REQUIREMENTS FOR MINIMISING THORACIC INJURY IN SIDE IMPACT ACCIDENTS

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1. INTRODUCTION

In side impact accidents to cars a very significant proportion of serious and fatal accidents to occupants involve injury to the thorax. There has therefore been a major effort over many years to investigate the causes of thoracic injury, and to provide more protection against the occurrence of such injuries by improvements in vehicle design. An important part of this effort has been the development of side impact dummies which can be used in experimental crashes to provide adequate information about the injuries which would have been suffered by human occupants.

The present paper starts by reviewing the information available about the biomechanics of thoracic injury, and suggests that some of the data available from non-automotive sources may not have been fully exploited in the past. Examination of the data leads to a theory of injury causation by wave propagation through the soft tissues of the body, which in turn leads to a re-examination of suitable criteria for estimating occupant injuries. The practicability of using such injury criteria has been investigated by the use of a number of one-dimensional simulation models to compare the mechanical response of a dummy to different types of impact. The initial simulation model was based on the EUROSID dummy and consisted entirely of lumped springs and masses. Subsequently a method of representing distributed mass elements by a wave propagation model has been developed, and distributed mass elements have recently been included in the model for comparative purposes.

Most of the dummy simulation work has been concerned with investigating the effects of simple "pendulum" impacts, in which it is seen that varying the mass of the pendulum impactor can have very significant effects on the kind of response shown by the dummy. The effect of introducing padding between the thorax and the impactor has also been investigated, and whilst it was found that a thin (25mm or less) piece of padding can almost eliminate contusion type injuries caused by compression waves, even thick padding is not likely to be very effective in reducing the internal laceration and rupture injuries caused by shear waves. The final development of the present suite of simulation models embeds the thorax model in a vehicle side impact collision model, and enables the effects of changing parameters such as the occupant position, stiffness of the impacted vehicle, and characteristics of the colliding object, to be studied. This model has already demonstrated the importance of controlling rebound characteristics in minimising occupant injury, but its full potential will not be exploited until the simpler models are more completely understood. The complete side-impact model is not dicusssed further in this paper, due to limitations of space.

2. MECHANISMS OF INJURY

2.1 Sources of Data. Data on injuries received in road accidents has been obtained from a number of sources, including in-depth investigation of real accidents, cadaver experiments, and animal experiments.

Although in the end we must be concerned with what happens in real accidents, the main use of accident data[1] is statistical, giving the relative frequency of occurrence of different types of injury in various types of accident. There are severe limitations in the use of such data, on its own, for investigating the detailed mechanisims of injury, as there is unlikely to be adequate information about the actual loadings imposed on vehicle occupants. Another problem is that complete information on injuries is usually only available for fatalilties, as those are the only subjects on which autopsies are performed.

Although such information is available from the results of animal experiments, current injury criteria for the thorax appear to be based mainly on the results of experiments using cadavers. These have variously been struck by moving masses (pendulum impactor experiments)[2,3] dropped sideways on to rigid or padded floor surfaces[4,5], or propelled sideways into rigid or padded walls [3,6]. Measurements have been made of the acceleration of ribs and vertebrae, and of the overall compression of the chest. Injury has been recorded in terms of rib fractures, internal lacerations, and other effects such as rupture of the aorta. The results of such experiments have commonly been explained in terms of lumped parameter systems, where the masses are concentrated at the ribs and spine, ignoring the fact that in reality most of the body mass resides in the soft tissues. In the light of the simulation experiments described later, a general weakness of these cadaver experiments is the lack of variation in the mass used to strike the thorax. In practice most experiments have been confined to two masses, pendulum impactors of about 23.4kg (51.5lb), and the effectively infinite mass of the earth (or walls fixed to the earth), although some earlier pendulum experiments have used smaller masses. Apparent discrepancies between experiments involving different impacting masses have been noted.

