IMPROVED MEASURES OF FOOT AND ANKLE INJURY RISK FROM THE HYBRID III TIBIA

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ABSTRACT

From a regulatory viewpoint, the design, construction and instrumentation of the Hybrid III tibia, and the related measures of injury risk, specified in Directive 96/79/EC, present some difficulties. The paper briefly describes limitations associated with the design of the tibia, the standard instrumentation, and the currently regulated measures of injury risk. Given the anticipated delay before more advanced legs become available, interim means of increasing the utility of current data are suggested. Improvements to the instrumentation and further modification of the ankle characteristics are briefly discussed.

INTRODUCTION

Vehicle performance measures predictive of occupant injury risk, and their associated regulated limits, should provide practicable levels of protection at reasonable cost. The present paper suggests that the regulated performance requirements of Directive 96/79/EC, intended to reduce injuries to the lower leg, ankle and foot, are of limited relevance and effectiveness.

Research on the biomechanics of motor vehicle injuries necessarily precedes the improvement of ATDs. The Hybrid III was originally developed more than 20 years ago, when injuries to the lower extremities attracted much less attention than they do to-day. Not surprisingly, the design and instrumentation of the Hybrid III tibia do not reflect the understanding of lower extremity injuries that now exists. The geometry of the Hybrid III tibia also incorporates features which are absent (or at best, very much less pronounced) in human anatomy, yet significantly affect the data the tibia provides. Moreover, only one of the regulated measures of injury risk has a clear and direct association with the types of lower extremity injury observed in real collisions.

Concern for the effectiveness of the regulation extends beyond the shores of Europe; Australia has already adopted it and work is in progress in the United States to produce a regulation harmonised with the European requirement. Canada is also proposing use of the test procedure, as a complement to the CMVSS 208 test. Pre-production versions of an improved frontal impact ATD are currently being evaluated by various organisations but it is likely to be some time before a new device is certified for regulatory use. New and modified feet, intended to address some of the limitations of the existing Hybrid III hardware, are available, but they are not without significant problems.

This paper therefore has two major purposes. The first is simply to document specific limitations, inherent to the existing regulation, that result from the geometry of the current Hybrid III tibia, from the formulation of the Tibia Index and in the application of the limiting value of the Index. The second purpose is to suggest ways in which the data currently available from the Hybrid III tibia could be used to provide more informative measures of the risk of injury to the lower leg, ankle and foot.

INJURIES TO THE LOWER LEG, ANKLE AND FOOT

Overview of Data from Motor Vehicle Collisions

Morgan *et al.*, (1991) reviewed 480 cases of occupant foot and ankle injury, of AIS 2 or greater severity in frontal impacts. The data were obtained from the hard-copy files of the National Accident Sampling System for the period 1979-1987. Of those cases, 28 percent had only foot injuries, 65 percent had only ankle injuries and 7 percent had both. The authors identified six injury mechanisms, each defined in terms of a specific interaction between the ankle or foot and the vehicle interior.

Of the occupants with <u>ankle injuries</u>, the mechanism could not be determined in 12 percent of cases. In 43 percent of cases, the mechanism was identified as contact with foot controls (drivers only), in 24 percent it was contact with the floor (half drivers and half passengers) and in 12 percent, entrapment of the lower leg between floor and instrument panel (7:5 drivers:

passengers). In slightly more than half of the cases, the specific injury was described. Fracture of the lateral or medial malleolus was the most common, followed by fracture of the talus and then of the distal tibia or fibula.

Corresponding figures for occupants with foot injuries were 8 percent undetermined, 47 percent (drivers only), foot controls, 24 percent, contact with floor (half drivers and half passengers) and 8 percent wheel well intrusion (7:1, drivers:passengers). In three-quarters of the cases, the nature of the injury was described. Fractures of the metatarsals accounted for half of all cases, with fractures of the calcaneus accounting for 15 percent and a further 7-8 percent comprising fractures of the cuboid or cuneiforms.

Lestina *et al.*, (1992) analysed data from a clinical sample of 23 drivers who suffered a total of 25 foot or ankle injuries in frontal crashes. They classified each case according to the scheme proposed by Morgan *et al.* (1991) for defining injury mechanisms. However, in terms of the biomechanical injury mechanism, Lestina *et al.* concluded that 12 of the 13 cases that resulted in malleolar fractures were attributable to inversion or eversion of the foot, not dorsiflexion.

