The Impact of Asymmetrical Loading on Tibial Characteristics and Bone Strength in High-Impact Athletes

Asymmetrical Loading and Bone Adaptation in Professional Fast Bowlers.

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Abstract

Background: Asymmetrical athletes produce movements where the external load is unequally distributed in the lower extremities e.g., cricket fast bowling. Loading magnitude is known to have an effect on bone adaptation. It is not understood if tibial characteristics differ between legs when they are exposed to different magnitudes of external load, as happens in asymmetrical athletes. Design: This study aimed to assess the association between external load and tibial characteristics and compare the effect that asymmetrical loading has between legs in asymmetrical athletes. Footballers were recruited as a comparator group. Methods: Inertial measurement units (IMU) were placed at the 14% site of the anteromedial tibia to measure external load during habitual training. Whole body Dual-energy x-ray absorptiometry (DXA) and tibial Peripheral Quantitative Computed Tomography (pQCT) scans were taken of the athlete within 2 weeks of the external load measurement. Results: Asymmetrical athletes experienced 34% higher peak acceleration and 28% higher peak positive acceleration (PPA) in the front planting leg compared to the back trailing leg and showed greater bone mineral content (BMC; 2%) and torsional tibial strength (7%) in the front planting leg. Positive correlations were shown between cumulative load and tibial strength (r=.638; p=.035) in asymmetrical athletes. Conclusions: Exposure to cumulative load showed higher tibial anteroposterior bone strength and transverse and torsional fracture resistance than the lesser loaded contralateral limb. The ability to monitor external load within the applied setting and how it impacts bone can help practitioners estimate the athletes' bone load throughout the season.

Keywords; Acceleration, Bone, Cricket, Load-Bearing, Wearables

Introduction

In some sports e.g., cricket, people experience different magnitudes of loading between limbs. For example, during cricket bowling, the front foot of a medium to fast bowler can experience a peak vertical ground reaction force (GRF) between 2 and 5 body weights higher than the back foot during a delivery stride[1]. These ‘asymmetrical’ sports people are an interesting population to study in the context of adaptations in bone, as the imbalanced movement patterns they habitually perform produce loads that can affect bone characteristics differently. The asymmetrical nature and force of the cricket bowling technique are shown to result in greater torsional and shear forces within the contralateral side of the vertebra. This is shown to result in greater bone mineral density (BMD) and bone mineral content (BMC) within the non-dominant side of the vertebra, which is contralateral to the bowling arm[2]. Load experienced at the lumbar spine, however, may not represent the loading experienced at distal parts of the skeleton, such as the distal tibia. There is only one study known to the present authors that has investigated bone characteristics in relation to the asymmetrical application of load in cricket bowlers. Despite differences in vertical GRF and loading rate between legs[1], there were no bilateral differences in femur bone characteristics[3], which is surprising when load magnitude is shown to be important for an osteogenic response[4].

Asymmetrical sports have previously been studied to examine the effects of divergent load on skeletal characteristics[5-6]. Limb-specific loading of the forearm in professional tennis players has been shown to lengthen the ulna and second metacarpal in the racket arm[5]. Baseball and softball pitchers display greater bone mass, cortical area, cortical thickness and BMC in the humerus of the throwing arm compared to the non-throwing arm[6]. The upper extremity shows the asymmetrical differences are likely caused by the habitual mechanical load experienced in the active arm as no external forces are exerted like in the lower extremity. External forces (*e.g.,* GRF), are prominent in the lower extremities within daily activity therefore understanding the specific influence of isolated mechanical load is more difficult to study due to the external interferences. The aforementioned studies have offered insight into how bone may react to its mechanical environment and how exercise can influence bone adaptation with an isolated bone-load interaction.

Fast bowlers can perform a bowling action up to 324 times per week[7] meaning they are habitually experiencing an imbalanced distribution of load within their lower extremities. This could have an impact on bone characteristics within the lower legs, specifically at the tibia, as the forces will be heightened at the distalmost part of the body. The loading stimulus (*e.g.*, force of action, accelerations) required to optimise bone health is not well established. An insight into how exercise variables (*e.g.*, impact load) relate to bone adaptation is important for elite athletes, as the risk of stress fracture injuries in cricketers has been associated with excessive loading[8]. Knowledge of loading metrics that may contribute to bone adaptation is important for practitioners wanting to prevent or treat bone conditions since they could use this information to prescribe exercise thresholds.