Much additional information about injury mechanisms can be obtained by studying the results of animal experiments on "blunt impact" and exposure to blast waves[8,9,10,11,12]. There is also some information on the response of living humans to vibration[13,14], and clinical information about the effect of blast on human beings, resulting from wartime observations[15,16]. Although many of the animal experiments were carried out in the course of studies not related to road accidents, the biomechanical problems of injury by blunt impact and blast appear to be identical to the problems encountered in road accident injuries. Apart from the advantage of knowing that the mechanical properties of tissues have not been altered by changes after death, the use of live (anaesthetised) animals also demonstrates the occurrence of types of injury which do not appear to have been observed in cadaver experiments. This is particularly the case for contusions of the lung (and gut), where very significant haemorrhage can occur without visible lacerations. This is a common cause of death after exposure to blast or impact by light non-penetrating masses.

It may be noted that lung injuries are recorded as a major cause of death in side impact accidents[1], with lacerations and contusions being about equally common. Lung contusions are noted much more rarely in "seriously injured" cases; it is possible that this is because they are less easily diagnosed when no autopsy has been performed. It may also be noted that although serious chest injuries in road accidents are frequently associated with multiple rib fractures, in the Birmingham Side Impact Study[1], 37% of the occupants suffering fatal chest injuries had no fractured ribs at all.

2.2 Types of Injury. Careful study of the available data suggests that three separate types of chest injury can be distinguished. These are:-

i) Contusions to the lung parenchyma, in which microscopic damage to the alveoli causes a general haemorrhage. It seems likely that this damage is caused by very localised high stress in which the gross movement or distortion of tissues is insufficient to cause lacerations or ruptures. There is evidence from animal experiments(7) of high pressure pulses occurring in the lungs within a very short time (1 msec or less) after impact. In the Birmingham side impact study this type of injury was suffered by 28% of fatalities.

ii) Lacerations or rupture of the heart and great vessels. These appear to be caused by major distortion and stretching of tissue in the chest cavity. Cine X ray film of animals clearly demonstrates the existence of this type of movement. The transmission speed of the disturbance is much slower than for the contusion-causing pressure pulses, with a transmission time from impact to peak distortion of the order of 10 msec in the pig. A comparable time would be expected for man.

iii) Rib fracture, leading to possible laceration of the lung parenchyma or heart by a broken end. In the absence of such lacerations, rib fracture by itself is not a serious injury, unless so many ribs are fractured that "flail chest" develops. Two types of rib fracture can be distinguished. The less serious is the simple break, in which the fracture is more of a "crack" in which the ends do not separate. This appears to be analogous to simple bending failure of a beam, under a slowly increasing load. In a multiple fracture a section of the rib separates and is pushed inwards, with a liability to cause lacerations of the soft tissues. This appears to be analogous to the failure of a section of a beam in shear, under a suddenly applied load.

2.3 A biomechanical model of chest impact. The detailed mechanics of rib failure have not yet been investigated. Qualitatively it appears that the analogy with bending or shear failures of a beam provides an adequate explanation of the observed facts. The remarks below are concerned with the soft tissue injuries.

In order to cause internal damage to the chest, it is necessary to have a mechanism for transferring energy from the impacting body (or external blast wave) to the interior. Most current models of the thorax are based on the concept of lumped masses, connected by springs and dampers to transmit forces, and hence energy. However, apart from the fact that such models are conceptually unrealistic, practical experience suggests that it is not possible to derive values of the mass/spring/damper parameters which will give a good 'fit' to experimental results over a wide range of impact conditions. Since real biological systems have distributed parameters, it seems desirable to develop models which also have distributed parameters.

It is found that the transfer of energy in distributed parameter systems is most easily described in terms of wave motion, and an outline description of such a model is given in Section 4. Two main types of wave motion are possible in a solid material, compression (longitudinal) waves, and shear waves. The propagation of these two types of waves is described in Reference 13.

