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The influence of basketball shoes with increased ankle support on shock attenuation and performance in running and jumping

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The aim of this study was to assess the influence of footwear with increased ankle support on ankle kinematics and on impact loads during landing from a vertical jump using high-speed cinematography, dynamometry and accelerometry in a series of tests in which a rebound action was simulated. To analyse the effect of this increased support on motor performance, two performance tests were designed: a vertical jump test and an obstacle course running test. Two prototype shoes with identical soles but different uppers were used. The first was designed to provide greater ankle support, with such features as a high top, heel counters and a rearfoot lacing system. The second prototype was a less supporting shoe, with low top and no heel counter or any other feature for support. In the shock attenuation test, the use of high-support shoes resulted in higher forefoot impact forces and higher shock transmission to the head, but showed lower shock transmission to the tibia. The use of high support shoes resulted in lower ranges of eversion and higher ranges of inversion of the ankle on landing. In the motor performance tests, the high-support shoes reduced the height jumped and increased the time to complete the running course relative to the low-support shoes. We conclude that increased ankle support reduces ankle eversion range but increases shock transmission, and reduces both jumping and running performance.

Keywords: Ankle support, performance, shock attenuation, sports shoes.

Introduction

A sprained ankle is one of the most common injuries in basketball (Garrick, 1977). It usually occurs when an athlete lands on an opponent's foot or some other obstacle, which may cause a large inversion moment (Stacoff *et al.*, 1990; Shapiro *et al.*, 1994). Most studies in the last 20 years have concluded that the support provided by high-top shoes, prophylactic ankle taping or ankle orthoses reduce the risk of ankle sprain in a number of sports, including basketball (Garrick and Requa, 1973; Robinson *et al.*, 1986; Rovere *et al.*, 1988; Barrett *et al.*, 1993; Barnes and Smith, 1994; Sitler *et al.*, 1994). An ideal prophylactic ankle restriction would support the ligaments of the ankle at the limit of the normal range of motion, thereby preventing abnormal movement (Robinson *et al.*, 1986; Barnes and Smith, 1994). If the greater ankle support reduces the normal range of movement, in particular ankle plantar flexion, the shock attenuation capability of the ankle joint can be reduced. This could mean increased risk of overuse injuries (Kaelin *et al.*, 1988), as well as diminished motor performance.

Impact force can be defined as the force generated by a shock (a collision between two objects) which reaches its maximum within 50 ms after first contact (Nigg and Herzog, 1994). The impact forces can be reduced by the shock-absorbing capability provided by the footwear. The shock-absorbing capability of a sports shoe is important for the prevention of pain and the development of degenerative musculoskeletal diseases, particularly those related to repetitive impacts (Radin *et al.*, 1980; Broom, 1986; Radin, 1987; Kaelin *et al.*, 1988; Ozguven and Berme, 1988). In addition to these impact forces, a shock wave is also generated, which is transmitted along the body, and which has been related

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to low back pain (Voloshin and Wosk, 1982). Shock attenuation can be defined as the reduction of this shock wave transmission and the rate of loading on the subject's body. Some joints, particularly the ankle joint, constitute an important shock attenuation mechanism (Gross and Nelson, 1988), mainly by means of an eccentric muscular action. Restriction of ankle range of motion, particularly plantar flexion, as occurs with most ankle support devices, can limit the function of the ankle as a shock attenuation mechanism. This can lead to higher impact force peaks on landing after jumping (Sussman *et al.*, 1988) and to an increase in the transmission of shock to the body. When plantar flexion of the ankle is restricted, the foot is in a less plantar-flexed position at landing, and the maximum impact force peaks could be generated much sooner after foot–ground contact. When landing after a jump, large and rapid rise-time peak forces (impact forces) are generated, reaching values as much as six times body weight (McClay *et al.*, 1994a). These large forces appear to be more damaging to the musculoskeletal system than lower forces, such as those involved in short sprints, even though the latter are repeated more frequently (McClay *et al.*, 1994a).

For these reasons, we believed that basketball shoes with ankle support designed to avoid acute injuries, such as ankle sprains, could increase the risk of overload injuries as well as diminish motor performance because they limit ankle joint mobility. The aim of this study was to analyse the effects of increased ankle support on ankle motion and shock attenuation during landing after jumping, and on motor performance in running and jumping.

Methods

Eight healthy male students from the Physical Education Institute of Valencia, who played basketball on a regular basis, were selected. The subjects had previously given their consent to take part in the study.

