AXIAL IMPACT CHARACTERISTICS OF DUMMY AND CADAVER LOWER LIMBS

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ABSTRACT

Axial impact tests conducted at the Medical College of Wisconsin on isolated cadaver lower limbs and the original version of the 50[th] percentile Hybrid III dummy lower limb were examined to characterize their dynamic response. Unlike the more recent version of the Hybrid III dummy lower limb, the original Hybrid III dummy lower limb allows only 30 degrees of foot rotation in dorsiflexion and lacks biofidelic heel pad compression characteristics. This original version of the Hybrid III dummy lower limb will henceforth be referred to as HIII-o dummy lower limb. Results of the tests suggested that the axial force measured in the HIII-o lower limb was higher than that measured in the cadaver lower limb for similar impacts applied to the plantar surface of the foot.

The dynamic properties of the cadaver and HIII-o dummy lower limb were characterized by representing the lower limb in the axial impact tests as a single degree of freedom system with a Kelvin element having linear stiffness and damping properties. The stiffness and damping coefficients of the cadaver and HIII-o dummy lower limbs were obtained from linear regression using the measured accelerations on the pendulum and the leg as input and output of the system, respectively. The average stiffness and damping coefficients were estimated to be 963 kN/m and 0.21 for the cadaver and 3256 kN/m and 0.26 for the HIII-o dummy lower limbs, respectively.

The axial force response ratio between the HIII-o dummy and cadaver lower limbs under similar impact conditions was obtained using Runge-Kutta simulations of the equation of motion for the single degree of freedom system. The axial force response ratio was represented as a function of the rise time of the axial force (the time to maximum force) in the HIII-o dummy lower limb. The axial force ratio between HIII-o dummy and cadaver is greater than one for HIII-o dummy axial force rise times below 55 msec (short duration tibia force pulse). The axial force in the HIII-o dummy and cadaver are approximately the same for long duration HIII-o dummy axial force pulse.

INTRODUCTION

Recent research efforts have attributed lower limb injuries to axial loading through the plantar surface of the foot. In an epidemiological study using data from a Level I Trauma Center, Dischinger et al. (1994) and Crandall et al. (1996) noted that axial load through the plantar surface of the foot contributed to 47% of the ankle fractures sustained in frontal automobile crashes. In another study, Ziedler et al. (1981) noted that distal tibia and fibula fractures were caused by axial compression alone or by a combination of compression, torsion, bending, and tension. Injury criteria for the foot and ankle complex have recently been developed based on the contact axial force at the plantar surface of the foot (Klopp et al., 1997) and axial force in the proximal tibia (Yoganandan et al., 1996). These criteria were developed using human cadaver lower limbs.

In order to evaluate countermeasures, these injury criteria must be incorporated into testing with anthropomorphic test devices and computational models.

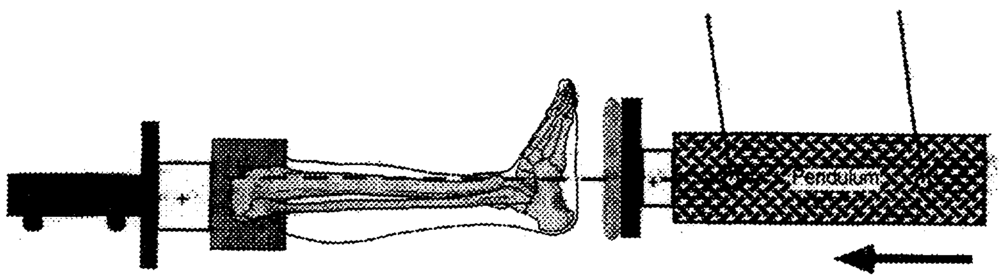

Figure 1. Setup for the axial impact tests at the Medical College of Wisconsin.

However, before applying any lower limb injury criteria to the test device, physical properties and response differences between the test device and human cadaver lower limbs need to be addressed. In particular, the mass, stiffness, and damping properties of the test device should be similar to that of the human lower limb such that the forces measured in the test device and human leg are similar under similar impact conditions.