In compression waves the material moves in the same direction as the wave front, and in most biological materials the wave velocity is of the order of 1500m/sec. An exception to this is in the lung parenchyma, where the wave velocity is only about 30 to 40m/sec.[7]. A consequence of this low velocity is that a severe impact on the outside of the chest wall can produce a compression wave in the lung material in which the "particle" velocity can become as high as the wave velocity. Under these circumstances the wave front becomes very steep and is effectively a shock wave. This can no longer be described in terms of the mathematics of "small" amplitude waves, and very high local stresses can be developed. A description of the physics of shock waves by Penney and Pike[17] suggests that such shock wave formation in the lung is probably the cause of the contusions observed after exposure to bomb blast; the same mechanism would be equally valid for contusions caused by blunt impact.

In describing the transmission of shear waves the material can be considered to be effectively incompressible, and in an infinite medium shear wave energy is transmitted by lateral waves, with no particle movement in the longitudinal direction (ie normal to the wave front). Such a wave motion could not be excited directly by a longitudinal impact. However in a finite body the situation is more complex and particles in most places would have both lateral and longitudinal components of motion. The mathematical description of shear waves moving in finite bodies is extremely difficult, and exact solutions are not generally possible, though of course numerical solutions can be obtained at some cost for computing. A solution can be obtained for surface waves at the edge of a semi-infinite medium (Rayleigh waves), and in this case the particles are found to move in elliptical paths in a way analagous to water waves, though in the case of Rayleigh waves the controlling influence is shear stiffness rather than gravity. In a shape like the thoracic cross section, shear waves would be expected to have a significant longitudinal component of particle motion, and so would be excited by a longitudinal impact. Lacerations and ruptures could clearly be caused by distortion strains of the type involved in shear waves, and it seems highly probable that such injuries are associated with shear waves.

Von Gierke[13] has discussed the propagation of shear waves in soft tissue, and estimates a value for the shear modulus which would give a propagation velocity of only 1.5m/sec. This is rather low to account for the phenomena observed in blunt impacts, which require a velocity of about 10m/sec. Von Gierke's estimate was obtained somewhat indirectly, and assumed that damping in the soft tissue could be regarded as viscous. It seems more plausible to regard soft tissue damping as being visco-elastic ("rubbery") in nature and this could well raise the propagation speed to nearer 10m/sec under the conditions relating to blunt impacts.

The total energy imparted to the thorax by a blunt impact will be distributed between the compression and the shear waves; subsequent to the impact the strain waves relating to the different modes will proceed entirely independently, so long as the material behaves linearly. This appears to be an adequate assumption for most practical cases. The actual motion of any particular particle will be the vector sum of the individual motions attributable to each of the wave modes. It would be expected theoretically that the partitioning of energy between modes would be a function of the "time duration" of the impact (described in some suitable way). This in turn would be expected to be a function only of the mass of the impacting object, and not of its velocity. This expection appears to be supported by the experimental data, which also shows that light impactors put proportionally more energy into the compression (longitudinal) mode, than heavy impactors. A proper theoretical basis for the partitioning of energy between modes has not yet been developed. However, Von Gierke[13] shows the results of calculations on a very simple physical system which suggest that under the conditions relevant to blunt impact on the thorax, the proportion of energy transmitted by the compression waves will always be very small (though they can still be very damaging in some tissue areas). It therefore seems a reasonable approximation to suggest that the energy in the shear waves can be calculated by assuming that all the absorbed energy is transmitted by this mode, and the amplitude of the compression wave can be estimated from the change of velocity of the chest wall occurring over a "short" time. The definition of "short" is not yet clear, but is probably of the order of one millisecond or less.

3. INJURY CRITERIA

Current injury criteria for the thorax are mainly based on the probability of rib fractures, on the assumption that this correlates quite well with the probability of soft tissue damage. This may well be true for a particular mass of the impactor, but if it is desired to develop injury criteria which are effective over a range of impactor masses, the evidence from animal impactor experiments suggests that this assumption is unjustified. Experience of injuries received in the more complex circumstances of real accidents suggests that each of the three injury types described in Section 2.2 above can occur independently of the others, and that three separate injury criteria should be specified in order to cover all possibilities of serious or fatal injury. Such criteria are being developed to meet requirements for the EUROSID side impact dummy[18].