In a more recent paper, Parenteau *et al.,* (1996) presented an analysis of 805 cases of AIS 2 or 3 injuries to the foot or ankle, resulting predominantly from frontal collisions of passenger cars. The objective of the study was to determine the influence of impact location, occupant seating position and occupant age on the frequency, incidence and rate of foot-ankle injury. Frontal impacts accounted for 76.3 per cent of the foot-ankle injuries analysed.

An interesting conclusion, relevant to the test procedure of Directive 96/79/EC, was that near-side oblique collisions, i.e., frontal impacts from the 11 o'clock direction for the driver or the 1 o'clock direction for the front passenger, were about 50 percent more likely to result in AIS 2-3 injuries to foot or ankle than simple frontal impacts. The most commonly occurring injuries were fractures of the ankle, (including the distal tibia), ankle sprains and mid-tarsal fractures. The conclusion should however be viewed with some caution, since the analysis did not account independently for the effects of impact direction and impact location.

From their own work, and from a review of prior research, Parenteau and collaborators concluded that both intrusion and vehicle deceleration contribute to foot and ankle injuries but that the exact mechanism of injury in any particular case is usually unclear.

Biomechanical Tolerance Data

Axial force in the tibia

Using data reported by Yamada (1970), Mertz (1984) proposed a value of 8 kN for the maximum tolerable axial compressive load in the 50th-percentile male tibia.

Yogandan *et al.*, (1996) tested 26 lower legs, separated at the knee joint, under impacts to the plantar surface of the foot. The proximal tibia was fixed in polymethyl-methacrylate and mounted on a small sled ballasted to 16 kg. The pendulum impactor was faced with synthetic rubber and aligned to achieve as nearly axial loading of the specimens as possible. The results of those tests were combined with those of 26 others, using somewhat different procedures, at two other laboratories. Some specimens that did not initially fracture on impact were subjected to one or more subsequent impacts.

The combined data were represented by a 2-parameter Weibull distribution with a function of age and impact force as the variate. However, disregarding age and gender, an axial force of 6.8 kN was associated with a 50 percent probability of fracture for the combined sample of 52 specimens. It should be noted that the subjects were predominantly middle-aged and elderly males. The extremes of the range of tolerance were defined by a specimen from a 27 year-old male which experienced 10.2 kN without fracture and the specimen from a 67 year-old female that fractured at 4.6 kN. Several types of fracture of the distal tibia-calcaneus complex were observed but their frequencies were not reported.

Bending of the tibia

Mertz (1984), again citing data from Yamada (1970), and Nyquist (1985) have reported estimates of the average strength of the tibia in symmetrical, three-point bending ranging from 225 to 320 Nm. The strength of the tibia at mid-shaft was reported by Nyquist to be essentially the same in anteroposterior and lateromedial directions.

Combined compression and bending of the tibia

The Tibia Index (Mertz, 1984) cited in Directive 96/79/EC appears to follow conventional engineering practice in estimating the failing strength of a column under combined compression and bending. However, as several others, including Tarrière and Viano (1995), have noted, the Index does not properly consider the combined effects of the two types of loading. It is a straightforward matter to improve the formulation of the Index, by taking account of the difference between the tensile and compressive strengths of tibial bone (though without considering either the non-linearity or the strain-rate dependence of the strength of bone). For completeness, Appendix A provides such a formulation.

In practice, however, the improved formulation is of limited significance, since the 8 kN limit on axial force severely restricts the range of loading conditions over which the difference in the tensile and compressive strengths of bone might otherwise be significant. Appendix A also shows the effects of the arbitrary increase in the maximum permitted value of the Tibia Index from 1 to 1.3.

A more fundamental issue is the relevance of the upper and lower Tibia Indices to the types of injury commonly observed in motor vehicle collisions. While moderate upper tibia x-axis moments do occur, typically if the lower leg is trapped, the predominant reason for the occurrence of large y-axis moments (and excessive values of the Tibia Index) is the unusual geometry of the tibia in the x-z plane, as explained in Appendix B. Real y-axis bending moments are necessarily difficult to generate in the vicinity of the pin joint at the knee clevis.

At the lower transducer, the Tibia Index is only slightly affected by the unusual geometry of the tibia, the effect of which can readily be discounted, as discussed in Appendix B. However, the regulated level of 225 Nm greatly exceeds tolerable moments in either flexion or inversion/eversion. While in the longer term, some form of index might evolve to cover combinations of flexion with inversion/eversion, the current Tibia Index is, for the present, more appropriately replaced by individual limits on each of the four modes of displacement of the foot at the ankle.