The novelty of studying bilateral differences in tibial bone characteristics of cricket bowlers offers insight into the most distal long bone that is likely experiencing the greatest magnitude of load in the body during the bowling action. How this translates to tibial bone characteristics is unknown and could offer insight into tibial adaptation from external load during habitual high-loaded exercise. Inertial measurement units provide an alternative non-invasive measurement since they can provide site-specific information on acceleration and direction of movement and are practical for assessing external loads during unrestricted exercise[9]. Research suggests tibial acceleration can act as a surrogate measure for impact force experienced on bone and moderate correlations have been shown between accelerations and GRF metrics (*e.g.*, impact peak) and tibial acceleration during lab running[10]. Furthermore, at the proximal tibia, accelerations have been shown to correlate positively with bone circumference and cortical cross-sectional moment of inertia[11].

Assessing sporting activity can provide valuable insights that may not be possible to obtain in a laboratory setting, where exercise is typically more controlled and restricted. By studying athletes in real-world settings, researchers and practitioners can gain a better understanding of how different types of external load and training regimes affect bone health in applied environments. The aim of this study was to assess the association between external load and bone characteristics in cricket fast bowlers. It is hypothesised that cricket fast bowlers will produce a greater external load and show higher bone characteristics in their front planting leg compared to back trailing leg.

Materials and methods

Participants and study design

11 male elite right-armed cricket fast bowlers (aged 24 ± 5 years old, height 1.91 ± 0.08 m, body mass 91.1 ± 12.1 kg, participation 12.9 ± 3.8 years) and 14 right-footed male elite footballers (aged 19 ± 1 years old, height 1.81 ± 0.04 m, body mass 77.0 ± 6.3 kg, overall participation 10.6 ± 2.9 years) recruited from professional clubs via pre-existing professional networks. Each athlete group was used as an independent sample to investigate the differences between legs. Fast bowlers typically bowled 35 overs, or 210 balls, per week whilst footballers typically trained for ~12 hours per week and played competitive matches for 1 - 2 hours per week. ~~In fast bowlers, the front planting leg during a bowl is referred to as the dominant leg and the back or trailing leg is referred to as non-dominant. In footballers, the preferred kicking leg is referred to as dominant and the standing or supporting leg is referred to as non-dominant.~~ The inclusion criteria for the study required participants to be aged between 18 and 40 years old, competing at an elite level (elite being defined as a professional athlete contracted to and competing in their chosen sport), a fast bowler (cricket only) or an outfield player (football only), injury free, not currently taking any medication that influenced bone metabolism and had not received a joint replacement or prostheses. Before taking part in the study, participants completed informed consent, a health screen questionnaire, pre-scan screening and an athletic and injury status questionnaire. Height (Stadiometer, Seca, Hamburg, Germany) and body mass (Seca, Birmingham, UK) were recorded wearing minimal clothing. The study was approved by the ethics committee (Ref 604) and the National Health Service Research Ethics Committee (Ref 260817).

Protocol

The external load was assessed in the fast bowlers whilst they performed six overs (36 balls) at a wicket and batsperson during a pre-season training session at the intensity they would do so during a competitive match. The external load was assessed in footballers whilst they performed a habitual warm-up that replicated the movements performed during match-play (*e.g.*, acceleration/deceleration, change of direction, hopping and jumping). Athletes attended a lab for DXA and pQCT scans within 2 weeks of their loading assessment during training.