Two prototype basketball shoes were designed and manufactured for this study. The shoes had identical soles and polyurethane midsoles; the differences between them were limited to the design of the upper. The first shoe was designed to give greater support to the ankle and incorporated a high-top upper, heel counter and a rearfoot control lacing system. The second shoe gave less ankle support with a low-top upper and no heel counter or any other special features (Fig. 1).

Two tests were carried out, the first of which assessed the effect of ankle support on shock attenuation when landing after jumping. The second assessed the effect of ankle support on motor performance, measuring jumping height and the time to complete an obstacle course.

Shock attenuation test

Voloshin and Wosk (1982) measured shock attenuation as the quotient of a lower segment and an upper segment acceleration. Shock transmission can be considered to be the opposite of shock attenuation (i.e. the reduction of the shock wave).

Five subjects took part in the test. The subjects were marked at six anatomical points on their skin (see Fig. 2): one marker was placed at the great trochanter (M1), two at the level of the external femoral condyle

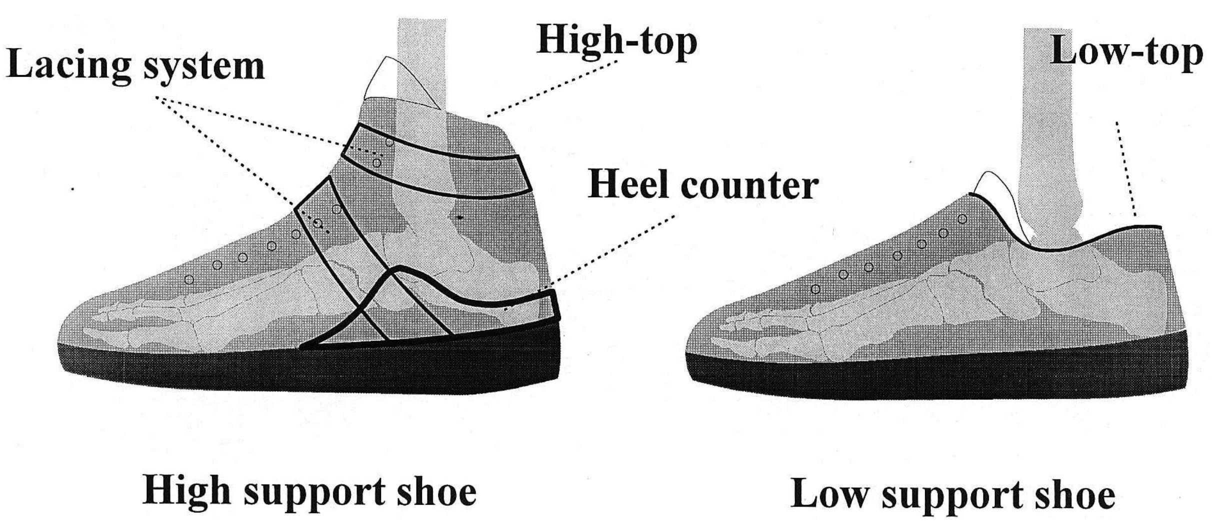

Figure 1 The prototype basketball shoes designed for the study.

(M2 rear and M3 front), two at the level of the head of the fibula (M4 rear and M5 front) and the sixth at the lateral malleolus (M6). Three further points were marked on the shoes, one of which was at the centre of the heel cap at the insertion of the Achilles tendon (M7), one at the centre of the heel cap just above the sole of the shoe (M8), and the third at the external side of the rearfoot on a screw drilled into the sole of the shoe (M9). Nine markers thus determined an anthropometric model of three rigid bodies, comprising the thigh, lower leg and rearfoot.

To measure the shock transmission through the body, an accelerometer (ICSENSORS 3031, piezoresistive, range 20 *g*, resonance frequency 1200 Hz, sensitivity 2.1 mV g^{-1}, mass 0.3 g) was attached to the subject's forehead. A second accelerometer was attached to his right leg, on the proximal anterior surface of the right tibia, 3–4 cm below the tibial tuberosity. The accelerometer, fixed to an aluminium frame, was attached to the skin with double-sided adhesive tape. The mass of the system was less than 2.5 g. An elastic bandage wrapped tightly around the shank was used to fasten the accelerometer and to preload the skin (to reduce movement of the skin along the bone). Using the same frame and an identical elastic bandage, the other accelerometer was fixed to the subject's forehead. The signals from both accelerometers were linked to a telemetry system connected to a computer.