In summary, it is clear from the foregoing that the Tibia Index and its associated limits are of limited relevance to the types of lower leg, ankle and foot injuries observed in motor vehicle collisions.

Displacements of the foot

As indicated above, significant injuries to the ankle and foot are associated with inversion, eversion, dorsiflexion and, less commonly, plantar flexion of the foot. Inversion is often associated with fracture of the medial malleolus, while eversion may result in fracture of the lateral malleolus. The same mechanisms may cause injury to the musculature and ligaments of the foot and ankle, whether or not fracture of either of the malleoli occurs. Dorsiflexion of the foot, induced by dynamic loading at the distal ends of the metatarsals, may result in fractures of those bones, as well as damage to the ligaments.

Parenteau and collaborators (1995) have provided some biomechanical data on which tentative tolerance levels for these modes of injury might be based. The data were obtained from quasi-static loading induced by rotation of the calcaneus in the appropriate plane and sense. Crandall et al. (1996) provided similar but rather more detailed data from volunteer subjects and cadaver specimens, though plantar flexion data were not included. The ranges of tolerable moments for inversion were generally quite similar. However, Parenteau's data were appreciably higher than Crandall's for eversion, while for dorsiflexion, Parenteau's cadaver data fell appreciably below the levels tolerated by Crandall's volunteer subjects.

The following table gives tentative tolerance levels, derived from these two sources, for the four independent modes of angular displacement of the ankle. Except in the case of plantar flexion, the values are based on Crandall's data for volunteer subjects. The angular displacement corresponding with the specified maximum moment is also given. In the particular case of dorsiflexion, the tolerable levels depend on the angle of flexion of the knee. The tolerable levels in the table therefore represent the average of results reported by Crandall (1996) for volunteers at zero and 90° of knee flexion. In the absence of actual data from volunteer subjects, the values shown for plantar flexion are estimates, based on the cadaver data, of the moment and associated displacement that might be tolerated by an average volunteer subject.

Table 1.

Tentative Tolerance Levels for Ankle Injuries

Mode	Tolerable moment (Nm)	Angular displacement (°)
Inversion	16	50
Eversion	40	40
Dorsiflexion	60	35
Plantar flexion	30	40

The subjects were young males with average weight and height approximating those of the current 50th-percentile Hybrid III ATD. For purposes of comparisons between the tolerance levels and crash test data, the data in Table 1 are therefore assumed to represent 50th-percentile male occupant responses. For comparisons with the inversion-eversion responses of 5th-percentile female dummies, the tolerable moments are scaled by the appropriate factor of 0.51.

COMPARISON OF ATD RESPONSES WITH BIOMECHANICAL DATA

Axial Force in the Tibia

The work of several investigators suggests that for dynamic loading of the heel, nominally aligned with the axis of the tibia-fibula complex, limiting the maximum force to 8 kN may be expected to limit the incidence of fractures of the calcaneus and fractures of the distal tibia, with or without extensions into the anatomic joints.

In pendulum impact tests of this type, attributed to Crandall by Tarrière and Viano (1995), the maximum forces measured on Hybrid III lower legs were of the order of twice the corresponding maxima observed in equivalent tests on cadaver legs. The ranges of peak force were 5500-7700 N for the Hybrid III and 1800-2500 N for the cadavers. In tests using the Renault test device to accelerate the heel directly from an initially zero velocity, the average peak force seen by the Hybrid III was 2678 N, while the corresponding figure for the cadavers was 1398 N. Depending to some extent on how the lower leg is accelerated, it thus appears that the axial force measured on the Hybrid III tibia is 2 to 3 times greater than the value observed on a cadaveric specimen. Limiting the axial force on the ATD tibia to 8 kN should thus ensure significantly smaller axial forces in the human tibia.

Flexion and Inversion/Eversion of the Foot

Interpretation of lower tibia moments

In principle, the two moments measured at the tibia transducers, may include the effects of external forces acting directly on the tibia shaft between them. However, as Saul and Zuby (1992) have pointed out, it is not possible to determine the contribution of such forces to the forces and moments observed at the transducers. A necessary assumption in interpreting the data from the Hybrid III legs is, therefore, that no contact occurred between the tibia shaft and the vehicle structure or any other external object, during the test. (Paint transfer or other simple means can be used to detect any such contacts.) Under the assumption of no external contact with the tibia shaft, it is then reasonable to attribute the moments observed at the lower tibia transducer to the forced displacements of the foot.

