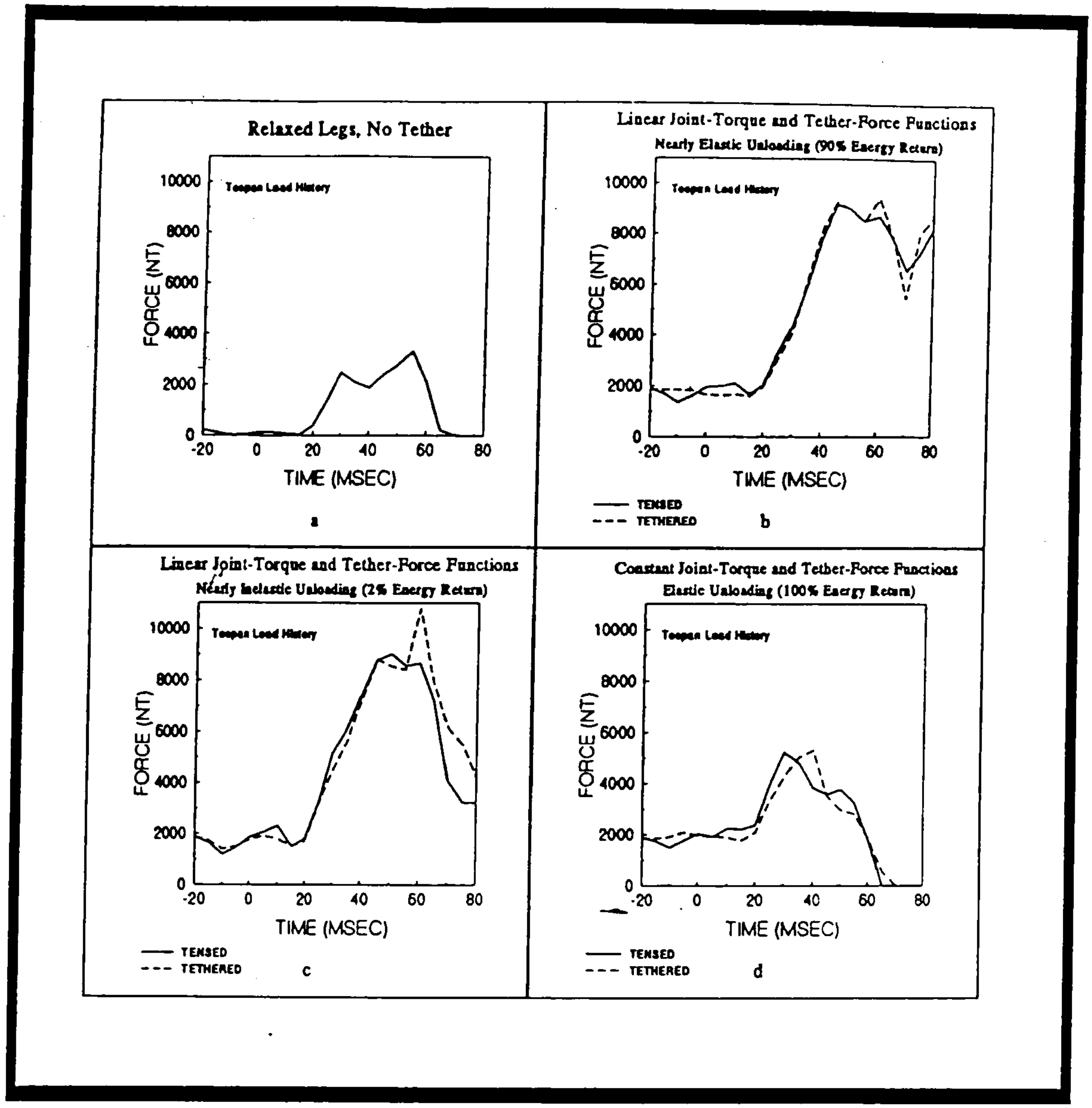


Figure 4. Resultant loads on the foot/toepan for differing load conditions.

The constant joint-torque and tether force functions with elastic unloading produced nearly identical occupant kinetics and kinematics. The resultant loads on the foot/toepan interface for both systems reached approximately 5300 N. The initial pre-load of 1800 N, however, precisely compensated for the difference in peak load for the relaxed condition (3500) and this case (Figure 5). Although occupant excursion for the constant joint and tether conditions was less than the relaxed muscle case, the occupant moved forward along the seat and flexed the lower extremities. Comparable to the relaxed case, the feet lost contact with the toepan at 60 ms. into the event.