External load monitoring

Prior to the activity, IMUs (dimensions 42 x 27 x 11 mm, mass 9.5 grams, operating range 200g; Blue Trident, Vicon Motion Systems Ltd, Oxford, UK), recording at 1600 Hz were secured with a self-adhesive overwrap (Lightpast Pro, Vivomed) to each leg at the 14% site of the tibial length measured from the distal end to match the 14% site of the pQCT scan. Tibial length was measured with a ruler between the medial aspect of the tibial plateau and the medial malleolus. Raw acceleration data were exported into Python (version 3.10) to calculate resultant peak acceleration, resultant peak positive acceleration (PPA), cumulative load and relative load. Resultant peak acceleration was calculated using the three-dimensional Pythagoras' Theorem formula. Resultant peak positive acceleration was calculated as the average of the 10 highest resultant peaks identified. Each step the participants performed was identified as starting when the acceleration surpassed 5 gravitational units[14] with one gravitational unit (g) being equal to 9.81m/s. The resultant peak acceleration found within 90ms of the initial 5 g threshold was calculated during each step. Cumulative load was calculated as the sum of the number of foot plants multiplied by the sum of peak accelerations adapted from the relative load and calculated by dividing the cumulative load by their body mass[15].

Bone characteristics

Dual-energy x-ray absorptiometry (iDXA, GE Healthcare, UK) was used to assess whole body BMD (g/cm^2), BMC (g) and bone area (cm^2). Participants were positioned supine on the scanner bed with their ankles and knees strapped to restrict involuntary movement. The participants lay motionless for the duration of the scan with their arms by their sides. All scans and analyses were performed by the same manufacturer-trained operator to keep the scans consistent. All attenuation materials (i.e., watches, jewellery, glasses, etc) were removed before scanning. If any movement artifacts were present, the image was classified as invalid and repeated. No participants were removed from the analysis due to artifacts.

Peripheral Quantitative Computed Tomography (XCT2000L, Stratec Medizintechnik) was used to assess Trabecular density (g/cm^3) at the 4% site. Stress strain index (SSI) (X (axial anteroposterior bone strength), Y (axial mediolateral bone strength) and Polar (torsional bone strength)), Cortical thickness (mm) and Periosteal circumference (mm) at 14% and 38% site and Cortical density (g/cm^3) at 14%, 38% and 66% site of the right and left tibia. The participant's tibial length was measured to the nearest mm, determined as the medial aspect of the tibial plateau to the medial malleolus. The participant's leg was placed in the scanner with their foot secured in a purpose-built attachment. The leg was aligned with an integral laser and clamped at the knee to restrict movement whilst the participant was directed to remain as still as possible during the scan. A reference point locating the scan was performed to confirm the location of the distal end plate, which acts as a positioning line. Sectional images were obtained at the distal sites (4%, 14%) and the diaphysis of the tibia (38%, 66%) from the positioning line. A voxel size of 0.5mm and slice thickness of 2.5mm was used for all measurements. A contour mode, with a threshold of 180mg·cm3, was used to separate soft tissue and bone. If any movement artifacts (inaccuracies in the measurement caused by motion) were present following the scan, the image was classed as invalid, and a repeat measure was performed. No participants were removed from the analysis due to artifacts.

Statistical analyses

Data were checked for normality of distribution with Shapiro-Wilks tests (IBM, SPSS Statistics, v.29). To determine differences in external load and bone characteristics between legs, paired samples t-tests were performed. To test the differences of change between legs across athlete groups independent samples t-tests were performed. Statistical significance was accepted at the 95% confidence level ($p<0.05$). If data were nonparametric then a Wilcoxon signed Rank test was performed. Pearson correlations were performed to assess correlations between bone characteristics and external load metrics for each leg.

Results

Physical characteristics

There was no difference in leg fat mass between athlete groups. Leg lean mass was higher in the dominant leg of footballers ($p=0.03$) but no difference was shown in lean mass between legs in fast bowlers.

Cricket fast bowlers

Fast bowlers displayed significantly higher resultant peak accelerations and resultant PPA in the front planting leg compared to the back trailing leg (Table 1; $p<0.01$). No differences were shown in cumulative or relative load. Fast bowlers had greater BMC ($p=0.02$) and tibial strength (X and Polar) (Figure 1, Table 2; $p<0.04$) in the front planting leg compared to the back trailing leg. No differences were shown in BMD, bone area, trabecular density, cortical thickness, periosteal circumference or tibial strength (Y).