As noted in Appendix B, however, both tibia transducers are displaced from the axis of the lower leg, extending from the centre of the knee clevis to that of the ankle joint. The lower tranducer is not, therefore, located at the ankle joint, but some distance above and behind it. A calculation is required to determine the moments acting at the nominal location of the ankle, assuming that the forces and moments acting at the transducer are entirely attributable to the forced displacements of the foot. In the absence of the requisite data, inertial effects are neglected.

For the dorsiflexion and plantar flexion data, the only calculation made here is to subtract the portion of the y-axis moment attributable to the axial force F_z in the tibia, as described in Appendix B. The calculation is approximate, since it ignores the inertia of that part of the leg between the transducer reference point and the centre of the ankle joint. The moment associated with the force F_x should also be considered, but F_x was measured in only two of the tests reported below.

In the case of the inversion/eversion data, from 5th-percentile female ATDs, it is required to determine the value of the x-axis moment at the ankle joint with respect to the axis between knee clevis and ankle joint. The angle between that axis and the axis of the tibia shaft, for the 5th-percentile female ATD, was 8.2°. The x-axis moment observed at the lower tibia transducer may therefore be resolved into two components, one parallel with the axis of the lower leg and one normal to that axis. The normal component of the observed moment is $M_x' = M_x \cos 8.2°$, which differs very little from M_x. Also relevant to the x-axis moment at the ankle is the lateral force F_y, measured at the lower tibia transducer. Following similar logic to that described above in estimating the flexion moments at the ankle, the effect of the lateral force F_y on the observed x-axis moment is also taken into account.

The interpretation of measurements obtained from the lower tibia transducer should be significantly improved by the recently announced additions to the standard sensor complement for the 50th-percentile Hybrid III tibia. Simultaneous measurements of the axial force at both upper and lower transducers provide information on the instantaneous axial acceleration of the tibia. The provision of both x- and y-axis bending moments at the lower transducer allows the consideration of flexion and inversion/eversion of the foot. However, the presumably unavoidable omission of F_y from the lower tibia transducer limits, to some extent, the interpretation of inversion/eversion moments.

Experimental data on flexion and inversion/eversion moments

The data of Parenteau *et al.*,(1995) and Crandall *et al.*, (1996), summarised in Section 2.2.4, provide a reasonable basis for estimating the displacements and associated moments tolerable by human subjects in these modes. However, the most recent Hybrid III foot, with 45- and 35-degree limits on flexion displacements at the ankle and a stiff rubber washer to prevent metal-to-metal contact, still provides no simulation of the resistive moments induced in the lower extremities over the range of such angular displacements. As Crandall *et al.*,(1996) show, for the Hybrid III, the moment resisting dorsiflexion is essentially zero until the displacement reaches 35°, at which point it begins to rise rapidly, tending to infinitely stiff at 45° dorsiflexion. An essentially similar response is to be expected in plantar flexion, with the limit at 35°. The same concerns are associated with the responses in inversion/eversion

Direct comparisons between the flexion moments observed in crash tests and the tentative injury levels given in Table 1 remain problematical as a result of the dynamic characteristics of the current Hybrid III foot and ankle. Notwithstanding the "soft-stop" rubber washer now embodied in the foot/ankle assembly, it is to be expected that many of the more severe exceedances of the tolerable dorsiflexion moments will continue to be associated with the abrupt change in angular velocity of the foot as it hits the rubber stop. In consequence, the observed injury measures are likely to exceed the values that would be observed if the ATD ankle provided a more progressive increase in resistive moment as angular displacement increased. Essentially similar concerns relate to the other modes of displacement of the foot.

While the comparisons that follow may, with some justification, be regarded as simplistic, they are arguably more informative as to the real risk of foot and ankle injuries than the continued use of the lower Tibia Index. As earlier noted, the current limit of 225 Nm on the value of the Index substantially overstates the moments that are tolerable without injury of the lower tibia, ankle and foot. Moreover, the Index obscures the differences among the differing moments tolerable in the four modes of displacement of the foot.

Figure 1 shows the dorsiflexion and plantar flexion moments for left and right leg pairs *in 56 km/h offset* frontal crashes into an EEVC deformable barrier (Welbourne, 1996). The broken lines indicate the tentative tolerance levels of 60 and 30 Nm respectively, for 50th-percentile male occupants. In these tests, it should be noted that the ankle joints were <u>not</u> equipped with the rubber washer at the ankle.