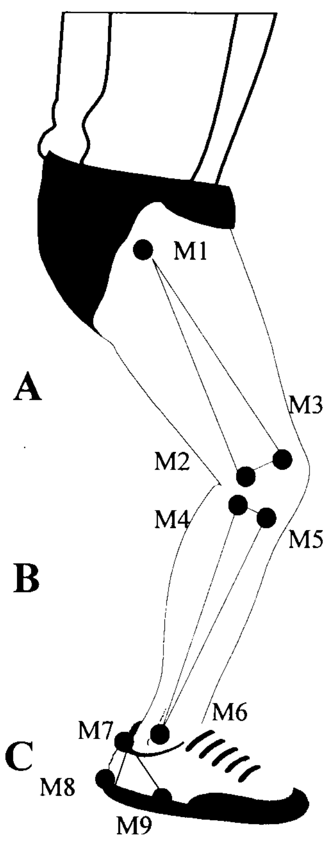

Figure 2 The anthropometric model with nine markers delimiting three rigid bodies: (A) thigh, (B) lower leg and (C) rearfoot.

The ground reaction forces from the five subjects when landing were recorded by a 'Dinascan-IBV' force plate. The sampling rate of the force plate and accelerometer was 1000 Hz, and the sampled signals of both were recorded for further analysis.

Each subject jumped and touched a switch located at the test height with one hand, before landing on both feet, with the right foot on a force plate, thus simulating a rebound action in basketball. The switch was used to electronically trigger the accelerometer and force plate recording, and to establish a lead pulse event for the synchronization of two cameras. The test height was 95% of the maximum vertical jumping capacity of each subject, measured by a mechanical device that records the height reached by the subject's hand, following a maximum effort jump, with a precision of 2 cm.

After a few practice jumps, each of the five subjects performed a total of 27 valid jumps. The first 18 trials were performed wearing the prototype shoes, 9 trials for each prototype, in a randomized sequence and in a series of three repetitions to avoid fatigue. As a reference condition, the last nine jumps were made barefoot. These barefoot jumps were performed at the end of the test, to avoid modification of the normal pattern that would affect the shod condition (Simpson *et al.*, 1988).

The landings of three of the subjects were recorded using two high-speed 16 mm cameras. The film speed was 150 frames per second. The camera axes were aligned at an angle of 60¡. All recorded frames were digitized manually, from the beginning of contact with the force plate until maximum knee flexion. The digitizing equipment consisted of a GP7/GRAFBAR/MARK II sonic digitizer with a NAC model DF-16C analysis projector. After the film was digitized, three-dimensional object coordinates of the markers were calculated by the direct linear transformation method (Abdel-Aziz and Karara, 1971). A typical sequence was digitized three times by each operator to obtain the noise variance (level of precision with which a coordinate is obtained when digitizing manually) for each coordinate of each marker. From these values, the three-dimensional coordinates were smoothed with quintic splines using the 'true mean-squared error' method (Woltring, 1986). After smoothing, the knee flexion angle, plantar flexion angle and ankle eversion angle (see Table 1) were calculated using a joint coordinate system model (Soutas-Little *et al.*, 1987).

A typical forefoot–heel landing pattern was observed in approximately 90% of the jumps. For these landings, two impact peaks were clearly detected in tibia acceleration and ground reaction forces, although in the case of forehead acceleration, only one impact peak was observed. For the statistical analysis of the results, only forefoot–heel landing jumps were considered, and several parameters related to the forefoot and heel impacts on the ground and their transmission to the body were studied (see Table 1).

Performance tests

With the aim of analysing the effect of ankle support on performance, two performance tests were designed, an obstacle course running test and a jumping performance test.

Obstacle course running test. The obstacle course, designed so that the subjects performed similar movements to those which usually occur in basketball games, was similar to that described by Robinson *et al.* (1986), and included forward and backward running, changes in direction of 90¡ and 45¡ to the right and left, and stops (Fig. 3). Photocells were set up at the start and finish of the course to register the time elapsed, with a precision of 0.001 s. Eight subjects participated in this part of the study and were asked to complete the circuit as quickly as possible. After several practice attempts, eight trials were completed in series of two, wearing one of the two prototype shoes in a randomized sequence. Rest periods of 3 min between trials and of 5 min between series were allowed to avoid fatigue.