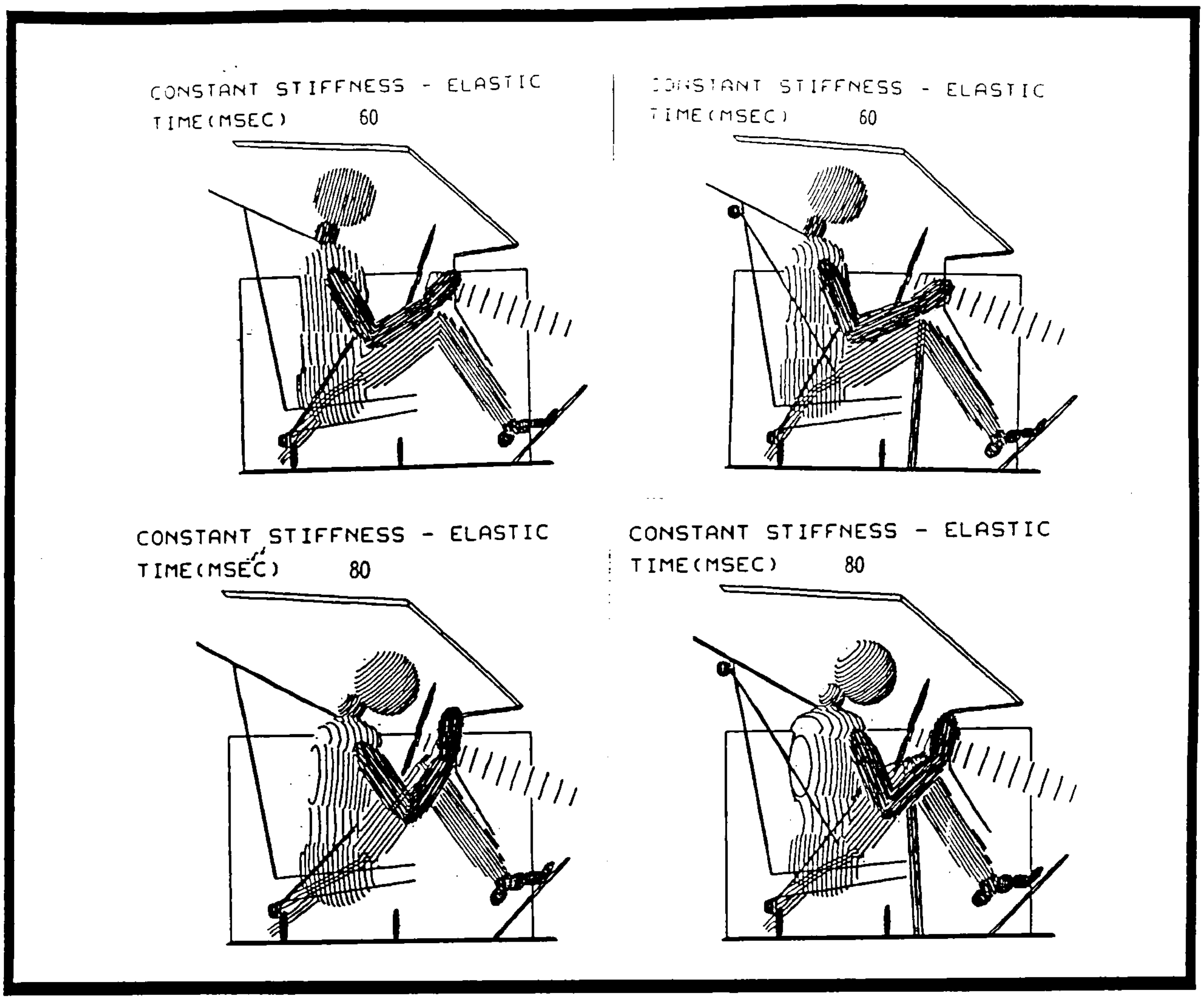


Figure 5. Constant joint torque and tether conditions at 60 ms. and 80 ms.

The linearly increasing joint-torque and tether-force functions with elastic unloading produced identical occupant kinetics and kinematics. The tethers and torques limited forward movement of the occupant and maintained contact of the feet with the toepan. The peak force on the feet against the toepan reached 9000 N (Figure 4). A secondary peak occurred at 75 ms. and indicated that the inertia of the lower extremities attempted to separate the foot from the toepan but was resisted by either the joint torque or tether. Figure 6 demonstrates that significant rotation of the ankle about the anterior posterior axes (i.e., inversion/eversion) occurred despite only longitudinal displacement of the intruding toepan.

