Heading in football. Part 3: Effect of ball properties on head response

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SUPPLEMENT

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Objectives: Head impacts from footballs are an essential part of the game but have been implicated in mild and acute neuropsychological impairment. Ball characteristics have been noted in literature to affect the impact response of the head; however, the biomechanics are not well understood. The present study determined whether ball mass, pressure, and construction characteristics help reduce head and neck can impact response.
Methods: Head responses under ball impact (6–7 m/s) were measured with a biofidelic numerical human model and controlled human subject trials (n = 3). Three ball masses and four ball pressures were investigated for frontal heading. Further, the effect of ball construction in wet/dry conditions was studied with the numerical model. The dynamic ball characteristics were determined experimentally. Head linear and angular accelerations were measured and compared with injury assessment functions comprising peak values and head impact power. Neck responses were assessed with the numerical model.
Results: Ball mass reductions up to 35% resulted in decreased head responses up to 23–35% for the numerical and subject trials. Similar decreases in neck axial and shear responses were observed. Ball pressure reductions of 50% resulted in head and neck response reductions up to 10–31% for the subject trials and numerical model. Head response reductions up to 15% were observed between different ball constructions. The wet condition generally resulted in greater head and neck responses of up to 20%.
Conclusion: Ball mass, pressure, and construction can reduce the impact severity to the head and neck. It is foreseeable that the benefits can be extended to players of all ages and skill levels.

Early footballs (soccer balls), developed in 1855, consisted of a rubber bladder and leather covering. The ball construction was a great improvement over previous generations that had used an inflated pig's bladder. Further improvements were made over the following years including better consistency in shape and durability but it was not until 1872 that the spherical shape was made mandatory by the English Football Association. Specifications for both the circumference (686–711 mm) and mass (368–425 g) were included. The official mass was later increased in 1937 to 397–453 g but the remaining requirements were unchanged. These requirements formed the basis for the current specifications in the Fédération Internationale de Football Association (FIFA) *Laws of the Game* specifying a circumference of 680–720 mm and mass of 410–450 g. Inflation pressures of 600–1100 g/cm^2 (0.6–1.1 bar) are also specified along with leather or other suitable cover material.

In more recent years the materials used in ball construction have changed considerably, improving upon the leather exterior with manmade materials. This has helped reduce water absorption and degradation of the balls, which was a problem with the older generation balls. Recent efforts to standardise the balls further were made upon the release of the FIFA Quality Concept (FQC) for Footballs[1] which stipulates the circumference (680–700 mm), mass (410–450 g), shape, size, water absorption (<20%), pressure loss, rebound, durability, and balance. All balls used for FIFA sanctioned match games must comply with the FQC requirements.

Continuing improvements to the balls have resulted in more consistent game play but little attention has been given to the severity of ball impact with the head during heading. With the recent media attention on neurophysiological and neuropsychological dysfunction related to heading, a better understanding of this issue is required. The discomfort of heading older generation balls, especially when wet, is generally acknowledged, but little research has been conducted to quantify the effects of ball impact on head response and the associated risks to the player. Isolated ball impacts onto steel plates and numerical studies have provided some insight but do not elaborate on the aetiology of injury or biomechanics of the impact.[2–5]

To date, there is much controversy on the effects of repeated head impacts where it concerns both the risk of injury after having sustained a concussion and the effects of repeated low level impacts to the head such as in heading the ball. This controversy remains partially due to the difficulty in assessing the severity of concussion and to the unknown effects of repeated head to ball impacts. For many years, recommended practices and guidelines for proper heading techniques have been based on the experience of players and trainers alike but these lacked quantification of the physical event and potential sequelae. Recently biomechanical studies in heading have started to be conducted but must be matured to develop a better understanding of the cause and effect for single and repeated impacts. Once an understanding has been reached, preventive measures could be taken to reduce the mechanical insult to the head and associated risks, if any.

Studies related to ball–head response under different impact conditions have been conducted in parallel with biomechanical studies. These have provided data concerning the peak force exerted to the head.[6–9] Estimates of the severity of the impact were made where peak forces were noted to be as high as 2000 N for ball speeds of 36 m/s.