Table I. Fast bowler's external load metrics

Figure 1. Individual fast bowler bone characteristics between front planting leg and back trailing leg.

Footballers

No differences in any external load metrics were shown between legs (Table 1). Footballers showed no differences in bone characteristics between legs (Table 2).

Table II. DXA and pQCT derived bone measurements from fast bowlers and footballers. pQCT measurements taken at the 4%, 14%, 38% and 66% sites of the tibia.

External load and bone characteristic correlations

In fast bowlers, cumulative load showed moderate positive correlations with SSIX ($r=.683$) and SSIPOL ($r=.638$) at the 14% site of the tibia in the front planting leg ($p=0.035$) (Figure 2). No correlations were shown between any other external load metrics and bone characteristics of the fast bowlers and footballers.

Figure 2. Pearson correlations between external load metrics and bone characteristics within legs of fast bowlers.

Discussion

The present study is the first study to show an association between tibial strength and cumulative load in fast bowlers**.** The association with cumulative load, but not peak load is surprising based upon research demonstrating that high magnitude loading is important for bone accrual[4] and the current study showing a 34% difference in peak acceleration in the fast bowlers' front planting leg compared to the back trailing leg. The mechanisms that drive the bone response to load have shown that high load creates large rates of deformation in the bone matrix, which promotes osteogenesis[16]. Most of the research exploring loading and bone accrual has, however, used rhythmic loading (replication of running, jumping[17-18]), whereas cricket fast bowling has unique loading cycles (run-up, pre-delivery stride, delivery stride, follow through). The intermittent nature of the foot planting cycles may help promote bone adaptation as the use of rest periods has been shown to regain bone mechanosensitivity, with 8 hours of rest being optimal for complete restoration of bone[19]. Fifty loading cycles in a single bout promote osteogenesis with bone becoming refractory to loading cycles beyond this[19]. It could be speculated the intermittent nature of cricket bowling may be used by support staff in cricket to help with bone adaptation. Using the guidelines of bone adaptation to loading[19], it could be possible in cricket to tailor training with optimal loading cycles and recovery time to assess the magnitude of loading and recovery time for optimised bone accrual in the applied setting.

Fast bowlers experienced 1.3x more peak acceleration PPA in the front planting leg compared to the back trailing leg, whereas no differences in external load were shown in footballers (Table 1). This is comparable to other studies that have shown high tibial accelerations are experienced during fast bowling[20] and the action creates a heightened impact force of 5 - 9 times the athlete's body weight when using force plates[1]. In comparison, footballers only generate ~2.5 times their body weight during sprinting[21]. Despite the front planting leg of fast bowlers being associated with greater tibial strength (SSIPOL, BMC) and greater external loading parameters (peak acceleration and PPA), no correlations were shown between these bone characteristics and external load parameters (Figure 2). Furthermore, although this was not a comparative study between groups, we analysed if there was a difference between the legs across the athlete groups. There was a significant difference between the front planting leg and back trailing leg resultant PPA, peak acceleration, BMC and SSIPOL (14%) in fast bowlers, but no differences in external load or bone characteristics in footballers' dominant and non-dominant leg. The technique of cricket bowling creates a large shift in the linear velocity of the centre of mass as the front planting leg produces a braking force when planted[22]. This heightened load may contribute to the stronger bone characteristics shown within the tibia of the front planting leg as a higher magnitude of load initiates a greater osteogenic response[4].

An advantage of the present study is the site-specific measurement of external load and the environment in which it took place. Ideally, a direct measurement of bone strain (*e.g.*, tibial mounted strain gage) during activity would be applied, but this is invasive and impractical, particularly in an elite athlete population[23]. GPS devices are the most used method among practitioners to estimate bone load[24]. Correlations have been shown between total distance, accelerations and decelerations derived from GPS and bone strength characteristics in football players across a season[25], suggesting quantifying load from wearable technology can assist in monitoring external load alongside bone during exercise. However, bone characteristics were assessed at the tibia whereas the external load was assessed in a unit placed on the upper back and therefore is not site-specific. The athletes in the present study were monitored using IMUs during habitual training rather than in a controlled laboratory setting. IMUs can provide information specific to the area under observation[12], in this case, the tibia. Changes in bone mass and geometry are sensitive to change at different sites (anterior, posterior, medial, or lateral) of the same bone. The current study reported the load experienced at the anteromedial distal site of the tibia where loads are observed to be highest[26] which is an improvement on GPS placement where it is not an approximate load transferred through the body[25].