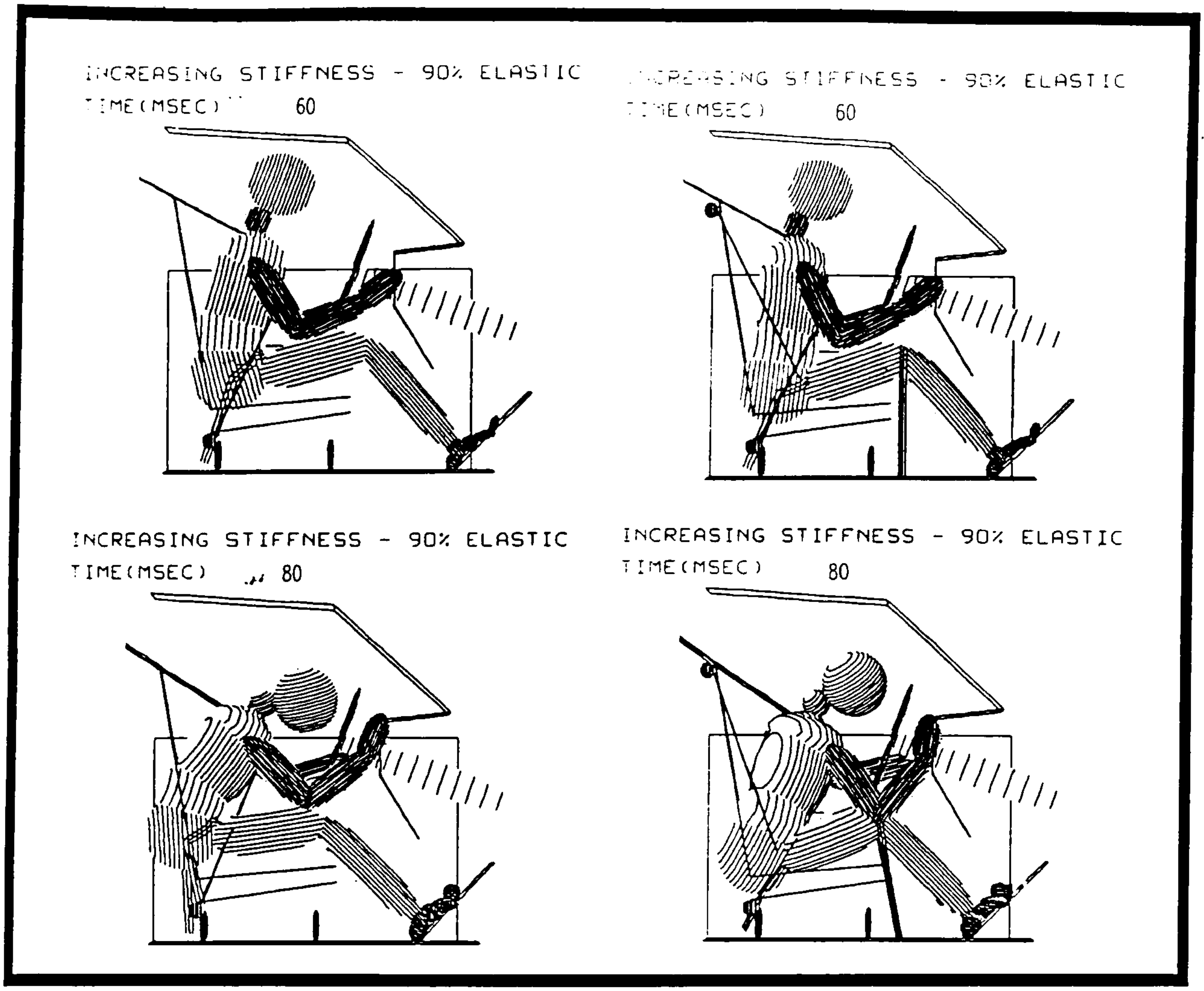


Figure 6. Linearly increasing joint torque and tether conditions with elastic unloading at 60 ms. and 80 ms.

The linearly increasing joint-torque and tether-force cases with inelastic unloading resulted in slightly different peak loads but comparable general shapes for the toepan load-histories. The joint-torque condition resulted in a peak force of 9000 N whereas the peak for the tethered condition was nearly 11,000 N (Figure 4). In both cases, the foot remained in contact with the toepan during the entire intrusion event. Similar to the elastic unloading cases, the increased joint torques and tether forces limited forward excursion of the occupant and flexion of the lower extremities (Figure 7).

