BODY PROTECTORS FOR HORSE-RIDERS

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ABSTRACT

Body protectors are intended to reduce the risk of soft tissue and spinal injuries to horse riders. Simple tests of current products between flat rigid anvils show that they are too thin to be effective in preventing soft tissue injuries. Assessment of the protection of the spine is more difficult because of its flexibility. Initial tests between a hemispherical striker and a rigid flat support give a different ranking of the designs than a test in which the back protector is mounted on a flexible spine. The best protection is given when there is a plastic skin of high bending stiffness over the foam layer; however this design may restrict the mobility of the rider to an unacceptable degree. Further development of the products is necessary.

1 INTRODUCTION

The need for back or body protectors for horse riders has been discussed at a meeting on "Equestrian Safety" in London in November 1988, held by the British Horse Society. At this meeting the injuries were described (broken ribs, damage to the kidneys, liver and other soft organs, and spinal injuries). It was discussed whether a British Standard could be developed in the same way the one for riding helmets, but this was felt to be premature because there was no consensus over the design criteria for such garments. In May 1989 the journal "Horse and Hound" had a special issue on body protectors(1). At that date the Jockey Club, and the British Horse Society both recommended particular makes of body protectors for use in competitive events. However no reasons were given why certain makes were recommended and others not. Given this confusion it seemed worthwhile to carry out some research on the protective performance of the existing garments.

Severe spinal injuries are quite common in riding accidents; of 237 patients seen at a Cambridge accident department there were 1 cervical and 4 lumbar spinal fractures (2). A Berkshire study of accidents in 1984 showed 10 spinal injuries, and 2 lumbar fractures causing partial paralysis, among 59 amateur riders, compared with only 2 spinal injuries among jockeys. A West Midlands survey(4) of 1000 injuried riders in 1986 showed that 10% of the 1554 injuries were spinal (10 cervical 10 dorsal and 17 lumbar fractures, 56 neck strains, the rest soft tissue injuries), 9% were chest and abdominal injuries (32 rib fracture(s), 5 kidney rupture, 4 spleen rupture, 5 liver rupture, 1 heart rupture and 90 bruising). In this survey less than 5% of the riders wore a body protector. Of the mechanisms of injury 29% involved jumping, 23% a fall, 14% the horse rearing, 10% a kick, 8% a vehicle, and 6% being crushed by the horse.

2 INJURY MECHANISMS

Before any test methods can be proposed the nature of the injuries and the mechanisms that cause them must be established. A division into soft tissue and spinal injuries can conveniently be made.

2.1 Soft tissue injuries

If a horse rider has a minor accident in which he/she falls vertically through 2 metres, with no forward component of velocity, the impact velocity with the ground is 6.26 m/s. Fig 1 shows the dynamic force - deflection charactarictic of cadaver chests; for deflections in excess of 20 mm the force required can be estimated as 3 ± 1 kN. The 'Viscous Criterion' has been proposed(5) as a means of assessing the probability of injuries of soft tissue in the chest. It is stated that such injuries are likely if the product of the instantaneous velocity V of the surface of the body and the compressive strain C of the chest cavity exceeds 1 m/s.
A calculation for a torso of mass 28 kg and chest thickness 250 mm that deforms at a constant force of 3 kN when it impacts a flat rigid surface at 6.26 m/s shows that the product VC reaches a maximum value of 1.55 m/s when the compressive deflection is 115 mm. Hence the criterion predicts that some body protector is necessary to prevent a chest injury. If the impact velocity is reduced to 5 m/s then the predicted peak VC is 0.9 m/s at a chest deflection of 80 mm. The subsequent peak chest deflection is 115 mm which should be compared with the typical thickness of body protector products of 10 mm.