This paper presents the research efforts concerned with the application of a lower limb injury criteria based on axial force, to the original 50th percentile Hybrid III dummy lower limb. Unlike the more recent version of the Hybrid III dummy lower limb, the original Hybrid III dummy lower limb allows only 30 degrees of foot rotation in dorsiflexion and lacks biofidelic heel pad compression characteristics. This original version of the Hybrid III dummy lower limb will be referred to in this paper as HIII-o dummy lower limb. The differences in the dynamic properties and the physical responses between the HIII-o and human cadaver lower limbs were examined. The first objective in this research effort was to determine the dynamic stiffness and damping properties of the human cadaver and HIII-o dummy lower limb. The second objective was to characterize the axial force response ratio between HIII-o dummy and cadaver lower limbs.

In order to achieve these objectives, the axial impact tests to the lower limbs of cadavers and HIII-o dummy, conducted at the Medical College of Wisconsin (MCW) were examined (Yoganandan et al., 1996).

TEST SETUP AT THE MEDICAL COLLEGE OF WISCONSIN

The test apparatus (Figure 1) consisted of a pendulum and leg specimen, attached to a mini-sled. The mini-sled was free to move on linear ball bearings over precision ground stainless steel rails after the impact. The pendulum, mass of 25 kg, impacted the plantar surface of the foot with velocities ranging from 2.2 m/s to 5.6 m/s. The human cadaver leg specimens and the HIII-o dummy leg were disarticulated at the knee and attached to the mini-sled. The mini-sled and leg assembly was ballasted to 16.8 kg in order to simulate the mass of the whole lower limb. Load cells and accelerometers were attached to the pendulum and the mini-sled system in order to measure accelerations and forces at the plantar surface of the foot and the proximal leg. The contact surface of the pendulum was padded with a one inch thick synthetic rubber (E.A.R. Composites Isodamp C-1000). Details of the test setup are provided in (Yoganandan et al., 1996).

PHYSICAL PROPERTIES AND RESPONSE DIFFERENCES BETWEEN THE HIII-o DUMMY AND HUMAN CADAVER LOWER LIMBS

The average mass of the human lower limb was obtained from a summary of cadaver segment mass data (Crandall, 1996 and NASA, 1978). The average mass of the HIII-o dummy lower limb was obtained from dummy specifications. The mass of various segments of the lower extremity are presented in Table 1. The total mass of the HIII-o dummy lower extremity (11.2 kg) is similar to the mass of the human cadaver lower extremity (11.5 kg).

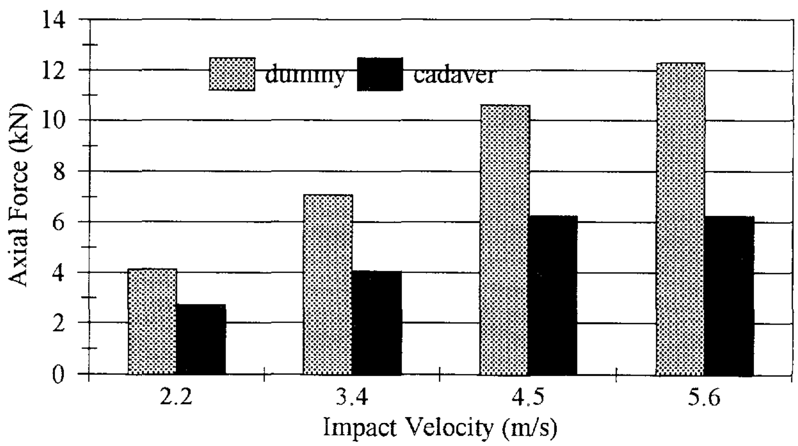

Figure 2. Cadaver and HIII-o dummy responses under similar impact conditions.

Table 1.
Mass of Segments of the Lower Extremity.

Segment	50% HIII-o Dummy (kg)	50% Human Cadaver (kg)
Thigh	6.0	7.3
Leg	3.7	3.2
Foot	1.5	1.0
Total	11.2	11.5

In order to examine response differences, the HIII-o dummy and approximately 50 percentile male cadaver lower limbs were impacted under identical conditions in the MCW test setup. Only those cadaver tests with no injury were considered. In each paired test, using HIII-o and human cadaver lower limbs, the pendulum impact velocity and impact energy was maintained the same. For these tests, the HIII-o dummy leg consistently experienced higher force (one and a half to two times higher) at the proximal end than the cadaver leg (Figure 2).