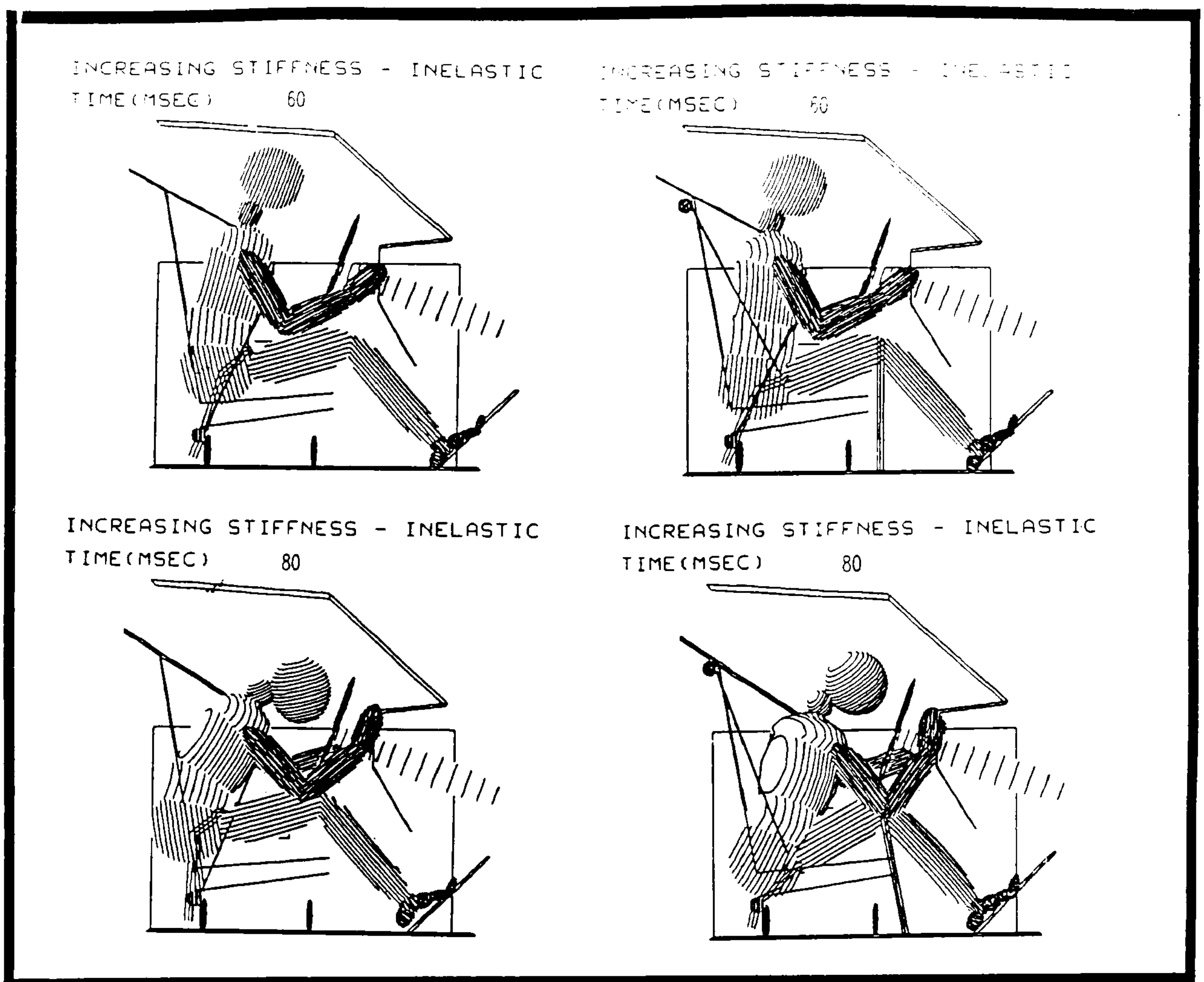


Figure 7. Linearly increasing joint torque and tether conditions with inelastic unloading at 60 ms. and 80 ms.

Conclusions

This paper has demonstrated the significant influence that muscle tensing can have on the kinetics and kinematics of a vehicle occupant during impact. Although preload levels for the lower extremities have been estimated from volunteer tests, there is little agreement on how tensed muscles respond to additional loading. The simulation model developed for this research proved to be an effective means to study the effects of various active and passive muscular responses. In particular, the four conditions (relaxed, constant, increasing resistance with elastic unloading, increasing resistance with inelastic unloading) identified significantly different loading behavior.

The simulation model suggests that a human surrogate outfitted with the tether system can successfully simulate the tensed occupant. The tether system demonstrated excellent qualitative agreement with the tensed condition achieved by joint torque functions. Although validation of the tether system must be performed through sled testing, the model proved to be an effective tool in designing a simple system to simulate muscle tensing of the lower extremities.

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DISCUSSION

PAPER: **The Influence of Muscle Tension on Lower Extremity Response**

SPEAKER: Jeff Crandall, University of Virginia

QUESTION: Don Friedman, Liability Research.

Have you thought about how to implement a mitigation procedure involving reduction of leg, ankle injuries?

A: I think we are approaching it from sort of the opposite end of the spectrum. What we're really trying to do is find out what causes the injury. I think that it's an uncertainty at this point. What we would like to have a better idea for is what's actually causing the high number of foot and ankle injuries that are being shown in the NASS data, and that we would like to work on the mitigation from that. I don't think we have the knowledge at this point to go directly and work on a mitigation mechanism.