Jumping performance test. Eight subjects performed 18 maximum counter-movement jumps divided into series of three. Rest periods of 3 min between series and of 30 s between jumps were allowed to avoid fatigue. Each series of jumps was performed with one of the two prototype shoes in a randomized sequence. To standardize the jumps, the subjects were instructed to keep their hands on their hips. The jump height was determined by the time of flight with a 0.001 s precision chronometer connected to a plate on the floor, and under the feet of the subjects. The following formula was used to calculate jump height: $h = \frac{1}{2} g \times (t/2)^2$, where h is the jump height, g is gravitational accelera-

Table 1 Nomenclature

Kinetic variables (Fig. 4)
AT1: first maximum of tibial acceleration (corresponding to forefoot contact)
AT2: second maximum of tibial acceleration (corresponding to heel contact)
MAT: maximum tibial acceleration of AT1 and AT2
FA: maximum of forehead acceleration
FZ1: first maximum of ground reaction forces (corresponding to forefoot contact)
FZ2: second maximum of ground reaction forces (corresponding to heel contact)
MFZ: maximum of ground reaction forces of FZ1 and FZ2
TFZ2 − TFZ1: delay between forefoot and heel impact force peaks
TAT2 − TAT1: delay between forefoot and heel acceleration peaks

There are two types of shock transmission ratios. Some are calculated as the quotient of two acceleration variables, and they express how much of the shock wave is transmitted; others are a quotient of acceleration and force variables, and they express how much of the impact force is measured as acceleration at the level of some body segments. To make the latter quotients non-dimensional, the acceleration measured is multiplied by the subject's mass, and the result divided by the measured force:

AT1/FZ1: ratio of shock transmission of the footprint impact to tibia
AT2/FZ2: ratio of shock transmission of the heel impact to tibia
MAT/MFZ: ratio of shock transmission of the maximum impact forces to tibia
FA/MFZ: ratio of shock transmission of the maximum impact forces to forehead
FA/MAT: ratio of shock transmission of the maximum acceleration of the tibia to forehead

Kinematic variables (Fig. 5)
R1: knee angle at the initial contact with the ground
MR: maximum knee flexion angle
T1: ankle dorsiflexion at the initial contact with the ground
MT1: maximum ankle dorsiflexion
MT0: mimimum ankle dorsiflexion
P1: eversion angle at the initial contact with the ground
PM1: mimimum eversion angle
PM0: maximum eversion angle

tion and t is the flight time. This method of calculation assumes that the height of the jumper's centre of mass above the plate is the same at take-off and landing (Bosco *et al.*, 1983); the error of measurement, when compared with film analysis, has been reported to be – 2% (Komi and Bosco, 1978).

For all the measured variables in the shock attenuation and performance tests, a two-factor analysis of variance (ANOVA) was used, with subject and condition (high support, low support or barefoot) as factors. The alpha level was fixed at 0.05, and a *post-hoc* analysis was performed with Fisher's LSD method, which controls the familywise error rate. Test power was calculated for all the variables (Scheffé, 1959).

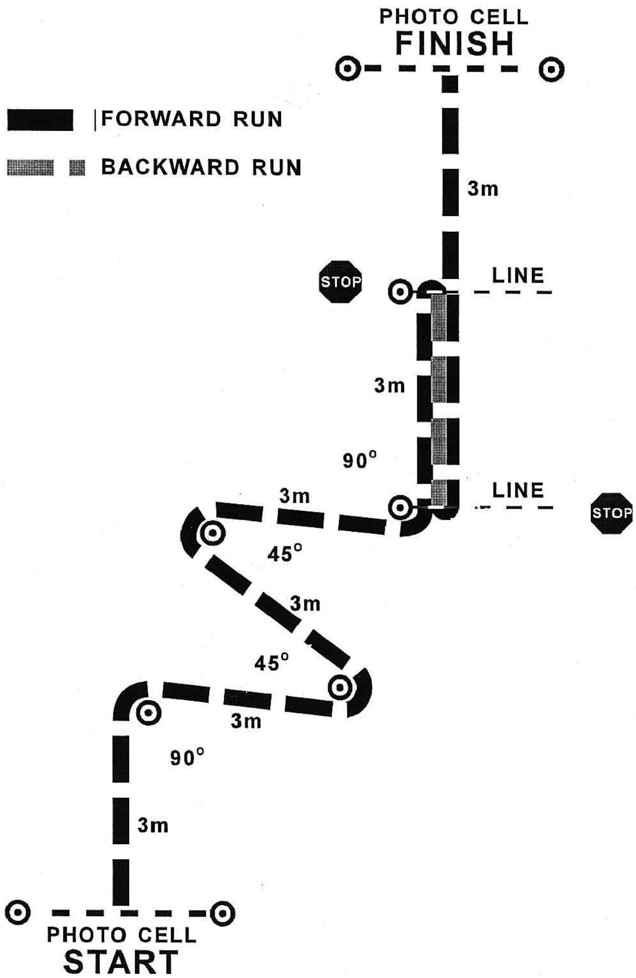

Figure 3 Obstacle course circuit for the running performance test.