Since the mass of the HIII-o dummy and cadaver lower limbs are similar, the results in Figure 2 imply that the dynamic stiffness and damping properties of the dummy and cadaver lower limbs differ. Due to the large response differences between cadaver and dummy lower limbs, the injury criteria, developed using human cadaver lower limbs, would be conservative when applied to the HIII-o dummy in this impact condition. In order to determine the conditions under which the axial force injury criteria would be conservative when applied to the HIII-o dummy lower limb, the first step was to characterize the dynamic properties of the HIII-o dummy and cadaver lower limbs

CHARACTERIZING THE DYNAMIC PROPERTIES OF THE CADAVER AND HIII-o DUMMY LOWER LIMB

The stiffness and damping coefficients of the human cadaver and the HIII-o dummy lower limbs were obtained by analyzing the MCW test data. The dummy and human cadaver lower limbs were represented by a single degree of freedom system with a Kelvin element (Figure 3). The acceleration measured at the impactor surface of the pendulum was taken as input into the system. The acceleration measured at the proximal end of the leg was taken as the output of the system.

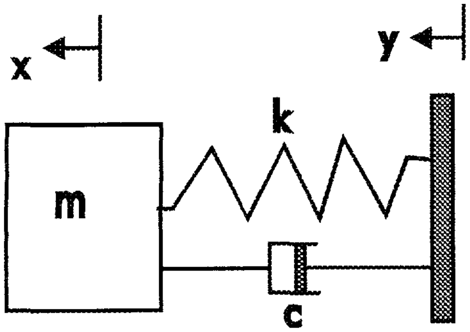

Figure 3. Single degree of freedom representation of the lower limb.

The development of a single degree of freedom linear model required several simplifying assumption which are listed below:

1. The first assumption is that the stiffness and damping coefficient are linear. During the initial phase of impact, the foot penetrates the padding material producing a growing contact area. This produces a nonlinear force versus deformation response. In order to ensure a linear stiffness and damping coefficient, analysis of the data was considered only in the range between 5% of the peak value on the leading edge of the input pulse and 10% of the peak value on the trailing edge of the input acceleration pulse. Analysis of the MCW data suggested that within this range of data, linear approximation was reasonable.
2. The second assumption is that there is no significant foot rotation during the impact. Foot rotation would also produce a nonlinear response for axial measures. Hence, only those tests were considered from the Medical College of Wisconsin where the impact was along the distal one-third of the tibia to minimize foot rotation. Film analysis of these MCW tests showed little foot rotation.
3. The third assumption is that there is no loss of contact between the foot and the impacting surface during the impact event. This is a critical assumption of linearity since any loss of contact would be expected to produce a strongly nonlinear response. This means, that the data analysis is restricted to the duration of the compression pulse measured by the pendulum load cell. This assumption can also be satisfied by considering the signals only between the range of 5% of the peak value on the leading edge and 10% of the peak value on the trailing edge.
4. The fourth and last assumption is that the stiffness and damping properties of the system are constant during the compression phase. Any injury or fracture in a cadaver specimen during the impact, would change its stiffness and damping properties. Hence, only those cadaver tests were considered in the analysis which did not sustain any injury.

For the single degree of freedom system in Figure 3, m represents the mass of the leg and mini-sled assembly (16.8 kg). The equivalent stiffness of the lower limb and padding on the pendulum impact surface is represented by k. Damping of the system is represented by c.

If $\ddot{y}$, $\dot{y}$, and y are the acceleration, velocity and displacement of the impacting surface, respectively, and $\ddot{x}$, $\dot{x}$, and x are the acceleration, velocity, and displacement at the proximal end of the leg, then the equation of motion for the system in Figure 3 is obtained from dynamic equilibrium for the lumped mass where

$$m\ddot{x} + c(\dot{x} - \dot{y}) + k(x - y) = 0 \qquad (1)$$

$$Let \quad z = x - y$$

$$\rightarrow \quad m\ddot{z} + c\dot{z} + kz = -m\ddot{y} \qquad (2)$$

$$or \quad \ddot{z} + 2\xi\omega\dot{z} + \omega^2 z = -\ddot{y} \qquad (3)$$

rearranging,

$$-\ddot{x} = 2\xi\omega\dot{z} + \omega^2 z \qquad (4)$$

The natural frequency of the system, ω, is √(k/m) and the damping coefficient, ξ, is c/2mω.