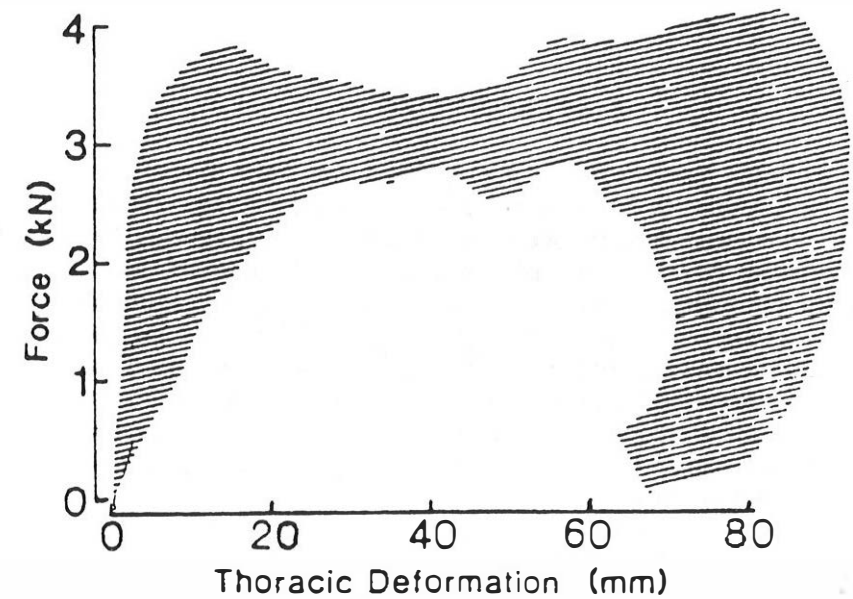

Fig 1. Dynamic force - deflection charactaristics of the human chest, derived from USCD cadaver data.

2.2 Spinal injuries

If a rider is thrown from a horse and hits rough ground or a wooden obstacle used in 'eventing', or is kicked while on the ground, then there can be localised impacts to his/her spine. Apart from the localised force on the vertebra(e) there will also be shear and bending forces on the spine, and these can be analysed as for other engineering beams. Riding back injuries range in severity from quadraplegia due to spinal column damage, through damage to the spinal processes to muscular sprains. A review of spinal injuries (6) states that 'few studies have been conducted to determine the shear forces for cervical injury' and only quotes data for experiments on monkeys. For these, forces in the range 1 to 2 kN produce failures respectively of the odontoid and of the C2-C3 endplate. If cervical vertebrae are isolated from the human spine and loaded axially in compression, the failure forces are in the range 3 to 5 kN (7).