Results

Typical ground reaction force and acceleration data are shown in Fig. 4; typical kinematic data are shown in Fig. 5. The results of the kinetic study are shown in Table 2, those of the kinematic study in Table 3 and those of the motor performance study in Table 4. No significant differences were found except for the following conditions and variables.

For the barefoot jumping condition, the forehead acceleration (FA) was found to be significantly lower than with the high-support shoe. An increase in the delay between forefoot and heel impact forces (TFZ2 – TFZ1) was found in this condition in comparison with the low-support shoe. The maximum shock transmission ratio to the tibia (MAT/MFZ) showed lower values for the barefoot condition than for the low-support shoe. Maximum knee flexion (MR) was higher than with the high-support shoe, and the

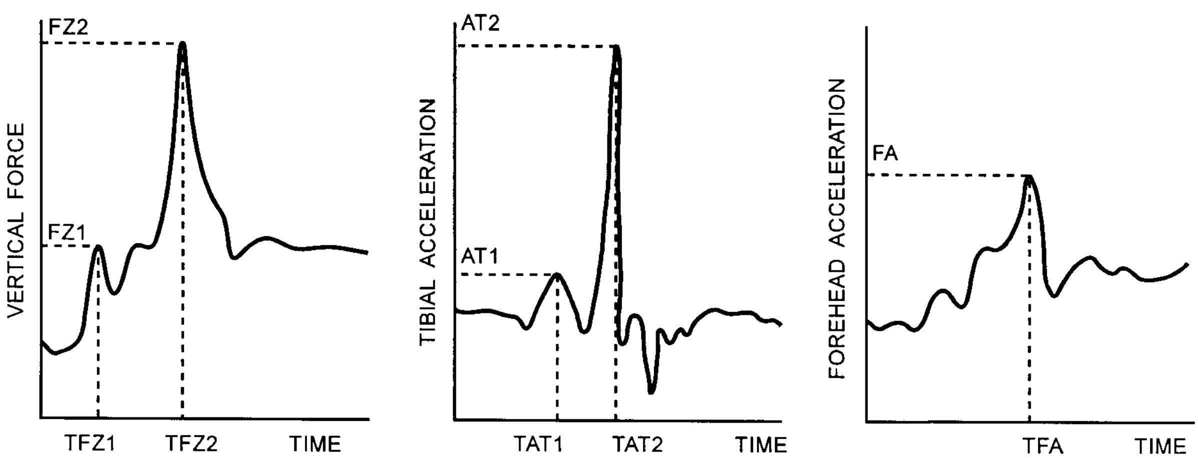

Figure 4 Typical curves and parameters for ground reaction forces and tibial and forehead accelerations. FZ1, ground reaction forces due to forefoot contact; FZ2, ground reaction forces due to heel contact; AT1, tibial acceleration due to forefoot contact; AT2, tibial acceleration due to heel contact; FA, forehead acceleration.

time to maximum flexion (TMR) was longer than for both shod conditions. The barefoot condition showed the highest plantar flexion values at the instant of contact with the ground (T1). Maximum dorsiflexion (MT1) was higher when landing barefoot than in the two shod conditions. The initial eversion angle (P1) in the barefoot condition was close to zero and lower than in the shod conditions. The maximum eversion angle (PMO) was lower for the barefoot condition compared with the low-support shoe. The minimum eversion angle (corresponding to maximum inversion) (PM1) was lower jumping barefoot than for both shod conditions.