The Equation 4 is a linear relation in z and $\dot{z}$ where these values are known for all corresponding $\ddot{x}$. This relation can be generalized for the measured data

$$-\ddot{x}_i = b_o + b_1\dot{z}_i + b_2 z_i + \varepsilon_i \qquad (5)$$

$$where \quad b_1 = 2\xi\omega \quad and \quad b_2 = \omega^2$$

where $\ddot{x}_i, \dot{z}_i, z_i$ are the acceleration, relative velocity, and relative displacement of the system at the ith instant, b_o is a bias or offset value, and ε_i is a perturbation. The bias can be attributed to systematic errors in the measures. The perturbation is due in part to the precision of the instruments and to the fact that the true system response is not completely described by a single degree of freedom linear model. If one assumes that the perturbation is a random variable that is normally distributed, then an optimal line or line of best fit to the test data under consideration can be obtained by using multivariate linear regression. This method produces least squares estimates for the parameters b_0, b_1, b_2. The estimates of ω and ξ are then obtained from Equation 5. The stiffness and damping values are obtained from $k=\omega^2 m$ and $c=2\omega m\xi$. The mass considered is the mass of the leg and sled assembly (16.8 kg) which represents the total mass of the lower limb.

RESULTS

This method of linear regression was applied to each cadaver and dummy test. The errors in the estimates of b_1 and b_2 for each test were small (within 10%) and the R^2 value of the analysis was high (cadaver average $R^2=0.9$ and dummy average $R^2=0.98$) suggesting that the values of ω and ξ are reasonably constant during the compression phase of the impact. This implies that the assumptions of linearity and constant dynamic properties during the compression phase were reasonable. The details of the tests and the results of the regression analysis are presented in Table 2.

Since the dynamic properties of the cadaver lower limb are to be compared to the 50th percentile HIII-o dummy lower limb, approximately 50 percentile male cadaver specimens were used in the study. The average estimated stiffness and damping coefficient for the cadaver lower limb is 963 kN/m and 0.21, respectively. The average stiffness and damping coefficient for the HIII-o dummy lower limb is 3256 kN/m and 0.26, respectively.

The results obtained are consistent with previous studies showing the HIII-o dummy leg to be stiffer than the comparable size human leg (Crandall, et al., 1996). Mizrahi and Susak (1982) investigated the in-vitro transmission of impact forces from the foot through the entire straight leg in human volunteers. A two degree of freedom linear mathematical model was employed to describe the mechanical behavior of the leg and the rest of the body during free fall from a height of 5 cm above ground. The volunteer legs were straight and vertical when the plantar surface of the foot impacted the floor causing axial loading of the lower limb. Assuming linear stiffness and damping properties for the leg, Mizrahi and Susak estimated the stiffness for the human leg to be 340 kN/m. The value of stiffness is lower than 963 kN/m obtained in the present study. The difference could be attributed to differences in the testing protocol and the considerably lower impact force levels (maximum impact force = 1700N and maximum acceleration =5 g's) in the Mizrahi and Susak study. Mizrahi and Susak did not control for the assumption of linearity in their model. The stiffness obtained from the Mizrahi model is lower than that in the present study since the Mirzrahi-Susak model includes the compression of the heel pad which is highly nonlinear and of lower stiffness than the rest of the leg.

Having completed the characterization of the dynamic properties of the lower limbs of the human cadaver and dummy, the next step was to characterize the response ratio between the forces measured in the dummy leg to that measured in the cadaver leg.

CHARACTERIZING THE AXIAL FORCE RESPONSE RATIO BETWEEN HIII-o DUMMY AND CADAVER LOWER LIMBS

The single degree of freedom representation of the lower limb (Figure 3) with the estimated average stiffness and damping properties for the HIII-o dummy and cadaver lower limbs was utilized to investigate the axial force response ratio between the cadaver and dummy lower limb. The input acceleration to the system was represented as a half sine wave, $A \sin \varpi t$. This representation of the input acceleration is similar to the characterization of floor pan acceleration in vehicle-to-vehicle crashes by Kuppa et al. (1995).

In order to illustrate the influence of the various dynamic parameters on the force in the lower limb, consider the motion of the system to be represented by Equation 1. For simplicity, ignore the effect of damping. The stiffness of the system is given by k. The axial force in the lower limb is given by F=k(x-y)=kz, which is

$$F = \frac{mA}{(1-\beta^2)}(\sin \varpi t - \beta \sin \omega t) \qquad (6)$$

$$where \qquad \beta = \frac{\varpi}{\omega}$$

where ω is the natural frequency of the system ($=\sqrt{(k/m)}$), m is the mass of the dummy lower limb, and ϖ is the frequency of the input acceleration (Clough, R.W. and Penzien, J.,1975).