3 BODY PROTECTORS AVAILABLE

3.1 Construction

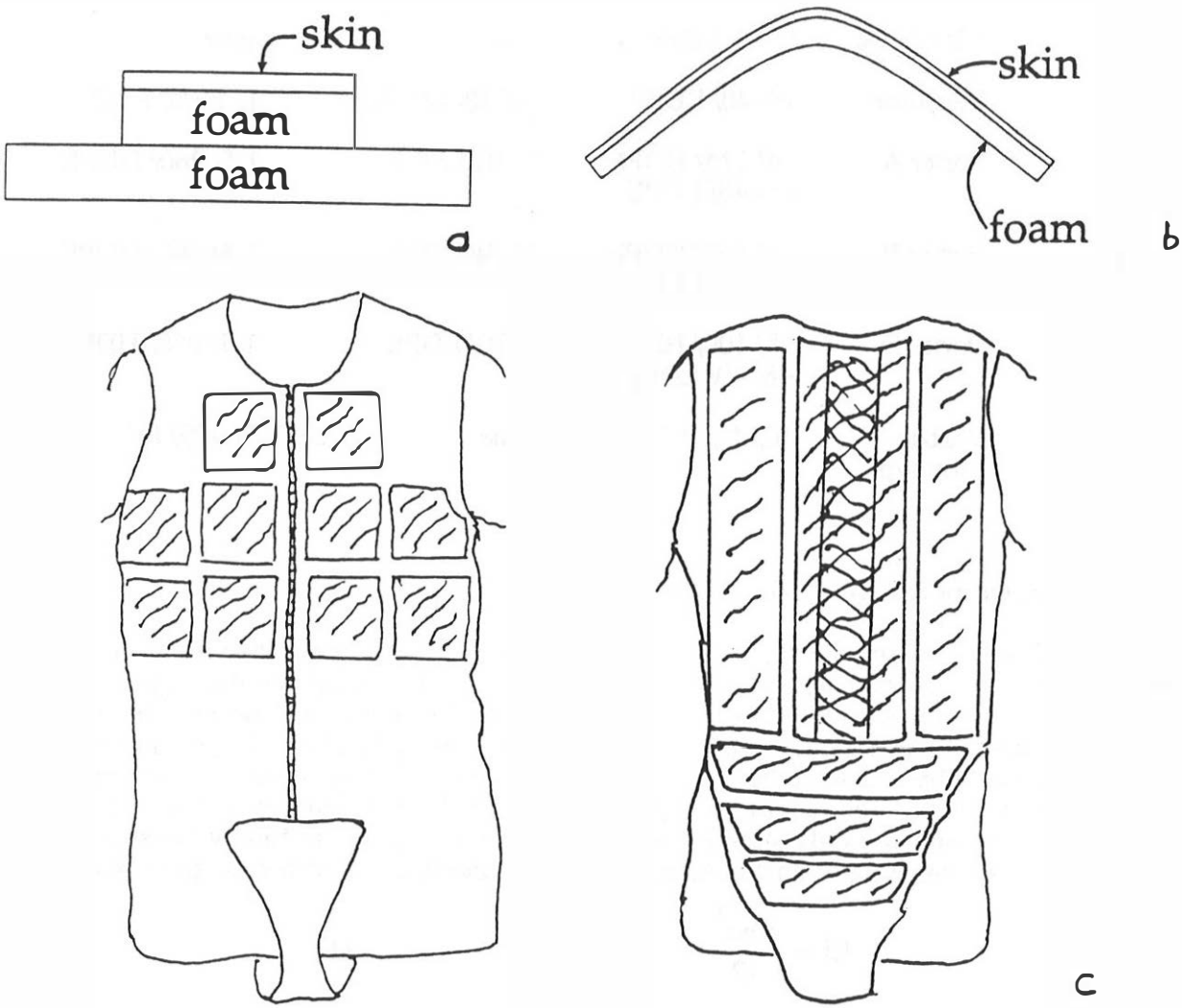

Fig 2.The cross section of two types of back protector a) Porter, b) Barber prototype, c) the foam locations in the front and rear of the Ransome body protector

Figure 2 shows the construction of the common types of body and back protector. In the former there is protective foam covering most of the exposed area of the back and chest, and there may be a strip of protective solid plastic 'skin' covering the spine. This foam is held in place in pockets in a zipped up garment, often with a crutch strap between the legs to prevent it riding up in use. In the latter there is only coverage of the spine and the lower part of the back; because the foam is thicker and therefore less easily bent the back protectors do not cover the shoulder blades, because to do so would restrict movement. Table 1 gives the constructional detail of the body protectors tested. The ones listed as being made by Porter are experimental constructions related to a commercial product. The three pieces of information given under each entry in the table are the thickness of the material in mm, the density of the material in kg/m^3, and the abbreviation for the plastic used. Thus LDPE is low density polyethylene, HDPE is high density polyethylene, ABS is acrylonitrile butadiene styrene copolymer, PU is polyurethane, PVC is poly vinyl chloride, and PP is polypropylene.The last entry in table 1 is a prototype made by Mr Barber, an orthopaedic surgeon, using materials normally employed to make surgical braces for the neck.

Table 1 Construction Materials and Dimensions

Type	Make	Main foam	Spinal strip foam	Spinal strip skin
Body	Air o Wear	6/ 40/ LDPE	none	none
Body	Ransome	10/ 40/ LDPE	10/ 40/ LDPE	1/ 1400/ PVC
Back	Porter A	4/ 175/ HDPE +5/40/ LDPE	5/ 30/ LDPE	1.5/ 960/ HDPE
Back	Porter B	5/ 100/ HDPE +10/ 40/ LDPE	5/ 30/ LDPE	1.5/ 960/ HDPE
Back	Porter D	8/ 100/ HDPE +8/ 40/ LDPE	5/ 30/ LDPE	1.5/ 960/ HDPE
Back	Barber prototype	10/ 40/ PU	none	5/ 909/ PP

3.2 Assessment of the solid skin materials

If the solid skin covering the spine of the wearer is to be of any benefit then it must distribute the impact forces over an area of more than one vertebra. Assuming that the impact is with a blunt object then the skin will not be cut, and force redistribution will occur if the skin has adequate bending stiffness. It therefore plays the same role as the shell of a protective helmet. In contrast with a helmet shell the body protector skin does not have doubly positive curvature (this is inherently stiff and requires a high force for the spherical shape to 'pop-in'); it is flat in the horizontal plane and only slightly curved in the vertical plane to follow the shape of the spine. Therefore the longitudinal bending stiffness of the skin of width w is given by

$$EI = \frac{Ewt^3}{12} \qquad (1)$$

where E is the Youngs modulus of the material , I is the second moment of area of the beam, and t is the thickness. For the Porter A product in Table 1 the values are t = 1.5 mm, w = 80 mm and E = 1.0 GN/m^2, so the bending stiffness is 0.022 Nm^2 . Although the skin is stuck on the surface of a layer of up to 20 mm of foam the Youngs modulus of this foam is so low that the neutral axis of the 'skin + foam' composite beam is still close to the centre of the skin, so there is no large increase in bending stiffness. Consequently the bending stiffness of this type of product is low.

The only back protector with adequate bending stiffness in its skin design was the prototype made by Mr Barber, who used a 5 mm layer of polypropylene moulded into a V shape in its horizontal section. This increases the I value of the beam dramatically (fig 2b); the equivalent values for the rectangular cross-section beam of the same bending stiffness are w = 15 mm and t = 120 mm, so using equation (1) with E = 1.0 GNm^{-2} gives a bending stiffness EI = 1660 Nm^2. Although the buckling of the sides of the skin will reduce this value when high forces are applied, it is clear that this design has a high bending stiffness, and this may make its load spreading capability much higher than the other products.