Impact studies relevant to work herein have involved assessment of the severity of the impact of the ball in relation to the pressure and mass when wet.[2] The tests were carried out by having the ball strike an instrumented flat steel surface. Twenty Size 5 balls inflated to 0.4, 0.6, and 0.8 bar were impacted at a nominal velocity of 9.7 m/s^2. Peak transmitted force, rise time, and impulse were computed. Increases in ball pressure were commensurate with an increase in peak force and impulse. Decreases in the rise time for the force and reduced contact durations were also noted. Balls conditioned in the wet experienced increases in peak force likely due to the increased mass. Ball construction methods were also found to be of interest where sewn assemblies were noted to absorb a greater percentage of water than moulded balls. Sewn balls were also observed to have higher peak forces, rise times, impulse, and contact duration. Although statistically significant differences were found between groups, absolute differences of 4–9% in peak force were obtained. It was suggested that the playing conditions within the game such as ball speed, impact direction, and actual inflation pressure may have an equal or greater influence. However, control of the ball pressure and mass was recommended to reduce the severity of head impact.

A similar test series was conducted on an instrumented plate with 20 balls inflated to 0.6 bar and impacted with a speed of 17–18 m/s^2.[3] Significant differences were found between sewn and moulded balls. Higher values of peak force, rise time, and impulse were noted for the sewn balls although the average ball mass for the sewn construction was higher than for the moulded balls. Concern was expressed about the peak forces as these were in the range of published skull and facial bone fracture levels. However, it should be noted that the study failed to address the issue of load distribution, which would increase the effective load tolerances. Impulse values were also noted to be below published concussion levels but movement of the head towards the ball under game play conditions would increase the impulse further. Ball construction was identified as a factor in the impact response and was suggested as a means for manufacturers to reduce the severity of the impact.

As an alternative approach, numerical simulation methods were employed to study ball to head impacts for children of different ages and variations in ball pressure and size.[5] The model was based on physical principles where the player's head and ball were considered linearly elastic and the contact surface between them flat. The study found that the ball pressure had little or no effect on the head response which included peak impact force, linear head acceleration, head injury criterion (HIC), and contact duration. However, an increase in ball size resulted in greater contact times and an increase in HIC values, which are sensitive to loading duration. The peak forces and accelerations did not change as these are independent of duration. Recommendations were made to use a smaller ball size for children to reduce the head responses.

All the current ball studies indicate that the ball characteristics and impact conditions play an important role in determining the impact response of the head. Changes to the physical ball characteristics such as mass, pressure, construction, and size can help reduce the severity of the impact but further insight is required to understand the relation between these and to determine if suitable guidelines can be developed for use by ball manufactures and regulatory bodies.

The objective of the current study was to gain a better understanding of biomechanics in heading to help determine if head responses can be changed by altering ball properties such as mass and pressure. The present study is a continuation of biomechanical research into heading techniques to determine if changes can be implemented within the player to reduce head impact response.[10] The analysis was extended in the current study to investigate external effects on head response consisting of variations in ball mass, pressure, and construction. The study was designed to investigate the sensitivity of ball characteristics with the aid of a numerical model in an initial effort to determine whether impact reductions are possible, and if so, which parameters are the most significant. The paper is the third of a three part series published in this supplement. The first part, ''Development of Biomechanical Methods to Investigate Head Response'', described analysis techniques and test methodology capable of measuring both kinematic and kinetic responses of the player.[10] The second part, ''Biomechanics of Ball Heading and Head Response'', focused on the implementation of a numerical model and guidelines for reducing head response.[11]

METHODS

We used a validated numerical model and human subjects to quantify the biomechanical response of ball impact under a limited set of heading scenarios. Initial tests were conducted with human subjects ($n = 7$) to obtain a preliminary understanding of heading techniques on biomechanical response and to validate the movements of a 50th percentile male numerical model. Linear and angular head accelerations and neck loads in the midsagittal plane were provided by the numerical model.[11] Subsequently, we used the numerical model to investigate the sensitivity of ball mass, pressure, construction, and condition on head response.

Study of the ball impact characteristics and interaction with the head first required detailed measurement of the ball's dynamic stiffness. We accomplished this with physical impact tests at 7 m/s onto an instrumented hemispherical anvil having a radius of 98 mm, approximating a 50th percentile male forehead. Measured anvil forces and ball deformations obtained by high speed video analysis provided data for assessment of the stiffness. Details of the instrumentation are provided in the first paper of the series along with the implementation of the numerical model.[10] We selected three balls (adidas Tri-Lance, Junior 290, and Junior 350; adidas, Germany) to investigate ball mass. The pressures investigated in the study were 0.4, 0.6, 0.8, and 1.1 bar for the adidas Tri-Lance ball. Both the mass and pressure ranges exceeded the guidelines provided by the FIFA *Laws of the Game* and may not be suitable for official matches.

