THE RESPONSE OF THE HUMAN LOWER LEG TO IMPACT LOADING

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INTRODUCTION

It has been widely observed and reported that injury to the leg is one of the most common forms of trauma associated with motorcycle accidents (1). Furthermore, it has also been observed and reported that the majority of motorcycle leg injuries resemble those experienced by pedestrians in that they do not involve crush. Rather, these injuries appear to involve only a direct impact between the leg and an opposing rigid object. Often the soft tissue of the limb is injured from the inside out in that sharp bone fragments and jagged ends lacerate the soft tissue as relative motion occurs. The complexity of understanding these results is due to the combination of impact effects, biological material properties, and human geometric considerations. The study of impact dates back to studies by Galileo in 1638. Research continues today to understand ballistic impact as documented by Werner Goldsmith (2). The biomedical aspects of impact have also received intense attention (3). This is especially true in the study of injuries obtained during automobile accidents (4). The majority of this work has focused on injuries of the head and neck. The work has also emphasized human body dynamics which has resulted in both specialized computer codes and the development of anthropomorphic dummies for testing. The use of these dummies has been limited, however, because of the lack of impact failure simulation. The literature on the determination of human and animal tissue material properties is extensive and well summarized by such individuals as Yamada (5). Although the majority of the work is properties obtained by slow loading, there has been significant work done in impact load by McElhaney (6) and others. The work in this paper differs from previous work in that complete bone and leg samples are used and the impact and failure occur while the velocity of the striking object is constant.

FIRST YEAR RESEARCH

In order to better understand lower leg impact injuries and hopefully to identify specific countermeasures, a laboratory apparatus was designed and constructed to allow impact testing at speeds up to 30 mph with impacting masses of several hundred pounds. Fifteen cadaver legs were then impacted by an automobile bumper. The impact was directed at or slightly above the distal one third of the tibia. Half of the impacts were delivered from the front (a-p) with the remaining delivered laterally. The automobile bumper was rigidly attached to a track guided cart traveling at either approximately 23 or 34 feet per second while the test specimen was supported by a single steel rod passing through the distal condyle of the femur. Upon impact, the specimen was free to rotate about this mounting rod with only its own inertia holding it in initial position.

In these 15 tests employing pinned - inertial specimen support, fractures (the majority comminuted) were obtained 14 times. Furthermore, it was observed that following fracture, the test specimen would literally drape itself around the impactor as the impactor accelerated it to a matching speed. In addition to direct test documentation, these specimens were X-rayed and then carefully dissected with good photographic documentation.

DISCUSSION

A careful review of the first year's data combined with an initial physical analysis suggests that the dynamic response of the human leg to impact must be viewed as a composite based upon the separate reactions of the various tissues as they provide a structural and/or inertial response to the impacting object. For example, in the case of an (a-p) impact delivered to the tibia, the crest of the tibia contacts the impactor after the offending object travels less than one mm (the approximate thickness of the tissue covering the tibia). The compression/extrusion of these covering tissues deliver a considerable force to the tibia. However, except for impacts involving very small velocity differences, this force is not adequate to "push" the limb out of the way before the tibia is reached. Once the impacting object reaches the tibia, these inertial forces rapidly rise to whatever level is necessary to accelerate the limb to a sufficient angular velocity for it to clear the impacting object. During this process, the tibia structurally flexes as it reacts to the inertial forces generated by its own acceleration as well as the acceleration of all surrounding tissue. Of course, if these inertial forces exceed the maximum structural strength of the tibia, fracture occurs. The limb must then be deflected or pushed away from the impactor through tensile responses and plastic deformation or be torn in two. The early failure of the tibia followed by large soft tissue displacement during limb acceleration was consistently seen at these relative speeds.