The following possibilities are suggested:

i) Lung contusions. These appear to be caused by a compression wave of sufficient intensity to produce particle velocities approaching the local wave speed in the lung parenchyma. Animal experiments suggest that a change in velocity of the chest wall of greater than about 10m/sec in a "short"time is potentially very dangerous. On theoretical grounds a velocity of 10m/sec at the chest wall would be expected to give rise to a particle velocity of 20m/sec or more in the lung parenchyma, which is close to the wave velocity.

ii) Lacerations and ruptures. A criterion is required which describes the level of shear strain in the appropriate parts of the thorax (ie in the mediastinum). On the basis of blunt impact experiments on rabbits, Viano[11] has suggested a "viscous injury criterion". This is the maximum of the instantaneous value of; (chest wall velocity X chest deflection). Although there is no apparent reason for this to be a good criterion for injury from the compression waves produced by light impactors, it is plausible that it may correlate quite well with peak shear strain. More work on the form of shear waves in a structure like the chest would be necessary to confirm this.

In practice, in order to produce a criterion which has dimensions of length rather than the square of length, it seems preferable to take the square root of Viano's criterion. the values of Viscous Injury Criterion quoted later in this paper are this square root, multiplied by 1000 to give units of reasonable size (using metres as the unit of distance). In simulation experiments on the EUROSID dummy (described below) it was found that this criterion correlated moderately well with the force on the rib damper, and it is suggested that for the present this damper force could be taken as a more suitable criterion for shear wave type injuries in experiments using EUROSID.

iii) Rib fracture. It may be desirable eventually to distinguish between bending fractures and shear fractures, but further work will be necessary to produce suitable criteria. Present criteria used for rib fracture include functions of rib accelerations (combining data from several ribs), spine acceleration, and rib deflection. At present it seems likely that for a dummy like EUROSID rib deflection is the most appropriate criterion.

4. SIMULATION MODELS OF THE CHEST

A one dimensional computer model has been developed as an aid to the design of the thorax for the EUROSID dummy[18,19]. The layout is shown in Fig. 1,

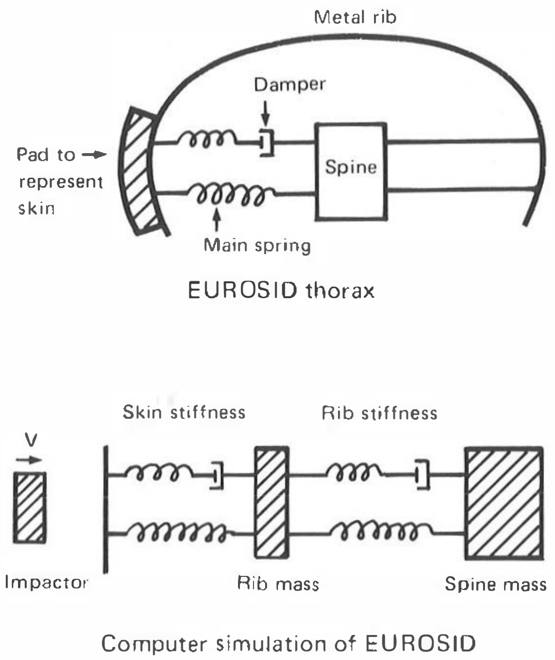

Fig. 1 EUROSID thorax and its representation in the computer simulation

in which it can be seen that there are basically two lumped parameter mass/spring/damper systems, one representing the skin layer over the ribs, and the other representing the connection between the impacted part of the rib and the spine. In EUROSID the skin layer is represented by a thickness of absorbent padding, and the ribs are represented by spring/damper systems having dynamic properties similar to those of the simulation model, which had previously been matched to the results of cadaver experiments. It will be noted that the dampers both have additional springs in series with them in order to give visco-elastic damping properties. This is thought to be more representative of biological systems than the use of simple viscous dampers. The properties of a system with viscoelastic damping are described in Reference 19.