3.3 Assessment of the plastic foam layers

Plastic foams have been assessed for use in riding helmets (8). In this the impact stress strain curves were integrated up to the point where the stress exceeded a value of 2.5 MNm^{-2}. This gave a value of the energy density absorbed by unit volume of foam before the local stresses on the skull were enough to cause head injury. The same principle can be used to assess the foams in body protectors, but the maximum stress level should be adjusted to allow for the different injury mechanisms. There is a further problem with some of these foams in that their structure can be temporarily or permanently damaged by the first impact. We have studied the change in the impact properties of HDPE foams of density between 73 and 98 kg/m^3 for impacts with and without intervening recovery periods(9). This showed that the useful energy density in a second impact might be only 60% of that in the first impact. Recovery for a week at 20° C would mean that the second impact could absorb 75% of the first impact energy density without exceeding a critical stress level. The softer LDPE and PU foams of table 1 are better at recovering rapidly because their cell walls are less rigid.

Figure 3 shows the impact stress- strain curves of one of the softer foams suitable for general use in body protectors. Although the stress is in the range 0 to 0.5 MNm^{-2} for strains up to 60% there is a rapid rise at strains of the order of 80%. At this strain the gas in the closed cells has been compressed adiabatically to about 17% of its original volume and the internal gas pressure is at least 0.6 MNm^{-2}. The theory of the stress strain curves of such foams is quite well understood(10), and it shows that it is impossible to have a constant compressive yield stress because the cell gas pressure contributes too much to the total stress.

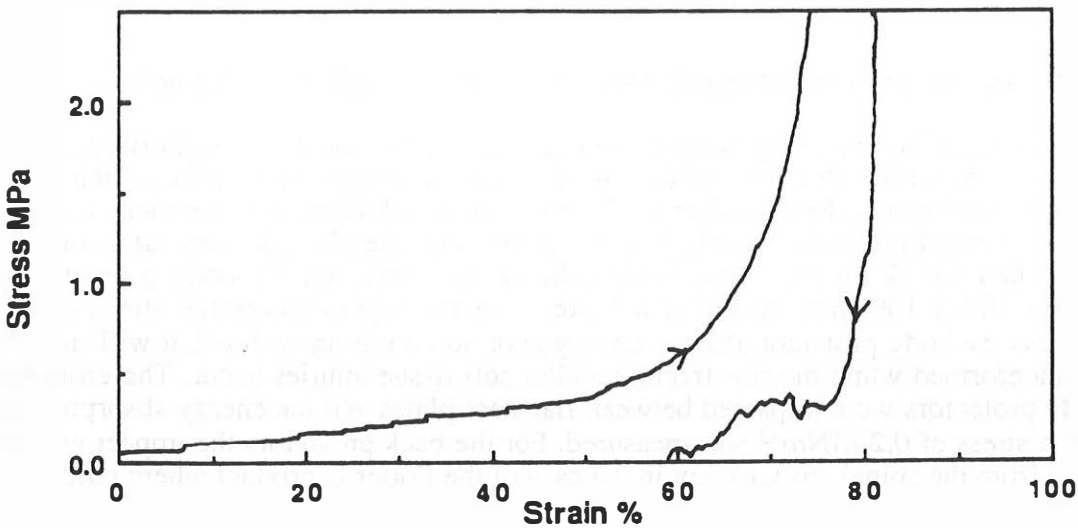

Fig 3. Impact stress strain data for a soft LDPE foam used in body protectors.

4 SIMPLE IMPACT TESTS FOR PRODUCT ASSESSMENT

4.1 Falls on to a flat surface

Consider a fall where a rider's back or chest hits a soft surface such as a grassy field. Figure 4 shows that there are at least three deformable elements in a simple mechanical model of the impact. In the simplest approximation inertial and viscoelastic effects in the chest will be ignored. We need some information about the impact properties of the ground surface; a limited amount was generated in connection with helmet impacts(8). This showed that deformations of

10 to 20 mm could occur on soft ground. The relative deflections of the three elements under the same load will give us a guide to the relative amounts of energy that they absorb. Typical body protectors contain plastic foam that is 5 to 20 mm thick. If the foam is of the correct compressive strength then it will deform by up to 80% before the stress becomes too high. Therefore the deflection of the body protector can be up to 16 mm.