The force in the lower limb is not only a function of the dynamic properties of the lower limb but also a function of the frequency of the input acceleration. The force F_c and F_d in the human cadaver and HIII-o dummy lower limb can be determined using Equation 6. Since the mass of the dummy lower limb (m_d) and cadaver lower limb (m_c) are similar, the ratio of the axial force in the HIII-o dummy to the axial force in the human cadaver lower limb, for the same input acceleration of $A \sin \varpi t$, is given by Equation 7.

$$\frac{F_d}{F_c} = \frac{(\sin \varpi t - \beta_d \sin \omega_d t)(1 - \beta_c^2)}{(\sin \varpi t - \beta_c \sin \omega_c t)(1 - \beta_d^2)} \tag{7}$$

$$\text{where} \quad \beta_c = \frac{\varpi}{\omega_c} \qquad \omega_c^2 = \frac{k_c}{m_c}$$

$$\text{and} \quad \beta_d = \frac{\varpi}{\omega_d} \qquad \omega_d^2 = \frac{k_d}{m_d}$$

k_c and k_d are the stiffness of the cadaver and HIII-o dummy lower limbs, respectively. Equation 7 suggests that the ratio of the force in the HIII-o dummy lower limb to that in the cadaver lower limb is a function of the stiffness of the dummy and cadaver lower limbs and the frequency of the input acceleration.

In order to solve the complete equation of motion which includes the effect of damping (Equation 3), a Runge Kutta simulation was conducted. using the dynamic properties of the human cadaver and the HIII-o dummy. Simulations were conducted with different input acceleration frequencies (ϖ). The input acceleration frequency was computed as $\varpi=\pi/$(pulse width). The force in the dummy and cadaver lower limbs were computed as $F = c\dot{z} + kz$. The ratio of the force in the dummy to that in the cadaver for different impact frequencies is shown in Figure 4.

The axial force response ratio between dummy and cadaver is greater than one for input acceleration frequency higher than 40 rad/s (short duration pulse). The maximum response ratio is approximately 1.86 for input acceleration frequencies greater than 200 rad/s. For input acceleration frequency between 7.85 and 40 rad/s, the response ratio dips below 1 to as low as 0.77 at an input acceleration frequency of 15.7 rad/s. The force ratio is approximately 1 for input acceleration frequencies below 7.85 rad/s (very long duration input acceleration pulse).

The frequency of the input acceleration is not always easy to estimate and so may not be the best parameter to determine the axial force response ratio (F_d/F_c) for a given impact condition. However, the time at which maximum axial force occurs (rise time) in the HIII-o dummy leg is unique for a given floor pan acceleration frequency as shown in Figure 5.

Recognizing this relationship between the floor pan acceleration frequency, ϖ, and rise time (T_{rise}) of axial force in the HIII-o dummy leg, the ratio, F_d/F_c for a given impact condition can be determined by the rise time for axial force in the dummy leg (Figure 6). The ratio, F_d/F_c, is greater than one for T_{rise} smaller than 55 msec. This ratio dips below one for rise times between 55 and 200 msec. The force response ratio is approximately one for T_{rise} greater than 200 msec.

The conservative approach to the lower extremity injury criteria using axial force assumes the ratio F_d/F_c =1. This implies that the axial force measured in the HIII-o dummy leg with small rise time (less than 55 msec) and a peak level greater than the critical force may not be injurious since the force response ratio is greater than one. On the other hand, the axial force in the HIII-o dummy leg with rise time between 55 and 200 msec, may be injurious for measured peak forces slightly below the critical force level.

This complex relationship between measured axial force and injury threshold level suggests the need for an improved design of the dummy lower extremity which has similar dynamic properties as the human lower limb.

CONCLUSIONS

Axial impacts to the plantar surface of the foot of isolated cadaver and HIII-o dummy lower limbs suggested that the forces measured in the HIII-o dummy lower limb is consistently higher than that measured in the cadaver lower limb, under similar impact conditions. These axial impact tests were further examined to characterize the dynamic properties of the HIII-o dummy and cadaver lower limbs and to determine the axial force response ratio between the dummy and cadaver.

