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Ankle Protection in Football Shin Guards

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ABSTRACT: Dynamic measurements were made of the forces and pressure distribution on an ankle model, with a variety of ankle protectors, when impacted with football boot studs and blades. The protection level could be improved by using thicker foams, foams of higher modulus, and domed shells of higher stiffness than in commercial products. Statistical analysis showed that the protectors did not affect player performance. Finite Element Analysis (FEA) was used to analyse protector deformation mechanisms and the biomechanics of load distribution to the ankle.

INTRODUCTION

Football is the world's most popular sport, with over 200 million registered players. Football accounts for the majority of sports injuries in Europe. Franke (1977) estimated that in Europe 50–60 % of injuries in sport were due to football. Studies of football injuries (Ekstrand et al (1983), Fried et al (1992), and Hawkins & Fuller (1999)) conclude that 75 – 93% of the total injuries involve the lower extremities, with 17 –26 % involving the ankle and 17 –23 % the knee. Sprains, strains and contusions account for 50–88 % of injuries, while fractures account for 1–10 %.

The shin guard is the only mandatory protective equipment, a rule introduced in 1990. However, there have been no subsequent studies to evaluate shin guard effectiveness in reducing injuries. Players may reject shin guards with ankle protection on the grounds that they inhibit performance, are bulky and uncomfortable. Materials selection and protector design have not been discussed in print. The size of the ankle protector must not inhibit the ability to put on the boot or inhibit performance. The protector is worn under the sock, usually attached to an elasticated strap under the arch. The foam must be sufficiently flexible to prevent discomfort or rubbing.

There is a parallel between the mechanics of ankle protectors and that of hip protectors for the elderly in falls (Mills, 1996). Both products have a shell over foam, and in both load can be shunted away from the vulnerable protruding bone.

BIOMECHANICS OF FOOTBALL INJURIES

The anterior border and the medial surface of the tibia at the front of the leg have very little soft tissue cover to absorb energy from impact blows (Marieb, 1995). The shin guard is specifically designed to protect these areas. The bony protuberances on both sides of the ankle, with very little soft tissue cover, are often exposed by shin guards, leaving them vulnerable to injury. These protuberances have a small contact area with any impacting flat or convex object, and they have high Young's modulus; consequently impact forces and pressures can be high, leading to injuries.

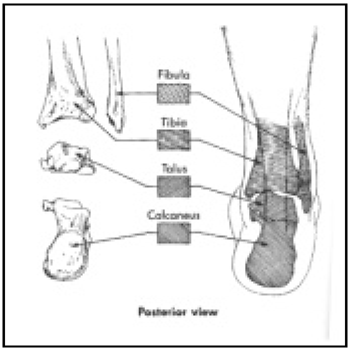


Fig. 1 Structure of the ankle (from Hall, 1995 with kind permission of Mosby Inc)

In a straight-legged tackle, with the bodyweight behind the leg, the kinetic energy can be as high as 992 Joules (Woods, 1994). No current shin guard can protect against tibia fracture with such high impact energies. No criterion could be found for fracture of the talus by direct impact. However the calcaneus fails at impact forces of 3.6 – 11.4 kN when impacted from below, as in a vehicle crash (Seipel *et al*, 2001). This provides a starting point for estimating the critical force to fracture the talus.

Biomechanical knowledge of soft tissue contusions (bruises) is extremely limited. Crisco et al (1996) impacted the leg muscle of rats with a 6.4 mm diameter nylon hemisphere to cause contusions. However there were no direct measurements of the pressure distribution. The average pressure over the projected area of the hemisphere was a maximum of 9 MPa. However, in a review of muscle contusion injuries, Beiner and Jokl (2001) could not propose whether force, pressure or some other mechanical variable was appropriate for a contusion criterion. We assume that the criterion for bruising the soft tissue of the ankle is probably peak pressure, and that the order of magnitude estimate for the minimum impact pressure to cause contusions is 1 MPa.

EXPERIMENTAL METHOD

Ankle model: Investment cast aluminium replicas were made of the ankle bones of a commercial plastic model (Sawbones –1132-1). The 3 parts were assembled using a polyurethane rubber adhesive. The soft tissues of the foot were simulated using Airex 5230 foam – this closed-cell flexible PVC foam has a low Young's modulus ≅ 100 kPa and Poisson's ratio of 0.5 (approximately incompressible). It allows load transfer away from the talar protuberance, which was covered with 2 mm of leather. Fig.2 shows the test rig. The impact site is on the lateral side of the ankle, which was oriented to be the upper surface of the test rig. The lower surface is rigidly supported on a flat table.

The **protectors** were a combination of a foam layer and a domed shell. The foams were either the currently-used ethylene vinyl acetate (EVA) of density 30 kg m^{-3}, or a foamed blend of 40% Dow Ethylene-Styrene Interpolymer (ESI) with 60% low-density polyethylene of density 53 kg m^{-3}, which has improved impact absorption.