The springs in the model are basically linear, but can be made to stiffen up as the deflection approaches a pre-set limit. This facility is important in the skin sub-model to prevent the impactor "passing through" the ribs in high velocity impacts. The output from the model includes various parameters which could be used as injury criteria, in accordance with the principles described in Section 3 above. These parameters include peak rib deflection, peak viscous injury criterion, change in velocity of chest wall (ie rib) and time required for this change, and peak load in the main damper in EUROSID, which it was thought might correlate quite well with the Viscous injury Criterion. Other parameters measured which have been used in the past as injury criteria are peak accelerations of rib and spine, peak load on the skin, and the Gadd severity index. With the possible exception of peak spine acceleration, these did not appear very relevant in the present studies.

The model has been used to investigate the effect on the various injury criteria of changing the mass and velocity of the impactor, and of providing energy absorbent padding outside the skin. Three types of padding were used, exemplified by the characteristics shown in Fig. 2. Surprisingly, changing the type of padding did not appear to have a very marked effect on the injury parameters, though the progressive force increase (type 3) seemed best on an overall basis.

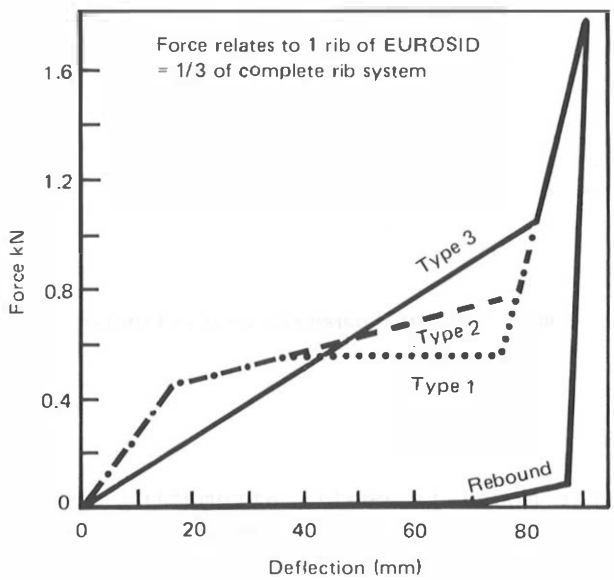

Fig. 2 Padding characteristics

A recent development of the model has been the inclusion of elements with distributed parameters. It has been usual to model distributed parameter systems by using finite element techniques, but this tends to be very expensive in computing time and computer storage requirements. The new technique models the distributed parameter element in terms of its wave propagation characteristics. These can be calculated from the density, elasticity , and damping properties (viscous or viscoelastic) of the material. The computer model consists of two "delay" arrays describing waves travelling in opposite directions through the medium, with appropriate subroutines describing boundary conditions at each edge of the medium where the waves interface with either another distributed parameter system, or with a conventional lumped parameter simulation model. Full details of the methodology will be given in a forthcoming report.

The new program element is at present restricted to one-dimensional distributed systems, and is strictly only applicable provided the system remains linear. The first application has been a thoracic model in which the rib spring and viscoelastic damping system have been replaced by a distributed parameter element, which also contains a large proportion of the mass which was previously concentrated in the spine. The layout is shown in Fig. 3. The characteristics of the model are broadly similar to those of the lumped parameter system, except that the spinal acceleration characteristics appear to be much closer to those observed in cadaver experiments.