BONE RESPONSE

Because of the obvious importance of the tibia in determining the dynamic response of the lower leg to impact, tests were conducted in order to identify the ultimate strength and force-deflection characteristics of the human tibia when it is removed from the leg. Three separate series of test conditions were utilized in evaluating the strength of this long bone when simply supported at each end. The first of this series of three was designed to provide reference data concerning the characteristics of the instrumentation system as well as to provide information on the characteristics of this bone with a low speed impact. The second test was intended to reflect only the effect of using a one inch pipe as the impacting object rather than the direct transducer - this was also a low speed impact. The third series was intended to provide information on the effect of impact speed in that the only difference between the second and third series was that the third series was conducted at a target speed of 25 feet per second rather than at 3 to 6 feet per second. The specific conditions of the first series involved the direct impact (a-p) of the force transducer with the tibial bone at approximately midshaft, used 9 specimens, and employed an impact speed of between 2 and 5 feet per second. The second series, also involved 9 specimens as well as an a-p impact, utilized a velocity range of from 3 to 6 feet per second and employed a one inch pipe impactor mounted directly on the force transducer. The last series involved six specimens, utilized the one inch pipe impactor, an a-p impact and an impacting velocity of 25 feet per second. A typical force time history is shown in Figure 1. The results of these three series of tests are summarized below:

	Mean	Standard Deviation
Direct Impact at 2-5 fps	347 lb.	146 lb.
Pipe Impacting at 3-6 fps	680 lb.	145 lb.
Pipe Impacting at 24-26 fps	547 lb.	63 lb.

Although the reason for differences between the direct impact and the pipe impact at low speed is not obvious, the most probable cause involves inadvertent contact occurring between the structure supporting the force transducer and the specimen as well as between the force transducer and the specimen. It is likely that there is no difference in the strength of the specimens under low and high speed impacts because of the large variance in the data and the small sample size.

SIMPLY SUPPORTED LEG

The next activity in this research involved impacting intact legs in a manner analogous to that of the simply supported bone. Here, the intact leg was mounted from a steel rod passing through the condyle of the femur with the heal against a very rigid steel shape. The impact was delivered with the one inch steel pipe at approximately the distal 1/3 of the tibia and with a striking velocity of 25 feet per second. A typical force time history is shown in Figure 2. Based on five specimens, the average peak force was 524 pounds while the standard deviation was 80.5 pounds. It is unlikely that there is a statistically significant difference between the impact strength of the intact leg and that of the bare tibia, at least not at 25 fps. However, our sample size is still quite small.

Review of the force transducer curves as well as of the high speed films provides considerable insight into the role of the soft tissue in maintaining leg integrity during impact but following bone fracture. In essence, the portion of the limb distal of the fracture must be accelerated by means of tensile forces delivered through the soft tissue around the area of the fracture. Unquestionably much of the internal soft tissue damage attendant to this type of impact must be related to this often violent stretching and bending in the immediate proximity of sharp bone fragments and splintered bone shaft.

ANIMAL TISSUE TESTING

Animal tissue testing has been conducted for the specific purpose of evaluating the effect of embalming on the strength of tissue and bone. Although the available goat limbs are clearly much shorter than comparable human long bones, matched legs were obtained from two goats. The two right limbs were embalmed and the two left limbs frozen. Upon testing, the following results were obtained.