Sheets of foam (10 mm thick) were thermoformed to a domed shape. 1 mm of glass fibre composite prepreg (GRP) was vacuum-bag moulded, and 2 mm polycarbonate (PC) sheet thermoformed, into domed shapes diameter 60 mm and radius of curvature 37 mm. Some commercial, 2 mm thick, injection moulded low-density polyethylene (LDPE) discs, with the same shape and diameter, were also used. The foam was bonded to the shell inner surface.

Impact tests. The perceived risk was from rigid projections on the sole of the opponent's boot, contacting small areas on the ankle. Impact tests were carried using metal (aluminium) and plastic (nylon) studs (truncated conical shapes), and with thermoplastic polyurethane (PU) blades (slightly curved, 30 mm long and 4 mm wide at the top) (fig 2.), attached to a vertically falling mass. The worse-case scenario was tested, with direct alignment of the stud with the centre of the ankle protuberance.

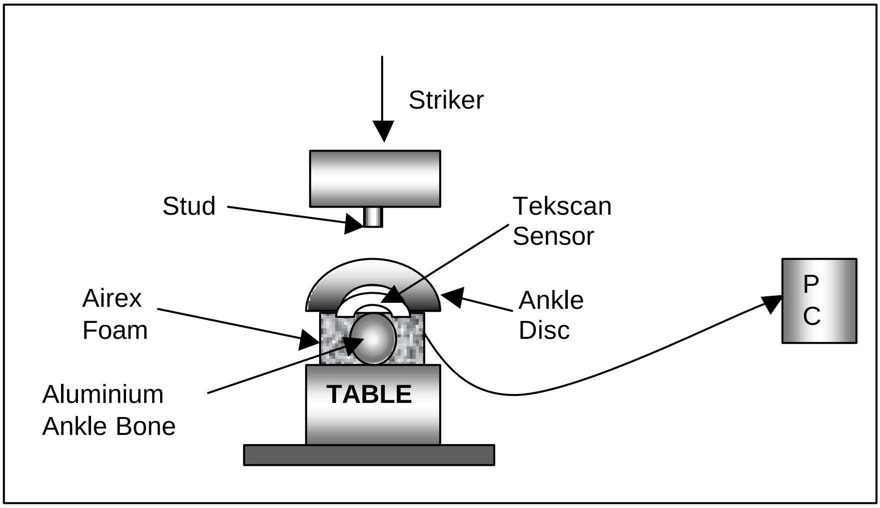


Fig. 2 The impact test rig

Although real impacts occur to a movable ankle, the concept of 'equivalent impact energy' was used. This was developed with circa 1980 motorcycle helmet standards. It is defined as the kinetic energy of a striker hitting a rigidly supported target, that produces the same peak deformation of the protective gear as in the real incident. Equivalent impact energies of 0.9 to 3.5 Joules were used. The signal from a quartz crystal accelerometer on the rear of the striker was recorded digitally. The total impact force was calculated from the striker acceleration. The deflection of the ankle protector surface was computed by numerical integration of the striker acceleration.

The pressure distribution, at the interface between the protector foam and the ankle model, was measured using a Tekscan F-scan thin-film sensor. This determines the pressures at approximately 4 mm intervals, at 176 Hz.

Wearability studies were carried out with volunteers, in a series of training exercises. The subjects were asked to carry out dribbling, sprinting and shooting tasks, while wearing 5, 10, 15 cm thick EVA foam ankle discs. They were timed and asked their opinion of comfort and restriction.