Part of distributed parameter element

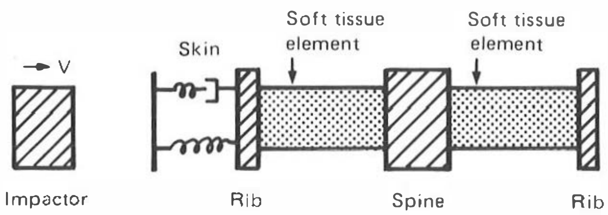

Fig. 3 Distributed parameter model of thorax

The distributed parameter model is really attempting to model the soft tissues in the thorax, which contain most of the mass in real animal and human systems. However, the ribs are responsible for a significant part of the externally observed stiffness of the chest, and it may be desirable to model them as an additional structure to the soft tissues. It is known from experiments on EUROSID that its model rib does not behave like a simple one-dimensional mass/spring system, and it seems likely that a two-dimensional model will be necessary to explain its behaviour. A suitable model element to simulate rib behaviour could provide a useful addition to the current thorax model. No attempt has yet been made to model the lung tissue as a separate element in the simulation.

5. SOME DETAILED RESULTS

5.1 Effects of varying impactor mass and speed. The response to impact of both the lumped parameter and distributed parameter thoracic models has been investigated over a wide range of impactor masses and velocities. It has been found that for any particular impactor mass the time constants (eg time to peak rib deformation) are nearly independent of impactor velocity, and all the parameters proposed as injury criteria are directly proportional to the velocity. These relationships only start to break down at high impact velocities where the "skin" deformation is driven well into the non-linear region (ie where force starts to increase more rapidly then linearly with deformation). This is the result that would be expected for a "linear" system, and is in accordance with the results of animal experiments(12).

The effects of changing impactor mass are much more complex, and it is not possible to relate even peak rib deformation directly to the input energy of the impactor. After impact the initial energy is partitioned between the impactor (the remaining energy depends upon the amount of "bounce"), the spring and damper relating to the skin over the ribs, and the spring and damper relating to the main rib system. The relative partitioning of the energy depends only on the impactor mass, and not on its velocity, provided the velocity is low enough for the system to remain linear.

Fig. 4 shows how the impactor velocity, and the corresponding input energy, required to produce a 40mm peak deformation of the ribs, vary with impactor mass. The energy is in Centre of Gravity co-ordinates, and refers to a single rib of the EUROSID dummy, ie one third of a complete rib system.

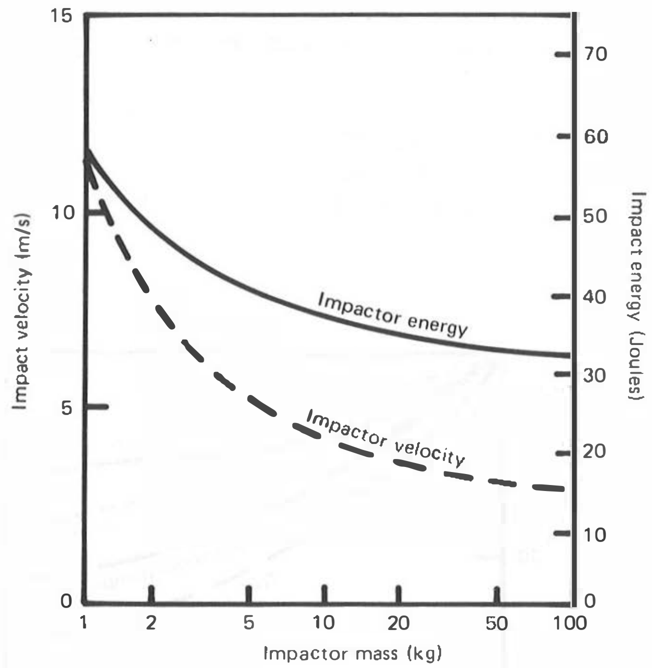

Fig. 4 Effect of impactor mass on velocity and energy necessary to give a 40mm rib deflection