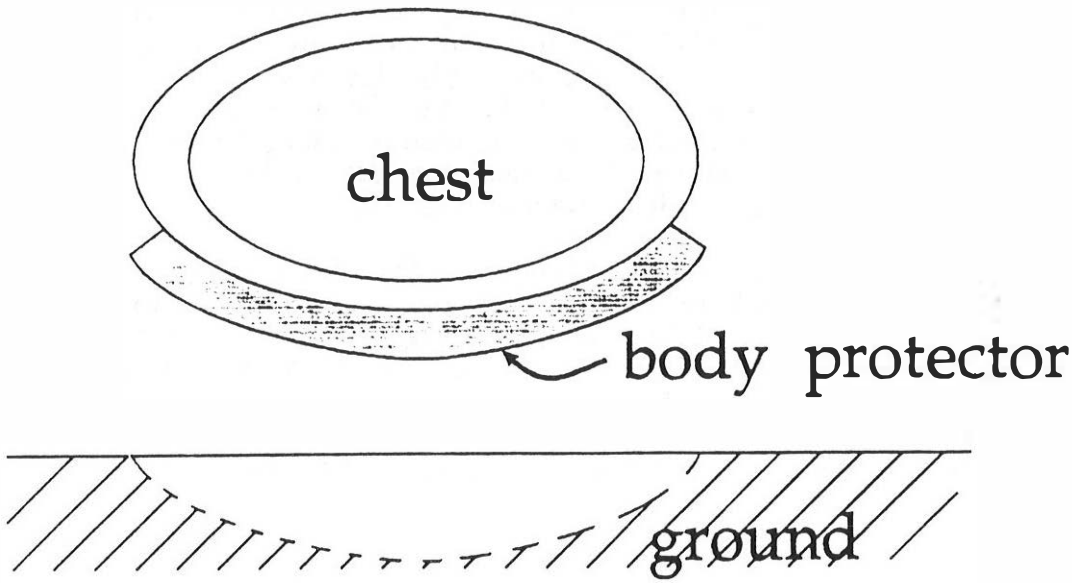

Fig 4. Deformation of the chest, body protector and ground, in a fall onto a flat field.

The simplest method for evaluating body protectors is to support them on a rigid flat surface, and to impact them with a flat steel striker. In this test the energy absorption of the body protector alone is measured; the contributions from the chest and the ground can then be added to estimate the overall protection level. Figure 1 shows that the chest deforms at a force of about 3 kN when struck on the front. Taking the contact area on the body protector as approximately 100 x 150 mm means that there is an average compressive stress of 0.2 MNm^{-2}. Unless the body protector absorbs energy at or about this stress level, it will survive the impact undeformed while the ribs fracture and/or soft tissue injuries occur. Therefore the front of body protectors were impacted between flat steel plates and the energy absorption in Joules up to a stress of 0.2 MNm^{-2} was measured. For the back protectors the impact was on an area away from the spinal strip, except in the case of the Porter D product where there was not room to do this.

The target energy value was assessed as follows:- If the torso has a mass of 28 kg and falls 2 metres its kinetic energy is 560 J. To avoid a risk of a serious chest injury the product VC of the Viscous Criterion should be less than 1 m/s ; the calculation in section 2.1 shows that this occurs if the impact velocity could be reduced to 5 m/s , an impact energy of 350 J. Therefore if the body protector can absorb 210 Joules at a compressive stress level less than 0.2 MNm^{-2}, ie before the chest deforms significantly, then it should reduce injury in this accident scenario. For hard ground the energy absorbed by the ground can be neglected.

The test area of 100 x 100 mm is 2/3 of the estimated torso contact area, so the test specimen should absorb 2/3 of the 210 J estimated above. Table 2 shows that the commercial body protectors do not begin to approach the target of ~ 140 J in the test. The Porter products are back protectors or prototypes thereof, but are included to show the effect of larger thicknesses of foam - low density polyethylene foam of density about 40 kg/m^3. The tests used a drop height of 1 m which was more than enough to crush the foam fully.

Table 2 Impacts on body protectors between flat steel surfaces of area 0.01 m^2

Name	soft foam thickness mm	Energy input at 0.2 MNm^{-2} J	Deflection at 0.2 MNm^{-2} mm
Air o Wear	6	4.1	6
Ransome	10	6.3	6
Porter A	5	2.7	3
Porter B	10	6.6	6
Porter D	13	8.8	9

The reason for only quoting the soft foam thickness in the table is that the dense HDPE foam in the Porter products will not yield at such a low stress as 0.2 MNm^{-2}. The energy values provide a ranking of the body protectors, but none offers more than a low level of protection in a fall which is not broken by prior arm contact etc. As a graph of the energy input versus soft foam thickness from table 2 has a slope of 65 J per 100 mm thickness of foam, it can be estimated that a total thickness of 215 mm of soft foam is required to give adequate protection! It is obviously impractical to wear this amount of foam.