	Embalmed	Frozen
Goat One	928	1002
Goat Two	1036	989

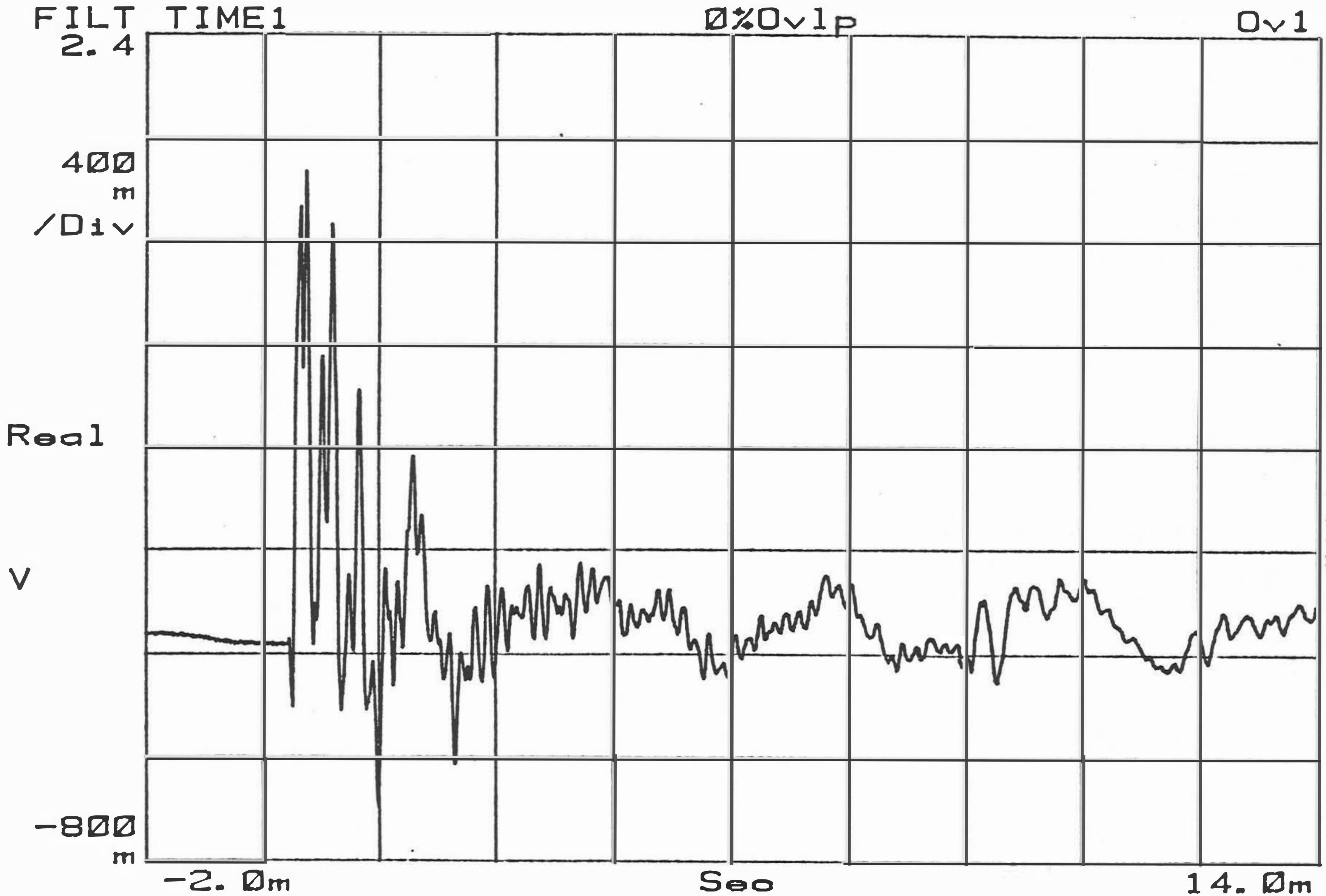

Figure 1: High-Speed Tibial Bone Test
One Inch Pipe 33L 25.60 fps

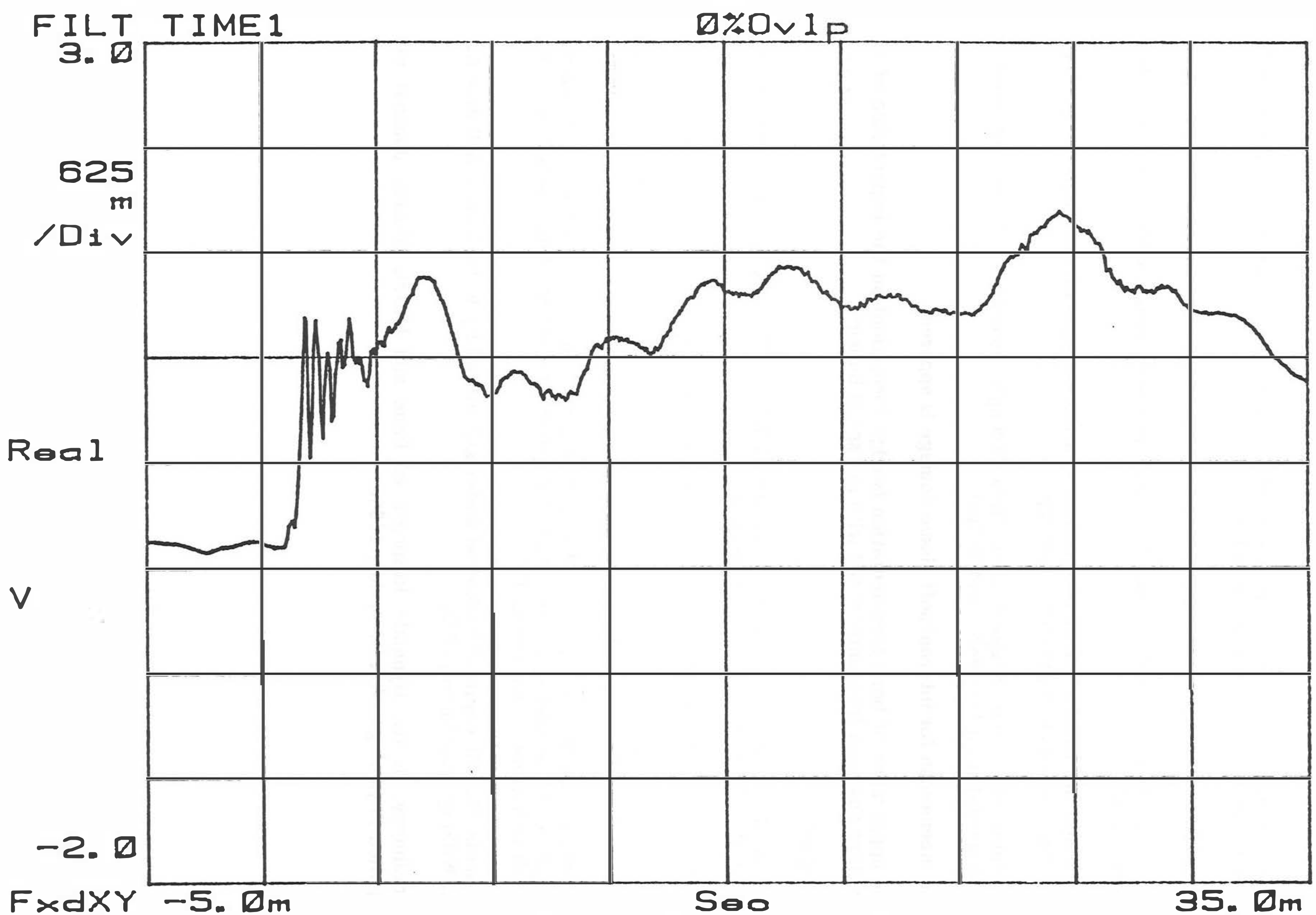

Figure 2: High-Speed Intact Leg Test
Force Data JE 883 25.10 fps

Although the sample size is unacceptably small, it appears that there is no significant difference between the embalmed leg and the frozen leg.

CONCLUSIONS

Although hampered by limited sample sizes, the following conclusions may be tentatively stated:

1) There appears to be little or no difference in the strength of the simply supported tibia and the simply supported intact leg.

2) Available data may suggest that tibia strength decreases as impact speed increases (at least between 5 and 25 fps).

3) The mechanism for internal soft tissue damage is apparent.

4) The appearance of bone fragmentation (wedge formation) on the impact side of a fracture has been frequently noted with low speed impacts.

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