The cadaver and HIII-o lower limbs were represented by a single degree of freedom system with a Kelvin element having linear stiffness and damping properties. The stiffness and damping properties were estimated from multivariate linear regression analysis using measured accelerations on the pendulum and the leg as input and output to the system. The average stiffness and damping coefficients were estimated to be 960 kN/m and 0.21 for the cadaver and 3256 kN/m and 0.26 for the HIII-o dummy lower limbs, respectively.

The ratio of the axial force measured in the dummy leg to that in the cadaver leg (F_d/F_c) was estimated using Runge-Kutta simulation of the motion of the single degree of freedom representation of the cadaver and dummy lower limbs. The estimated average stiffness and damping values of the cadaver and dummy lower limbs were used in these simulations. The force response ratio (F_d/F_c) was characterized as a function of the rise time (time to maximum value) (T_{rise}) of the measured axial force in the HIII-o dummy leg.

The force response ratio, F_d/F_c, is greater than one and as high as 1.8 for T_{rise} less than 55 msec. This implies that for small T_{rise} values, the measured axial force in the dummy leg which is above the critical threshold value may not be injurious. On the other hand for T_{rise} between 55 and 200 msec, $F_d/F_c <1$. Therefore the measured axial force in the dummy leg which has a T_{rise} between 55 and 200 msec may be injurious though the maximum force is slightly below the critical force value.

The presence of padding has the effect of increasing the rise time of the measured force. Therefore, the axial force response ratio (F_d/F_c) decreases when padding is added to the system. This would change the relationship between the measured force in the HIII-o dummy and the injury threshold level.

This complex relationship between the force measured in the HIII-o dummy leg and the critical force suggests that there is a need for a lower extremity device which demonstrates comparable axial compliance with the human leg.

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Table 2.
Test Data and Results of the Analysis

Test No.	Age	Sex	Test Velocity (m/s)	freq. ω Rad/s	damp. ratio ξ	stiffness kN/m	max. impactor force (N)	max. tibia force (N)
CADAVER TESTS								
PCLE110-1	27	M	2.2	235	0.19	928	3662	2666
PCLE114-1	46	M	2.2	242	0.15	984	3756	2719
PCLE116-1	27	M	3.4	282	0.22	1336	5919	4490
PCLE117-1	55	M	4.5	277	0.21	1289	7819	6226
PCLE129	66	M	3.4	230	0.22	889	5900	4607
PCLE131	64	M	2.2	206	0.18	713	3564	2776
PCLE132	50	M	3.4	209	0.3	734	4236	3093
PCLE133	50	M	5.6	223	0.24	835	7983	6239
AVERAGE				238	0.21	963		
HIII-o DUMMY TESTS								
PDLE119-1			2.2	356	0.2	2129	5447	4125
PDLE119-2			3.4	440	0.25	3252	9434	7058
PDLE119-3			4.5	498	0.27	4166	13960	10604
PDLE119-4			5.6	455	0.32	3478	17840	12278
AVERAGE				437	0.26	3256		

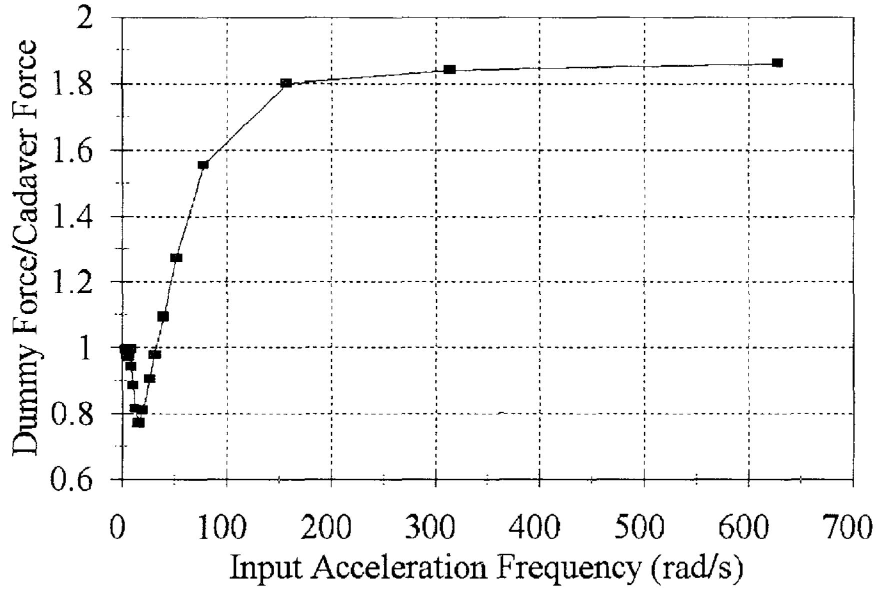

Figure 4. Ratio of HIII-o dummy / cadaver axial force for different input acceleration frequency.