4.2 Protection for the spine against a localised impact

The test used was a hemispherical steel striker with a radius of 50 mm falling onto the part of the body protector that covers the spine, which rested on a fixed flat steel plate. The striker has a kinetic energy of 25 Joules, and it tests the ability of the thermoplastic skin of the protector to distribute a localised force, and the denser foam to absorb energy. Table 3 gives the results. The area under the force deflection graphs like fig 5 while the deflection is increasing is equal to the energy input in Joules.This area was calculated for force limits of 2 kN and 5 kN, as these are of the same order as those required to damage a vertebrae or dislocate the spine, and the results given in table 3 . For the Ransome protector in fig 5 the force rises very rapidly once the deflection is appoximately equal to the thickness of the foam(which must then be fully compressed) so the energy input level to obtain a force of 5 kN is little greater than that needed to produce a 2 kN force.

None of the energy levels in table 3 are large; the energy of a horse kick has been estimated as being at least 400 Joules. Consequently the protection level afforded by these garments is low. A further concern is that the comparative energy levels in table 3 may be wrong because the flexibility of the spine has been neglected. This could mean that the wrong type of designs could be recommended to riders. A more realistic type of test is examined in the next section.

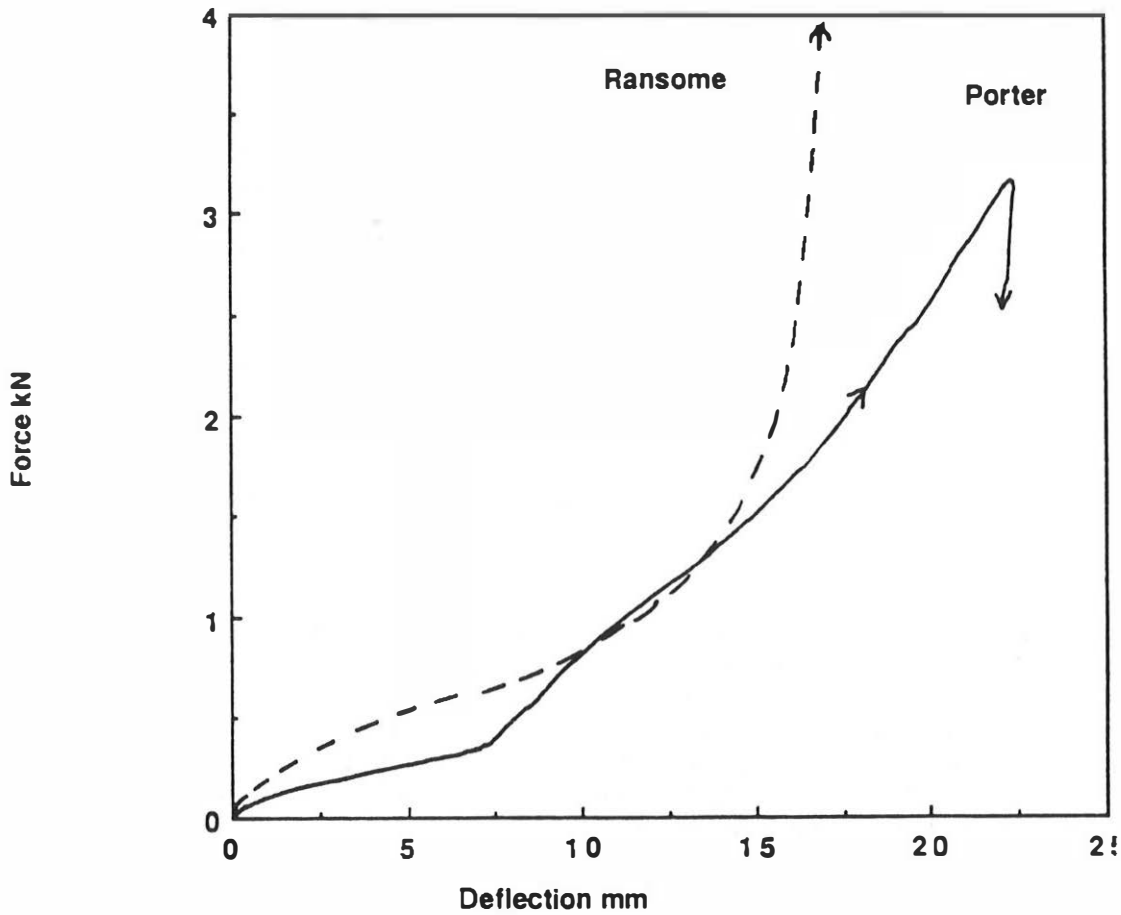

Fig 5. Force - deflection graph for a hemispherical striker impacting the spine region of 2 types of back protector