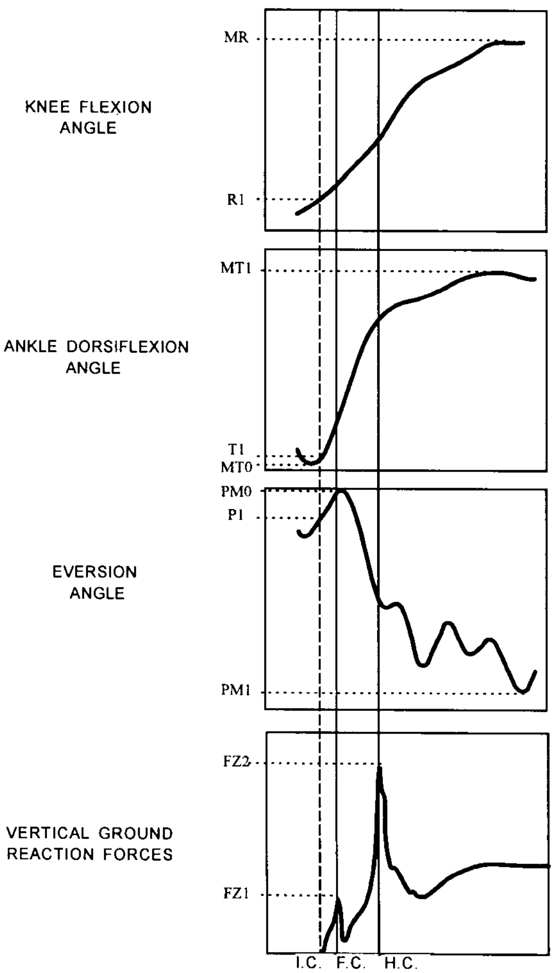

Figure 5 Synchronized graphics of kinematic and ground reaction force variables of the landing. R1, knee angle at the initial contact with the ground; MR, maximum knee flexion angle; T1, ankle dorsiflexion at the initial contact with the ground; MT1, maximum ankle dorsiflexion; MT0, minimum ankle dorsiflexion; P1, eversion angle at the initial contact with the ground; PM1, minimum eversion angle; PM0, maximum eversion angle. I.C., initial contact; F.C., forefoot contact; H.C., heel contact.

For the low-support shoe, there was a larger shock transmission ratio to the tibia (AT2/FZ2) for the heel impact compared with the other two conditions, and larger values of the maximum shock transmission ratio to the tibia (MAT/MFZ) compared with the barefoot condition. The initial eversion angle (P1) and the maximum eversion angle (PMO) were both greater for the low-support shoe compared with the other two conditions. The minimum eversion angle (corresponding to maximum inversion) (PM1) with the low-support shoe was zero (i.e. no inversion), compared with negative values (inversion) for the other two conditions.

With the high-support shoe, the forefoot impact forces (FZ1) were higher than in the other two conditions. The transmission ratio of the maximum acceleration of the tibia to the forehead (FA/MAT) was found to be the largest. The maximun dorsiflexion (MT1), the initial eversion angle (P1) and the maximum eversion angle (PMO) were smaller than for the low-support shoe. The minimum eversion angle (corresponding to maximum inversion) (PM1) was smaller for the high-support shoe than for the low-support shoe, meaning greater levels of inversion. In the performance tests, the high-support shoe was found to reduce the jumping height by 3% ($P < 0.0001$) and increase the time required to complete the circuit by 1% ($P = 0.048$), compared with the low-support shoe.

Discussion

The ground reaction force and tibial acceleration patterns and values during landing in this study (see Fig. 3) were similar to those reported by others (Valiant and Cavanagh, 1985; Gross and Nelson, 1988; McClay *et al*, 1994a), and the parameters obtained from the kinematic study were similar to those reported by McClay *et al.* (1994b) (see Fig. 5).

The increased restriction provided by the high-support shoe was found to limit the ankle joint range of movement in terms of both eversion and plantar flexion. However, and surprisingly, the high-support shoe resulted in larger maximun inversion angles during landing. This could be due to a forced contact of the sole on the ground caused by the increased vertical rigidity of the shoe, while the legs incline laterally during knee flexion.