It can be seen that five of the ten legs exceeded the suggested dorsiflexion limit and nine exceeded the corresponding limit for plantar flexion. However, only one leg exceeded the current 225 Nm limit on tibia bending moment. Basically similar results, albeit with higher maximum moments, were observed in similar tests at 60 km/h. Referring the forces and moments to the nominal location of the ankle joint generally has the effect of reducing dorsiflexion moments and increasing plantar flexion moments, because of the predominantly negative observed values of F_x and F_z.

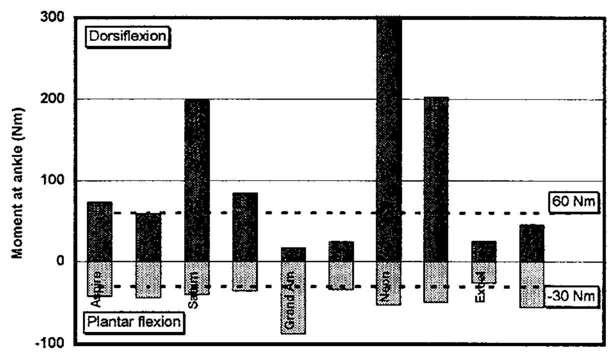

Figure 1. Dorsiflexion and plantar flexion moments in offset frontal crashes at 56 km/h into EEVC barrier.

Figures 2 and 3 show, respectively, left and right leg inversion and eversion responses of 5th-percentile female drivers in 40 km/h offset frontal crashes into EEVC deformable barriers. In considering the results, it should be noted that that speed is considerably lower than the 56 km/h specified in Directive 96/79/EC.

Eversion moments were within the tentative limits, for both feet, for all nine vehicles. For the right foot, all inversion moments were also less than the suggested 8 Nm. For the left feet, tolerable inversion moments were exceeded in four of the nine cases.

The effect of referring the moments to the nominal location of the ankle was essentially neutral for inversion of the left feet and eversion of the right. For eversion of the left feet and inversion of the right, the reference tended slightly to increase the magnitude of the moments.

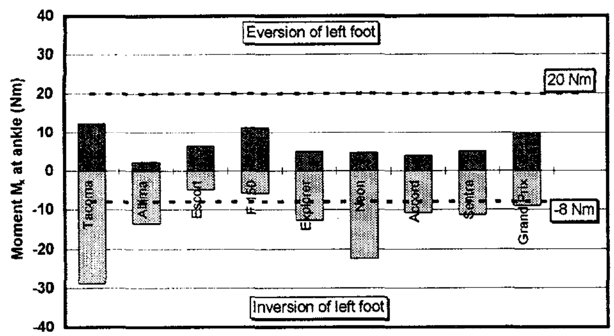

Figure 2. Inversion/eversion moments for 5th-percentile female drivers in offset frontal crashes at 40 km/h into EEVC barrier.

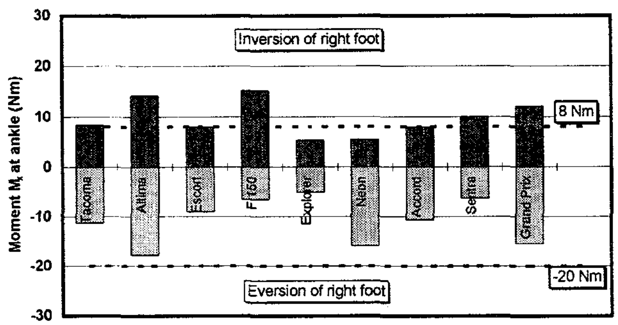

Figure 3. Inversion/eversion moments for 5th-percentile female drivers in offset frontal crashes at 40 km/h into EEVC barrier.

CONCLUSIONS

The Tibia Index is largely ineffective as a measure of the risk of injury to the lower extremities and in particular, to the foot, ankle and distal tibia-fibula complex.

The value of the Index at the upper transducer location is often inflated by the unusual geometry of the tibia, which induces a y-axis moment proportional to the axial force in the tibia. Owing to the proximity of the pin-jointed knee clevis, real y-axis moments of any significance are, however, unlikely to occur at the upper end of the tibia.

At the lower transducer location, the Tibia Index limit of 225 Nm permits moments greatly in excess of the tolerable levels for the flexion and inversion/eversion of the foot. It also serves to obscure the differences among the tolerable moments in the four modes of displacement of the foot.