Table 3 25 Joules hemispherical impacts onto spine area of body protectors supported on a flat steel plate

Name	skin thickness mm	Hard/soft foam thickness mm	Input energy J for force of 2 kN	5 kN
Airo Wear	-	-/12	--	6
Ransome	1.0	-/18	9	12
Porter A	1.5	5/10	15	18
Porter B	1.5	5/10	12	--
Porter D	1.5	8/13	14	--
Barber	5.0	- / 10	11	--

5 SPINAL DAMAGE ASSESSMENT TESTS

5.1 Development of the test method

The drawbacks with the previously described simple test methods are twofold

a) the use of a fixed anvil to represent the back

For some Newtonian mechanics problems there are analyses that allow the comparison between tests carried out using a fixed (or infinite mass) target, and the real life situation of a movable finite mass target. For instance in the development of British Standards for motorcycle helmets there has been a progression from a fixed headform test in BS2001:1960 to a "swingaway " test in BS 2495:1977 (neither is in use now). To allow comparison between the kinetic energies at impact in the two types of test the concept of the "equivalent impact energy" was introduced. For a two body impact problem this is defined as

Equivalent impact energy E = energy input to the helmet up to the time when the masses M_s and M_h have a common velocity

where M_s is the mass of the striker and M_h is the mass of the headform that is free to move. The equivalent impact energy is equal to the striker kinetic energy in a fixed headform rig, and the energy input to the helmet is this value when the striker velocity is zero. When the striker and headform have a common velocity, the crushing of the helmet foam has a maximum value. If the mass of the helmet is ignored, then the equivalent impact energy is

$$E = \frac{M_s V_s^2}{2}\left(\frac{M_h}{M_s + M_h}\right) \qquad (2)$$

When we apply this concept to impacts on the spine, M_h is replaced by some mass associated with the spine. The vertebrae are linked so there is a relationship between the velocities of neighbouring vertebrae. A simple way to deal with this is to use the "effective mass" concept; in this the spine is replaced by a single lumped mass M_L under the impact point. This mass has the same momentum at a velocity V as the sum of the momenta of the vertebrae in a segmented spine. Although it is difficult to obtain an accurate estimate of M_L it is possible to use fixed anvil tests to represent impacts on the back, with equation (2) used to calculate the equivalent impact energy, and with M_L replacing M_h.

b) use of a rigid rather than a flexible spine

Changes in the geometry of the spine during the impact will have a major effect on the load distribution effectiveness of a back protector, if its bending stiffness is comparable or larger than that of the spine. Therefore it was decided to build a more realistic test rig to see the effect of spine flexibility.

5.2 Flexible spine test method

Figure 6 shows the details of the test equipment. The intention is to measure the impact forces on individual vertebrae, using piezoelectric film pressure transducers affixed to the top surfaces of the aluminium blocks that simulate the vertebrae. To avoid excessive complexity the 175g aluminium blocks (25 x 50 x 50 mm) are articulated using pin joints so the 'spine' can bend in one plane only. The support was designed to have the same order of elastic stiffness as the cadaver chests in figure 1, viz 200 N/mm. This was acheived by using 100 mm of artificial playing surface consisting of bonded rubber crumb .

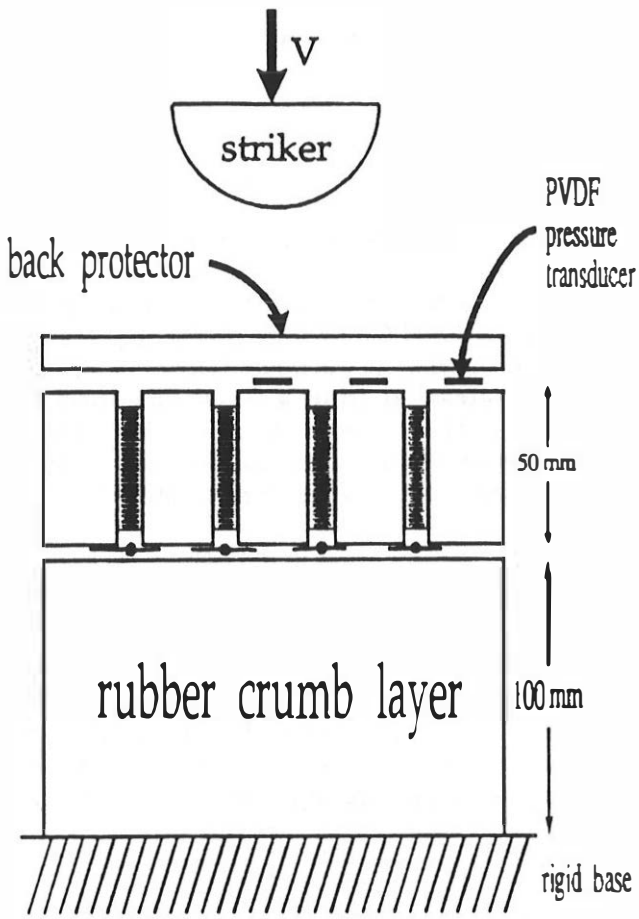

Fig 6. The flexible spine test rig for measuring the force distribution by back protectors.