This higher ankle support, contrary to the results of Sussman *et al.* (1988), increased the impact forces on landing after jumping, possibily because of restricted plantar flexion. However, this difference was seen at forefoot impact but not at heel impact, although the statistical power (the ability to detect differences) was similar (13 and 15% for forefoot and heel impact peak forces respectively). It should be noted that the high-

Table 2 Results of the kinetic analysis of the landing

Variable	P	Mean $\pm s_{\bar{x}}$			Multi-range LSD test	Detectable differences for power >0.80
		Barefoot	Low support	High support		
AT1 (g)	0.1359	6.4 – 0.8	5.9 – 0.8	6.0 – 0.9		0.4 (5%)
AT2 (g)	0.2034	15 – 1	16 – 1	15 – 1		1.0 (8%)
MAT (g)	0.3275	16 – 1	17 – 1	17 – 1		1.0 (6%)
FA (g)	0.0068*	3.2 – 0.4	3.5 – 0.4	3.8 – 0.4	Difference between barefoot and high support	0.5 (15%)
FZ1 (BW)	0.0001*	0.89 – 0.05	0.73 – 0.04	0.83 – 0.05	Difference between low support and the other two conditions	0.11 (13%)
FZ2 (BW)	0.4709	2.8 – 0.2	2.5 – 0.2	2.5 – 0.2		0.4 (15%)
MFZ (BW)	0.4426	2.8 – 0.2	2.5 – 0.2	2.6 – 0.2		0.4 (15%)
FA/MFZ	0.0071*	1.2 – 0.1	1.3 – 0.1	1.4 – 0.1	Difference between barefoot and high support	0.2 (14%)
FA/MAT	0.0388*	0.51 – 0.03	0.52 – 0.04	0.59 – 0.03	Difference between high support and the other two conditions	0.08 (15%)
AT1/FZ1	0.5754	3.5 – 0.5	3.7 – 0.4	3.6 – 0.5		0.7 (19%)
AT2/FZ2	0.0125*	2.13 – 0.09	2.5 – 0.1	2.2 – 0.2	Difference between low support and the other two conditions	0.4 (17%)
MAT/MFZ	0.0204*	2.4 – 0.1	2.7 – 0.1	2.6 – 0.1	Difference between barefoot and low support	0.3 (12%)
TFZ2 − TFZ1 (ms)	0.0118*	60 – 3	52 – 3	55 – 4	Difference between barefoot and low support	9.0 (16%)
TAT2 − TAT1 (ms)	0.3802	36 – 2	32 – 3	35 – 4		12.0 (35%)

Note: All acceleration variables are expressed as multiples of the gravitational acceleration (g) and all variables corresponding to the ground reaction forces are expressed as multiples of the subject's body weight (BW). The impact transmission variables are non-dimensional. Time is expressed in milliseconds. Power = statistical power of the F-test. *$P < 0.05$.

Table 3 Results of the kinematic analysis of the landing, where knee extension, ankle dorsiflexion and ankle eversion values are positive

Variable	P	Mean $\pm s_{\bar{x}}$			Multi-range LSD test	Detectable differences for power $>$0.80
		Barefoot	Low support	High support		
R1 (¡)	0.3639	19 – 2	17.9 – 0.7	17 – 1		3.4 (19%)
MR (¡)	0.0047*	87 – 2	85 – 2	82 – 4	Difference between barefoot and high support	6.0 (7%)
TMR (ms)	0.0035*	271 – 14	236 – 14	227 – 21	Difference between barefoot and the other two conditions	40.0 (16%)
T1 (¡)	0.0000*	-13 – 2	-25 – 2	-28 – 2	Difference between barefoot and the other two conditions	5.0 (24%)
MT1 (¡)	0.0000*	48 – 1	28 – 1	23.7 – 0.8	Differences between the three conditions	3.0 (9%)
TMT1 (ms)	0.3390	209 – 20	199 – 14	188 – 14		60.0 (30%)
MTO (¡)	0.0000*	-14 – 2	-26 – 2	-29 – 2	Difference between barefoot and the other two conditions	6.0 (24%)
TMTO (ms)	0.2679	-5 – 2	-7 – 1	-8 – 1		6.0 (89%)
P1 (¡)	0.0000*	3 – 4	11 – 1	7 – 1	Differences between the three conditions	5.0 (66%)
PM1 (¡)	0.0000*	-19 – 2	0 – 1	-5.4 – 0.8	Differences between the three conditions	3.0 (41%)
TPM1 (ms)	0.0023*	176 – 25	105 – 12	112 – 12	Difference between barefoot and the other two conditions	43.0 (33%)
PMO (¡)	0.0014*	7 – 2	14 – 2	8 – 1	Difference between low support and the other two conditions	5.0 (52%)
TPMO (ms)	0.0163*	237 – 7	10 – 3	9 – 3	Difference between barefoot and the other two conditions	10.0 (9%)

Note: Power = statistical power of the F-test. $^{*}P < 0.05$.