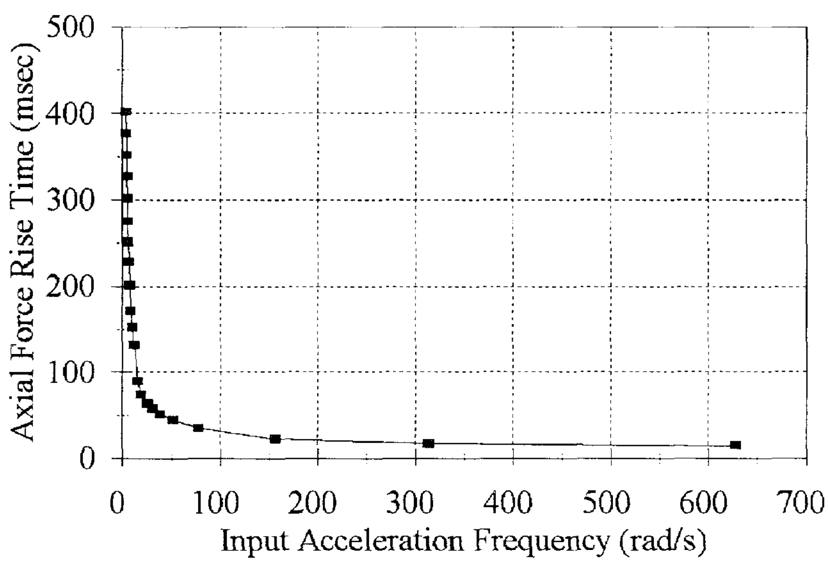

Figure 5. Axial force rise time in HIII-o dummy leg versus input acceleration frequency.

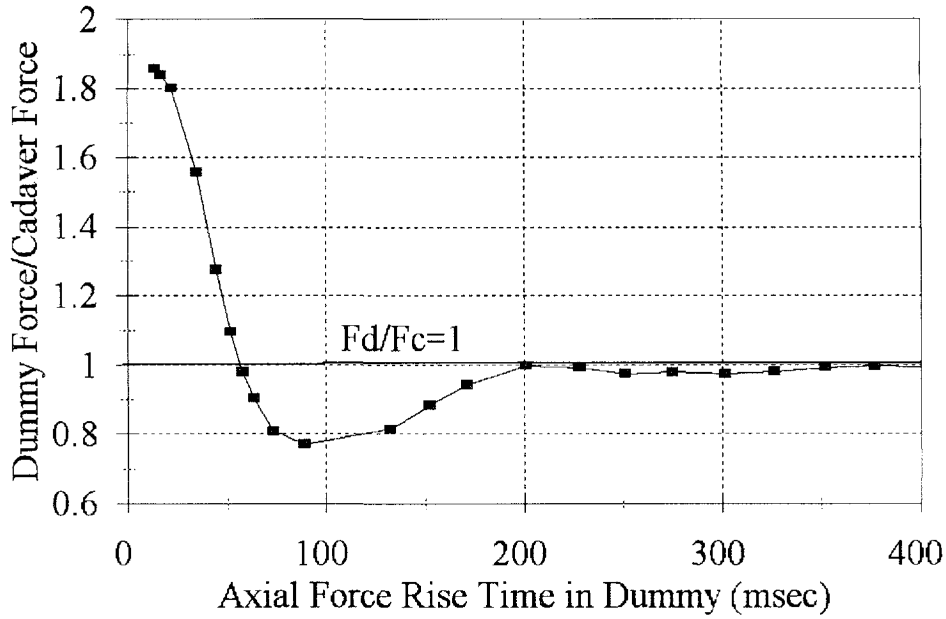

Figure 6. Ratio of HIII-o dummy/cadaver axial force versus axial force rise time in HIII-o dummy.