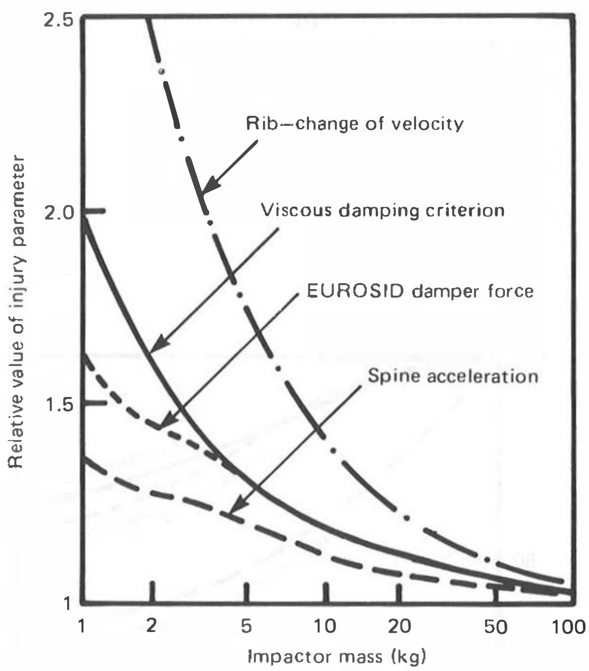

Fig. 5 Effect of impactor mass on relative value of injury parameters

Fig. 5 shows the variation with impactor mass of a number of other parameters which could be used as injury criteria. The impactor velocity has been adjusted to give a 40mm peak rib deformation, for all values of impactor mass. (It has been assumed that a 50mm deformation would be about the survivable limit.) The parameter values are shown as the ratio of the value at the relevant mass to the value for a very large impactor mass (ie the rigid wall type of test). The actual base values of the parameters are tabulated below:

Parameter	Value for lumped parameter model	Value for distributed parameter model
Impactor velocity	2.78 m/s	2.73 m/s
Viscous damping criterion	227	226
EUROSID damper force	575 Newtons	680 Newtons *
Rib change of velocity	3.45 m/s	3.07 m/s
Peak spine acceleration	14.8 g	25.5 g

* Peak force in chest "spring" element near spine

TABLE 1. Base values of possible injury parameters for very high mass impactor (ie rigid wall test), with velocity set to give peak rib deformation of 40mm.

It will be noted that apart from spine acceleration, the values for the distributed parameter model are fairly similar to those for the lumped parameter one. This also applies to the variation of parameter values with impactor mass, except for small masses of less than 2kg, and the results are not plotted here.

5.2 Effect of padding. Figs. 6 and 7 show the effect of interposing padding between the impactor and a lumped parameter representation of the thorax for two different impactor masses. The impactor velocity is the same as that

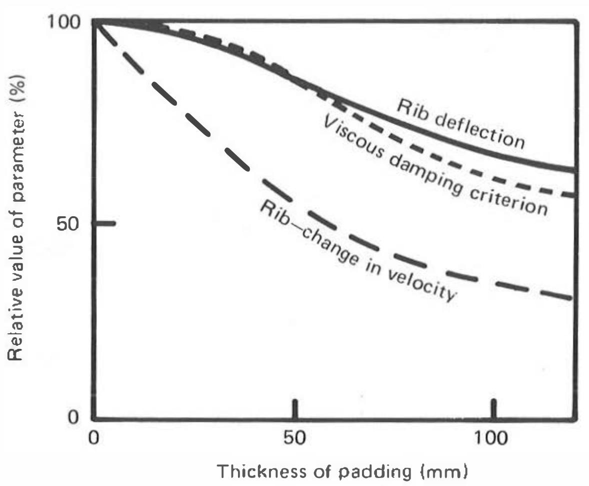

Fig. 6 Effect of padding thickness on injury criteria
Impactor mass = 2kg — lumped mass model