RESULTS

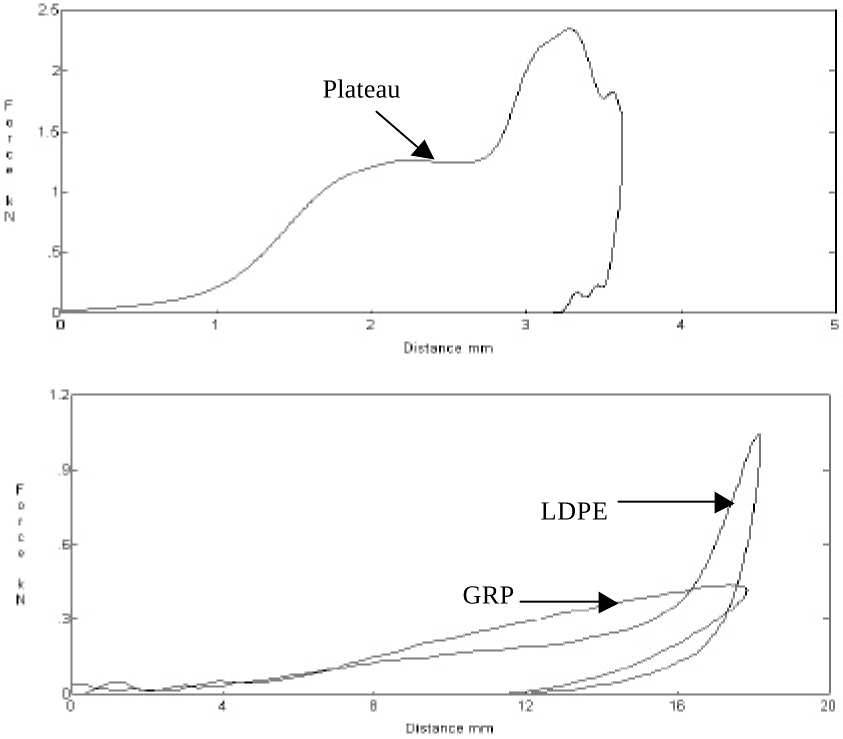
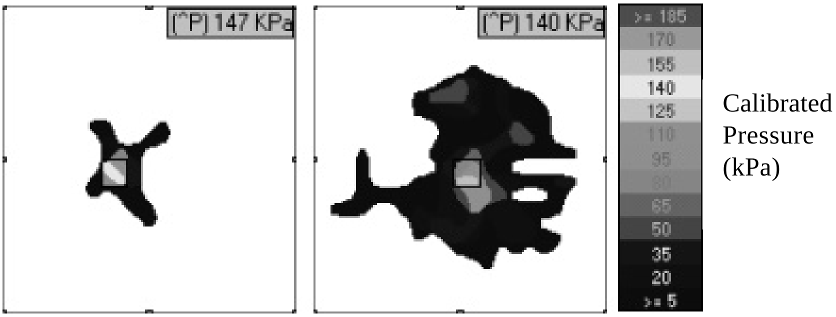


Fig. 3 Force – deflection graphs for impact energy 3.5 J a) metal stud impact on ESI foam, no shell, b) blade impact on LDPE and GRP shells, with ESI foam.

Typical impact force-deflection graphs, for protectors without a shell, show 3 stages. In the first, rising slope phase, the foam between the stud and ankle compresses. The second, with a plateau value of 1 to 2 kN, seems to be the response of the segmented aluminium ankle bone model, once the foam has bottomed out. There is a further force rise in the third stage, when there are no more deformation mechanisms (fig. 3a). There was evidence of the stud damaging and penetrating the foam.

For ankle protectors with a shell, the first energy absorption mechanism is the compression of the foam between the shell and the ankle. The foam compression increases the duration of the impact, and reduces the peak impact force. However the total impulse delivered to the ankle bone will be the same unless another mechanism (load shunting) comes into play. The LDPE, PC and GRP shells provide a range of bending stiffnesses. Experimentally it was found that the GRP shells performed best with the studs and the blades. The combination of GRP shell and ESI foam reduces the force to the lowest level, around 0.5 kN (fig 3b).

The Tekscan sensor shows the **load transfer** away from the talar protuberance, when a shell is used (fig. 4). This is the second, and more effective method of reducing injury risk.



a. b.

Fig. 4 Tekscan images for blade impacts with kinetic energy 1.7 J on ESI foam a) without and b) with GRP shell. Note the load spreading over a greater area in b.

At the higher input energies (up to 3.5 J) the stiffer GRP and PC shells reduced the maximum force more effectively than the LDPE shell. This suggests that the talus would experience high forces for impact energies > 3.5 J, if current commercial ankle protection were worn. Without a shell on the protector, impacts with energies > 10 J would cause forces >5 kN on the talus, and it is probable that fractures would occur.

One way Analysis of Variance (ANOVA) using SPSS, was used to investigate the within – subjects effects in the dribbling and sprinting experiments. The probability values were 0.174 and 0.092 respectively. As these values exceed 0.05, there was no significant change in performance with ankle disc thickness.

FEA OF PROTECTOR DEFORMATION AND LOAD DISTRIBUTION

The model had an axis of rotational symmetry. The stud is an aluminium tapered cylinder of height 10 mm, flat end with diameter 12mm, and edge radius 1 mm. The protector geometry was the same as in the impact tests. The talus geometry was simplified to be a 16 mm diameter cylinder with a 8 mm radius hemispherical end.

The ankle soft tissue was modelled as a nearly-incompressible gel-like material, the hyperelastic model in ABAQUS with Ogden shear moduli $m_1 = m_2 = 200$ kPa, exponents $a_1 = 2$ and $a_2 = -2$, and inverse bulk modulus $D = 1.0 \times 10^{-9}$ Pa^{-1}, values taken from previous modelling. The EVA foam was modelled as a hyperfoam with shear modulus $m = 55$ kPa, Ogden exponent $a = 0.5$, and Poisson’s ratio = 0. The talus had a Young’s modulus of 10 GPa, while the GRP, PC and LDPE shells had Young’s moduli of 15, 2.4 and 0.1 GPa respectively, and appropriate yield stresses.