Interpretation of the data obtained from the tibia transducers will be facilitated by recently announced improvements to the instrumentation of the 50th-percentile tibia. Pending the availability of more advanced ATDs for frontal impact, a modified Hybrid III ankle design providing a progressive increase in resistive moments with angular displacement of the foot would further improve the validity of flexion and inversion/eversion moments measured in regulatory tests.

ACKNOWLEDGMENTS

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APPENDIX A: THE TIBIA INDEX

A1 Introduction

The Tibia Index, defined by Mertz (1984) as in Equation (A1) below, appears at first sight to follow standard engineering practice in estimating the failure load of a column under combined bending and compression. It does not, however, follow that practice in accounting for the generally different strengths of materials in tension and compression. Some confusion as to the significance of the Index has therefore resulted.

It should be noted that the quantitative results given here derive directly from an assumed (though plausible) ratio of the quasi-static tensile and compressive strengths of tibial bone. Those results are, however, used purely to illustrate the consequences of considering the difference between tensile and compressive strengths. In the interests of simplicity, the analysis also preserves the convenient assumption of linear, elastic material behaviour. It is <u>not</u> the purpose of this Appendix to propose an alternative formulation of the Tibia Index, in view of the doubtful utility of the concept in controlling injuries of the distal tibia, ankle and foot.

A2 Tibia Failure under Combined 3-point Bending and Compression

Mertz (1984) defined the Tibia Index as:

$$TI = (F_z/35\,900 + M_r/225), \qquad \text{(A1)}$$

where F_z is the axial (compressive) force and the resultant bending moment, M_r is given by:

$$M_r = (M_x^2 + M_y^2)^{1/2}\,. \qquad \text{(A2)}$$

In order to derive an index of the basic form of Equation (A1), it is usual to consider both the maximum stresses acting at the critical section and the strengths of the material in tension and compression. To illustrate the effect of accounting for a difference in tensile and compressive strengths, it is sufficient here to assume an arbitrary value of the ratio of compressive to tensile strengths in bending, say 1.25 to 1.

Since the tensile strength is less than the compressive strength, we conclude that the basic failure mode of the tibia in pure three-point bending is tensile. Similarly, we associate the maximum crushing force of 35.9 kN, with a compressive stress of 1.25 times the maximum tensile stress.

With the assumed ratio of the tensile strength of the tibial bone to its compressive strength of 1/1.25 or 0.8, failure of the tibia under a purely tensile force would therefore be expected to occur when that force attained a value of 0.8(35900) or 28720 N.

With the assumed linear relationship between the maximum stresses in the critical section and the bending moment at that section, <u>tensile</u> failure of the tibia will occur when:

$$M_r/225 - F_z/28720 = 1, \tag{A3}$$

since the compressive load F_z reduces the tensile stress at the critical section.

Failure of the tibia in compression will however, occur when:

$$M_r/225 + F_z/35900 = 1 \tag{A4}$$

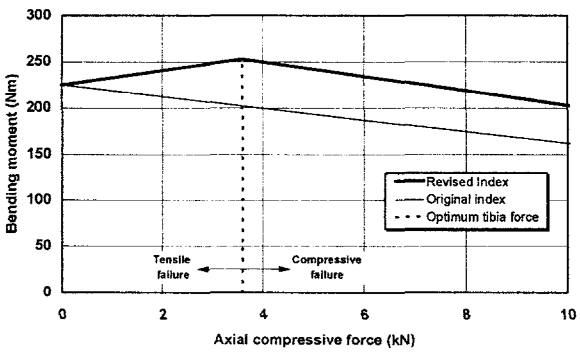

Figure A1. Revised Tibia Index.

Figure A1 shows the revised and original tibia indices in graphical form. The horizontal axis has been truncated, since the maximum axial compressive force may not exceed 8 kN.

The vertical dotted line separates the two primary failure modes, tension to the left and compression to the right. It is evident that considering the tensile and compressive failure modes separately makes a relatively modest difference to the value of the index, at least with the ratio of tensile to compressive strength assumed here. The difference is greatest at the boundary between the two failure modes, where the tolerable bending moment is 25 percent or about 50 Nm greater than if the tensile and compressive strengths of the tibia are assumed equal.