The 5 kg cylindrical steel striker of radius 50 mm is aligned with the central of the five 'vertebrae'. Its impact energy is 50 Joules. The dimensions of the PVDF transducers at 12 by 30 mm means that they only cover 29% of the surface area of the Aluminium block ,to which they are fixed by double sided adhesive tape so they cannot bend. Impact calibration of the transducers was done under conditions where there is a uniform pressure on the top surface of the block. Unfortunately with the cylindrical striker there can be a higher pressure at the centre of the middle block (where the transducer is) than at the sides, so absolute force measurements become impossible. There is also the problem of spurious signals (due to the flexing of the leads?), noticeable when apparently negative forces sometimes occur on the outermost transducer. Table 4 gives the results of the tests.

Table 4 50 Joules cylindrical striker impacts onto spine area of body protectors supported on the flexible spine + 100 mm of rubber crumb

		Maximum force (N) measured on		
	Striker	**Vertebra 1**	**Vertebra 2**	**Vertebra 3**
Ransome	2500	~5000	0	0
Porter A	2300	~4200	170	400
Porter D	2200	3200	170	100
Barber	2200	~5000	650	950

The output of the central 'Vertebra 1' transducer was clipped by the voltage limit of the transient recorder in several cases. The load spreading ability of the back protector is revealed by the magnitude of the forces on the neighbouring vertebrae 2 and 3. Table 4 shows that these are negligible compared with the 'noise' level of ~200 N, except for the Barber back protector. The Porter D product differs from the A product in having a spinal reinforcement strip that is 170 mm wide instead of 90 mm, but this makes no significant improvement to the force spreading ability. For the Barber prototype, which was calculated in section 3.2 to have a high bending stiffness, the proportion of the total force taken by the 4 neighbouring vertebrae could be as high as 50% allowing for the measurement problems.

6 DISCUSSION

Tests on body protectors, for a fall onto a flat surface, show that the level of energy absorbed safely in them is less than 10% of the 140 Joules estimated to be needed to keep the viscous criterion less than 1 m/s for a fall from 2 metres. Although this does not prove that these products are of little protective value, it does mean that they should be critically evaluated from accident statistics. Since their use is now mandatory in certain types of competitive equestrian sport, comparison of accident figures will have to be made between years before they were worn, and current figures.

The tests on the spine protection ability of back protectors give different rankings if these products are tested on a flexible 'spine'(where the stiff-skinned Barber prototype is best) or if they are tested on a rigid flat support (where the Porter products that have thicker dense foam but a thinner skin are best). This suggests that the former more realistic test method is essential if misleading comparisons are not to be made. The spine can bend significantly during an impact because of the compressibility of the thoracic cavity, and therefore the skin of a back protector undergoes high bending deformation. The tests show that the protective skin of typical commercial products is far too flexible to contribute to load spreading along the spine. A radical re-design is necessary to acheive adequate protection against a locally applied impact on the spine.

The choice of PVDF film force tranducers for the flexible spine test rig was with hindsight a mistake. Both this type of transducer (11) and the more expensive but much more accurate quartz crystal force cells (12) have been used for measuring the force distribution on rigid aluminium dummy faces. As the spine model can make large movements, it is felt that the errors due to moving leads are unacceptably large. Quartz load cells would be better.

There are important ergonomic factors that will decide whether any design of back or body protector is acceptable for use. It must be possible to control the horse, and this means that the mobility of the rider must not be unduly restricted. All these garments also act as thermal insulation for the rider, and in the case of racing their weight must also be considered. Therefore the best that can be expected of laboratory tests at the moment is that they could label products with some energy level, representing the estimated protection in certain types of accident. If such labelled garments could be worn in competitions for a couple of years, and the injury patterns correlated with the marked energy levels, then the benefits of wearing such garments could be assessed. Until such an exercise is carried out, it will be impossible to put forward a National Standard for the testing of body protectors.

7 CONCLUSIONS

It has been shown that some simple tests can be used to evaluate back protectors; for falls onto flat surfaces a simple impact test on to a rigid flat surface is adequate, but to evaluate spine protectors requires a segmented 'spine' on a compliant support. Currently available products are shown to be of limited protective value.

Acknowledgements :

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