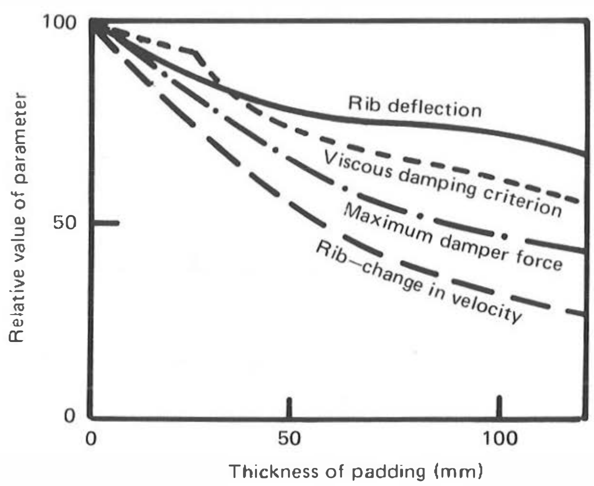

Fig. 7 Effect of padding thickness on injury criteria
Impactor mass = 10kg — lumped mass model

required to give a 40mm rib deformation without padding. In Fig. 6 relating to a 2kg impactor, the curve for the EUROSID damper force follows the Viscous Damping Criterion very closely, and is not shown separately. In Fig. 7, relating to a 10kg impactor, the kink in the viscous damping criterion arises because there are generally two peaks of viscous damping criterion at different times of impact. At the 'kink', the absolute maximum is shifting from one peak to the other. In both figures it will be noted that increasing the thickness of padding has far more effect on the rib velocity criterion relating to compression wave injuries, than on the other criteria which are probably indicative of shear wave injuries. In fact the real effect on compression wave injuries would be more marked than suggested by these curves, as padding increases the time over which the velocity change occurs, as well as decreasing the magnitude of the change.

Fig. 8, relates to a distributed mass representation of the thorax, and shows an interesting effect in which moderate thicknesses of padding actually increse the viscous damping criterion. This is due to an impedance matching effect; it has also been observed under some conditions for the lumped mass model, and can even increase peak rib deformation.

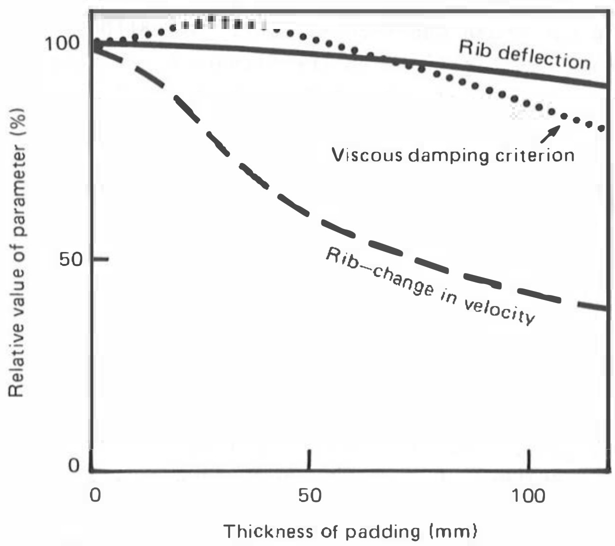

Fig. 8 Effect of padding thickness on injury criteria
Impactor mass = 2kg — distributed mass model

6. SUMMARY AND CONCLUSIONS

A comprehensive study of the data available on thoracic injury, especially from animal experiments, suggests that the causes of injury are complex and, apart from those due to rib fractures, best described in terms of two types of wave propagation. It is proposed that three independant criteria should be used to assess the probability of injury.

Simulation models have been developed to investigate the probable behaviour of EUROSID under a variety of impact conditions, and to compare its performance with that of a more sophisticated model using distributed parameter elements.

Preliminary experiments with padding suggested that if the padding has to operate over a wide range of impact types, a material in which the force increases progressively with deformation is likely to be more suitable than one with a constant crush force. Padding thicknesses of only about 25mm deep can reduce the compression waves causing contusion type injuries substantially, but even greater thicknesses are unlikely to be very effective in reducing shear wave type injuries.

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