A3 Effect of Increasing the Tibia Index to 1.3

The final version of ECE R 94/01 limits the value of the Tibia Index to 1.3 rather than the conventional 1.0. The increase in the limit effectively eliminates the index as such, so that the optimum combination of axial force and bending moment consists simply of the two individual maxima. Under such loading, the value of the Index is 1.223: the value of 1.3 is not attainable without exceeding one or other of 8 kN or 225 Nm, the individual axial force and moment limits. The effect of the change is shown in Figure A2.

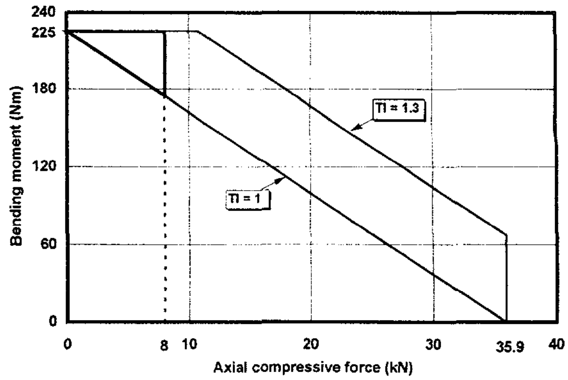

Figure A2. Maximum Tibia Index of 1.3.

The heavily outlined triangle indicates the actual extent of the increase in combined loading permitted in practice by a nominal Tibia Index limit of 1.3.

APPENDIX B: BENDING OF THE HYBRID III TIBIA INDUCED BY COMPRESSIVE FORCES

B1 Introduction

In any currently practicable ATD, gross simplifications of the human prototype are unavoidable. It is nonetheless essential that the quantities measured on the ATD and compared with proposed critical values be consistent with the injury mechanisms they are intended to control in human subjects. It is not clear that that is the case in the regulatory application of the Tibia Index currently proposed by WP29 and EEVC WG11.

In this Appendix, the primary issue of concern is the alignment, in the sagittal plane, of the compressive load paths in human and Hybrid III tibia. The consequences of the unusual geometry of the Hybrid III tibia for the observed bending moments and for the application of the injury measures discussed in Section 3 of the paper are outlined in the following sections.

B2 Geometry of Human and Hybrid III Tibiae

B2.1 Human tibia

A recent edition of Gray's Anatomy [B2] provides comprehensive descriptions of the form and function of femur, knee and tibia.

The femoral and tibial condyles, which provide the bearing surfaces for compressive forces transmitted between the two largest bones in the body, extend medially, laterally and in the posterior direction with respect to the long axes of both bones. (The small anterior extensions of the condyles are negligible in the present context.) Viewed from the side, i.e., in the sagittal or x-z plane of the leg, the posterior extension of the condyles is apparently associated with a rearward offset of the load path with respect to the long axis of the tibia, of about one quarter of the width of the tibia in that plane (op. cit., Fig. 5-70). Flexion of the knee does not appear to change the compressive load path significantly (op. cit., Fig. 4-210, Fig 5-60).

For one adult male subject of nominally 50th-percentile height but lesser mass, the posterior displacement of the load path was estimated to be about 13 mm. However, given the variable cross-sections of the tibia and fibula and their irregular shape, it is not possible estimate the position of the neutral axis of the tibia-fibula complex with any confidence, from two-dimensional images. Whatever the true magnitude of the local displacement of the load path, it is almost certainly much less than the offset of the 50th-percentile Hybrid III knee clevis from the long axis of the tibia. In the absence of any obvious alternative, a straight line between the knee clevis and the ankle joint is therefore used as the reference axis for the forces and moments that act on the tibia-fibula complex.

B2.2 Hybrid III tibia

Figure B1 shows the essential geometry of the Hybrid III tibia, viewed in the sagittal or x-z plane. [Anon.(1994)] It can be seen that the posterior displacement of the knee clevis with respect to the shaft of the tibia is 1.67 inches (42.4 mm). A similar, though lesser anterior displacement of the ankle joint is also apparent. The reasons for the discontinuities in the load path between the knee and ankle joints are not apparent. Their consequences with respect to the forces and moments observed at the upper and lower tibia transducers are readily demonstrated, however.

B3 Effect of Hybrid III Tibia Geometry on Injury Measures

B3.1 Static compression

In the lateral view of the tibia geometry in Figure B1 below, unit compressive forces (1 Newton) are assumed to act at the knee and ankle pivots, such that the tibia is in static equilibrium. The adjacent free-body diagram of the shaft of the tibia, shows the values of the bending moments, axial and shear forces, acting at the transducer reference points, which are required for equilibrium of the shaft under the unit forces applied at the pivots. In particular, it can be seen that a moment M_y equal to 0.02802 Nm is induced at the upper tibia transducer, a corresponding moment equal to 0.00633Nm is induced at the lower tibia transducer and that the force in the tibia shaft is 0.98944 of the force acting between the two pivots.