Previous studies have shown that accumulated impact-based loads produce positive adaptations to bone strength and can act as predictors for bone characteristics at the tibia[27-28]. However, no correlation was shown between peak acceleration and tibial strength in the present study. The reasons for this could be 1) only using metrics from IMUs where other external load methodologies and metrics can be used and 2) no segmental analysis being performed from the pQCT scans which would have enabled correlations to be viewed alongside specific areas of the tibia. The lack of correlation in the present study may be due to the placement of the IMUs. The present study placed the IMUs at the 14% site of the distal tibia, therefore, the accelerations and inferred load were only measured at the anteromedial site of the tibia. It may be hypothesised that the loading experienced at other tibial sites (e.g., posterolateral) may differ from the site measured. Tibial accelerations can fluctuate across different locations of the tibia and loading applied across bone does not act uniformly[29]. The action of cricket bowling necessitates linear movement patterns so that bone accrual may occur at a specific location (e.g., anteriorly) in response to the load experienced. Therefore, movement can cause an excessive load on bone in one direction whilst simultaneously unloading the other[17].

Although BMC and torsional tibial strength (polar) were different, there were no differences in BMD, bone area, trabecular density, cortical density, cortical thickness, periosteal circumference, and tibial strength (X and Y) between the front planting leg and back trailing leg of fast bowlers. This may be explained by each leg being habitually exposed to mechanical loading daily. Unlike previous studies[5-6], the observations in the present study were made in the lower limbs, which means the application of habitual loading may have occurred outside of bowling sport-specific actions. This habitual load may create a higher baseline loading threshold for adaptation, whereas the upper limbs do not have any regular exposure to GRF, which has been associated with bone accrual[30]. However, this is not shared by other research that has shown to dissociate from the relationship between GRF and tibial bone load[31].

External load was monitored during a single training session due to the time constraints of the participants, therefore between-session loading reliability was not assessed. IMUs have previously been shown to be a reliable measure of sporting movements[12] and tibial accelerations during fast bowling[13]. No differences in bone characteristics were shown between legs within footballers, although there was a difference in leg lean mass, which may be ascribed to being the kicking foot. Injury history was recorded by the participants during this study, however, no dietary tracking was performed even though it is well-established that diet can influence bone health. As the study aimed to make inter-limb comparisons in the same individuals, diet is unlikely to have significantly influenced the findings. It should however be noted that dietary differences between the fast bowlers and footballers could influence inter-group comparisons. The present study was cross-sectional in nature therefore, it cannot distinguish whether the sport alone or maturation influenced the bone characteristics measured. Longitudinal studies are required to assess the effect of growth and maturation on bone characteristics.

Conclusions

Higher cumulative load may be associated with an increase in anteroposterior and torsional tibial bone strength. This offers applied practitioners insight into how bone accrues to habitual high-loading activity. Using IMUs as a method to estimate bone load and subsequent bone adaptation may offer an alternative solution to using GPS for measuring external load for applied practitioners. It would be insightful to observe the relationship between external load and bone during interventions where loading magnitude differs between groups. This would help to observe the effects of quantified repetitive loading on bone characteristics in human exercise protocols, where groups are habitually exposed to different magnitudes during the same type of activity.

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Authors' contributions

Reece Scott has given substantial contributions to the conception, design, acquisition, analysis and interpretation of the data of the manuscript. Ian Varley has given substantial contributions to the conception and the design of the manuscript. Reece Scott and Ian Varley have participated to drafting the manuscript, Craig Sale, Ruth James and Cleveland Barnett revised it critically. All authors read and approved the final version of the manuscript.