Table 4 Results of the performance tests

Variable	P	Mean $\pm s_{\bar{x}}$		Detectable differences for power >0.80
		High support	Low support	
Counter-movement jump (cm)	0.0000	42.6 – 0.2	43.8 – 0.2	0.76 (1.8%)
Running (s)	0.0482	8.7 – 0.1	8.6 – 0.1	0.1 (1.7%)

Note: Power = statistical power of the F-test.

support shoe had heel counters. Studies of running shoes have shown that heel counters improve heel shock attenuation, probably because of heel soft tissue confinement (Jorgensen, 1990; Ferrandis *et al.*, 1993). The lack of significant differences for the heel impact forces can be explained as a compensation of two opposite effects of increased ankle support and heel confinement; greater ankle support increases heel impact forces but soft tissue confinement decreases these forces. A consequence of this is that shoes with greater ankle support should incorporate materials that improve shock attenuation, particularly in the forefoot.

No differences in tibial acceleration were observed between the prototype shoes with respect to forefoot contact (AT1), heel contact (AT2) or maximum acceleration (MAT). However, the statistical power for these variables provided a greater discriminating power (5, 8 and 6% respectively) than for the other variables in this study. This implies that there were no differences between the prototype shoes regarding these variables, or that such differences were less than the percentages indicated.

The shock transmission ratio of ground reaction forces to the tibia showed no significant differences between the prototype shoes at forefoot contact (AT1/FZ1). This may be because the statistical power for this variable was low, only allowing the determination of differences above 19%. In contrast, for the variable corresponding to heel impact transmission (AT2/FZ2), the differences observed were significant. In this sense, for the prototype shoe with the lower ankle support, force transmission to the tibia was greater than for the high-support shoe. This may be attributed to the soft tissue confinement effect of the heel counter.

The results obtained for impacts transmitted to the head (FA), and the transmission ratios (FA/MFZ and FA/MAT) calculated for the high-support shoe, indicate that the latter allow greater impact transmission to the upper body. Both knee and ankle flexion–extension constitute natural shock-absorbing mechanisms (Gross and Nelson, 1988). Thus our results show greater knee flexion in barefoot landing and greater ankle dorsiflexion during landing when wearing low-support shoes than high-support shoes. Wosk and Voloshin (1985) showed that an increase in the shock attenuation at the spine reduces the incidence of lower back pain. Thus shoes that give higher ankle support are not recommended for players suffering from back pain.

To protect against ankle inversion injuries without diminishing the natural capacity for shock attenuation and performance, the ideal ankle movement control should exclusively limit excessive eversion–inversion without limiting flexion–extension. This study has shown that the high-support shoe increased the magnitude of forefoot impacts during landing and decreased motor performance, probably because of the restriction of the plantar-flexion range of movement.

With respect to motor performance, our results show that higher ankle support reduces player performance both in vertical jumping and in running with directional changes. The running performance tests, as in the study of Robinson *et al.* (1986), showed increased times with increased restriction of ankle movement (1% in the present study and 2% for Robinson *et al.*). However, this decrease in performance does not affect both actions equally; the vertical counter-movement jump was affected to a greater extent than the obstacle course run. The jumping performance results reveal inter-prototype differences of 3% with worse performance for the high-support shoe. This result agrees with that of Burks *et al.* (1991), who found differences between 3.4 and 4.6% in ankle tapped and ankle braced vertical jumping conditions, but does not agree with that of Bocchinfuso *et al.* (1994), who did not find significant differences between using and not using ankle braces, probably because of a low statistical power, even though they reported differences of about 4 and 5%.

It is recommended that players wear high-support shoes when the risk of injury is great because of a higher jump frequency and frequent rebounding. Likewise, such footwear is advisable for players with a history of repeated ankle sprains. Low-support shoes are recommended for players who perform fewer jumps under the basket, generally lighter and shorter individuals who perform more frequent rapid movements and who need all of their performance capacity.

In conclusion, we found that increased ankle support reduces ankle eversion range of motion, increases shock transmission and reduces both jumping and running performance. Protection against ankle sprains is provided by limiting the eversion–inversion range of movement, and both the increase in impacts and decrease in performance are particularly influenced by limitation of ankle flexion–extension. It was not possible to establish the relationship between each element of the shoe and the effects observed. Further research is required to discriminate the individual effects of each element of the high-support prototype shoe. These aspects should also be taken into account both when designing and purchasing basketball shoes.

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