The large deflection option was used. The ankle tissue was bonded to the ankle bone, as was the protector foam to the protector shell. A friction coefficient of 0.7 was used between the foam and the ankle surface, and between the stud and the protector shell. No attempt was made to model the dynamics of the moving masses. Simulations with aluminium studs produced very similar results to those with PU blades. It has not been possible yet to simulate the semi-rigid ESI foam.

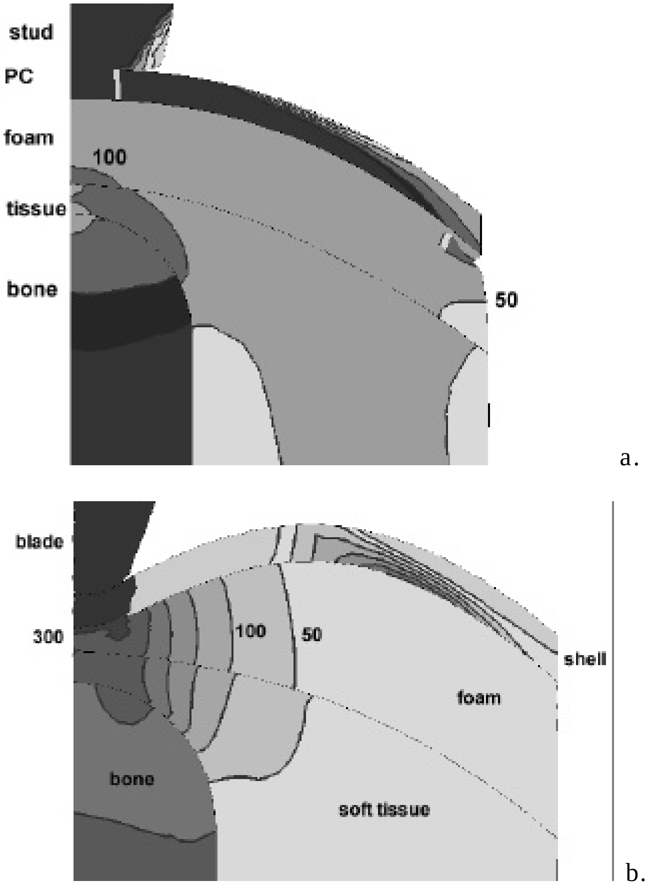


Fig. 5 Predicted deformation mechanisms, for protectors with a) PC, and b) LDPE shells. Contours of the vertical compressive stress in kPa are shown.

The domed PC shell is sufficiently stiff to keep its shape (fig. 5a), so the EVA foam is almost uniformly compressed between the shell and the ankle flesh. However, the LDPE shell buckles at its centre (fig. 5b), reducing the volume of highly-deformed foam. The computations become unstable at deflections (table 1) lower than those in the impact tests. The predicted force vs deflection graph was very close to that measured with 10 mm of EVA foam under a PC shell. The ESI foam (fig 3) provides a higher energy absorption than the EVA foam by a factor of about 3.

Table 1 Data for the simulations shown in fig. 5

Shell	input energy J	peak force N	Peak deflection mm	peak pressure kPa
PC	0.44	168	4.5	100
LDPE	0.33	104	9.6	300

DISCUSSION

The domed protector shell spreads contact loads to a large foam volume, and hence to a large area of the ankle surface. However, the 'soft tissue' of the physical ankle model affects the protector ranking in the experiments. The Airex foam is probably of too low modulus, and too thick, to correctly model the load transfer to the sides of the talar protuberance. Further experiments will use stiffer foams.

FEA can usefully predict both shell deformation mechanisms and load spreading to the surface of the ankle. However, the mechanical properties and shape of the soft tissue of the ankle are, at present, first approximations. High-speed photography could confirm the predicted buckling of the LDPE shell, and instrumentation of the load transferred to the talus could confirm the amount of load spreading.

Given the lack of contusion injury criteria, and statistics of the impact velocities to the ankle, it would be useful to examine football players' contusion patterns, and relate them to the type of ankle protectors worn. This, with further FEA modelling of the muscle and cartilage around the ankle, will eventually lead to optimisation of the ankle protectors. At present it is not possible to specify the optimum shell bending stiffness necessary for a protector.

The impact energies used in the draft EN (2001) standard for football shin protectors have not been justified either by statistics or by kinematic analysis of tackling incidents. Eventually it will be possible to relate the high kinetic energy of a tackling player's leg and foot to the effective kinetic energy of a laboratory test on to a rigidly supported ankle model. Until then it is difficult to specify impact energies for ankle protector tests.

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