TABLES

Table I. Fast bowler external load metrics

Variable	Fast bowlers			
	Front planting leg	Back trailing leg	**% Difference**	***P***
Resultant peak acceleration (g)	26.9* ± 7.9	17.6 ± 6.1	53%†	.00
Resultant PPA (g)	21.1* ± 6.3	15.5 ± 4.9	38%	.00
Cumulative load	94756 ± 40019	90609 ± 47036	5%	.58
Relative load	1038 ± 469	997 ± 556	4%	.62

Values are represented as mean (±1SD). *depicts a significant difference between the front planting leg and the back trailing leg ($p<0.05$). †depicts a significant difference in change between legs across athlete groups ($p<0.05$).

Table II. DXA and pQCT derived bone measurements from fast bowlers and footballers. pQCT measurements taken at the 4%, 14%, 38% and 66% sites of the tibia.

Variables	Fast bowlers				Footballers			
	Front planting leg	Back trailing leg	% Difference	p	Dominant leg	Non- dominant leg	% Difference	p
DXA								
BMD (g/cm^2)	1.62 ± 0.12	1.59 ± 0.42	1.9%†	.13	1.59 ± 0.16	1.57 ± 0.17	1.3%	.19
BMC (g)	820* ± 94	804± 220	2.0%	.02	738 ± 81	736 ± 81	0.3%	.47
Total bone area (cm^2)	507 ± 43	504 ± 40	0.6%	.20	466 ± 24	469 ± 24	0.6%	.27
pQCT								
4%								
Trabecular density (g/cm^3)	276 ± 22	269 ± 32	2.5%	.13	304 ± 45	298 ± 42	2.0%	.24
14%								
Cortical density (g/cm^3)	1106 ± 14	1100 ± 19	0.5%	.07	1091 ± 18	1094 ± 17	0.3%	.33
Cortical thickness (mm)	3.11 ± 0.46	3.07 ± 0.38	1.3%	.73	3.11 ± 0.39	3.16 ± 0.46	1.6%	.35
Periosteal circumference (mm)	90 ± 7.9	90 ± 6.9	0.0%	.92	88.3 ± 6.1	88.3 ± 5.3	0.0%	.94
SSIX	1405 ± 255	1366 ± 277	2.8%†	.06	1339 ± 199	1315 ± 159	1.8%	.67
SSIY	1441 ± 259	1424 ± 245	1.2%	.65	1404 ± 206	1439 ± 181	2.4%	.35
SSIPOL	2511* ± 470	2337 ± 516	7.0%	.00	2386 ± 352	2408 ± 284	0.9%	.83
38%								
Cortical density (g/cm^3)	1139 ± 17	1136 ± 16	0.3%	.52	1138 ± 20	1142 ± 18	0.4%	.22
Cortical thickness (mm)	6.35 ± 0.79	6.34 ± 0.66	0.2%	.98	6.64 ± 0.35	6.54 ± 0.51	1.5%	.41
Periosteal circumference (mm)	86.9 ± 4.6	84.9 ± 5.7	2.4%	.11	81.7 ± 3.3	81.1 ± 3.0	0.7%	.52
SSIX	1768 ± 295	1657 ± 382	6.3%	.06	1513 ± 171	1473 ± 147	2.7%	.36
SSIY	1500 ± 223	1482 ± 290	1.2%	.53	1346 ± 164	1303 ± 157	3.3%	.35
SSIPOL	2746 ± 326	2619 ± 506	4.6%	.13	2448 ± 305	2424 ± 284	1.0%	.73
66%								
Cortical density (g/cm^3)	1108 ± 20	1104 ± 16	0.4%	.61	1086 ± 18	1098 ± 18	1.1%	.09

Values are represented as mean (±1SD). *depicts a significant difference between the front planting leg and the back trailing leg ($p<0.05$). †depicts a significant difference in change between legs across athlete groups.

FIGURE CAPTIONS

Figure 3. Individual fast bowler bone characteristics between the front planting leg (DL -□) and back trailing leg (NDL -○). Black depicts group mean. SSIX and SSIPOL were assessed from pQCT. BMC assessed from DXA.

Figure 4. Pearson correlations between external load metrics and bone characteristics within legs of fast bowlers. Black dots and red line depict dominant leg. White dots and black line depict non-dominant leg.