That part of the observed value of M_y which is attributable to an external moment may be calculated for the upper tibia as:

$$M_y' = M_y + 0.02832\ F_z. \tag{B1}$$

where M_y is the measured moment at the upper tibia transducer and F_z is the (constant) axial force in the tibia shaft. Similarly, at the lower tibia transducer:

$$M_y' = M_y + 0.006402\ F_z. \tag{B2}$$

The signs of the observed moments and forces are significant in the foregoing equations.

An interesting consequence of the upper tibia geometry is that the (original) maximum Tibia Index value of 1 is reached before either of the independent limits on axial force or bending moment is attained. Under a static compressive force acting between the knee and ankle joints, the index reaches unity when:

$$F_z = -6505\ \text{N};\ M_y = 184.2\ \text{Nm}. \tag{B3}$$

Under such loading it is therefore impossible to attain the critical axial load of 8 kN in the Hybrid III tibia, without having previously exceeded the combined bending and compression limits.

Regardless of the particular loading conditions, it is desirable to refer the observed forces and moments to the knee-ankle axis and, more importantly, to the ankle joint.

B3.2 Dynamic equilibrium

Provided that no external contacts with the shaft tibia are observed during a test, an informative analysis of the dynamic equilibrium of the tibia appears feasible, given adequate instrumentation. However the subject is not considered further in this paper.

B4 References

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Gray, H. (1985) Anatomy of the human body. 30th American Edition (C.D. Clemente, Ed.), Philadelphia: Lea and Febiger.

Anon. (1994) Drawing Number B3071. Dummy load cell reference guide, Robert A. Denton Inc.

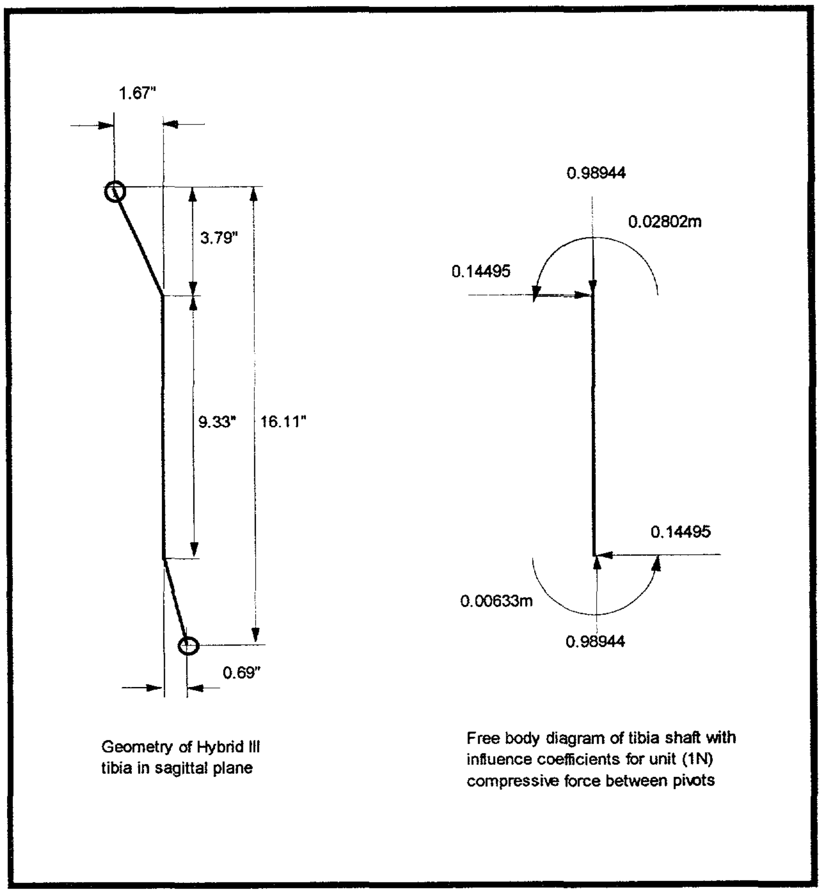

Figure B1. Geometry and influence coefficients for 50th-percentile Hybrid III tibia in static compression.