The numerical model was exercised under a number of different ball mass and pressure characteristics and for a series of old and new generation balls, all Size 5. We conducted the tests with the low speed heading scenario (LS2; see table 1 in reference 10 (Part 1) for details of the heading scenarios) under which the model was validated. The LS2 heading designation was used and represents the ball targeted 5.5 m down in front of the player with a forehead impact. Further details are given in Part 1.[10]

The range of trials and ball characteristics are detailed in table 1. The ball stiffness in loading represents the measured dynamic stiffness required for use with the numerical head and neck model. The ball radius was also fixed to match the size in the simulation but this did not affect the interactions between the head and ball since the impact characteristics were determined from the experimental tests with ball and anvil. The mass moment of inertia values were calculated assuming a thin outer shell.

Verification of the simulation results was carried out with human subject trials ($n = 3$, one repeat) conducted at the same time as for the analysis of heading techniques described in Part 2.[11] A single heading scenario (LS2) was again selected to study the effects of ball pressure and mass. Details of the subject trials are presented in table 2.

The subjects were instrumented to measure linear and angular acceleration at the mouth with an intraoral device. Kinematic measurements of the head and torso were also recorded with high speed video but are not reported here as the ball characteristics are studied and not heading

Table 1 Details of parameters investigated with the numerical model

Ball type	Ball condition	Radius (mm)	Mass (kg)	Mass moment of inertia (kg-m^2)	Pressure (bar)	Stiffness in loading (kN/m)
Fevernova Tri-Lance (baseline)	Dry	110	0.444	0.0021	0.8	33.6
Fevernova Tri-Lance	Dry	110	0.444	0.0021	1.1	39.1
Fevernova Tri-Lance	Dry	110	0.444	0.0021	0.6	31.1
Fevernova Tri-Lance	Dry	110	0.444	0.0021	0.4	17.7
Fevernova Junior 290	Dry	110	0.299	0.0014	0.8	35.8
Fevernova Junior 350	Dry	110	0.351	0.0017	0.8	34.8
Fevernova Tri-Lance	Wet	110	0.458	0.0022	0.8	39.5
Mitre White	Dry	110	0.428	0.0021	0.8	25.9
Mitre White	Wet	110	0.603	0.0029	0.8	32.8
adidas World Cup 1974	Dry	110	0.439	0.0021	0.8	43.7
adidas World Cup 1974	Wet	110	0.534	0.0026	0.8	30.2
adidas Santiago (Orange)	Dry	110	0.445	0.0022	0.8	36.7
adidas Santiago (Orange)	Wet	110	0.463	0.0022	0.8	38.9
adidas Santiago (Brown)	Dry	110	0.412	0.0020	0.8	37.4
adidas Santiago (Brown)	Wet	110	0.604	0.0029	0.8	31.0

Table 2 Details of human subject trials

Heading scenario	Ball speed	Ball modification	Ball target	Heading code
Passing	Low, 6 m/s	Baseline, 444 g, 0.8 bar, Fevernova Tri-Lance	Front, down, 5.5 m from player	LS2 (baseline)
Passing	Low, 6 m/s	Low pressure 0.6 bar, Fevernova Tri-Lance	Front, down, 5.5 m from player	LS2-P2
Passing	Low, 6 m/s	high pressure 1.1 bar, Fevernova Tri-Lance	Front, down, 5.5 m from player	LS2-P3
Passing	Low, 6 m/s	Low mass 299 g, Fevernova Junior 290	Front, down, 5.5 m from player	LS2-M3
Passing	Low, 6 m/s	Low mass 351 g, Fevernova Junior 350	Front, down, 5.5 m from player	LS2-M2

biomechanics. All measurements were taken in the mid-sagittal plane. The detailed methodology of subject measurement is reported in Part 1.[10]

RESULTS

Numerical simulations

The results of the head and neck responses for the numerical simulation are presented in table 3. The peak resultant head responses are relative to the centre of gravity of the head, and the neck loads are taken at the occipital condyles. Linear accelerations in the range of 11–18 g (107–180 m/s^2, 1 g = 9.81 m/s^2) were obtained for the various ball conditions. Angular accelerations were in the chin down direction in the range of 286–446 rad/s^2. The neck loads relative to the top of the neck have rearward shear values of 279–433 N and axial compressive values of 497–608 N.

Power values were based on the calculation of the head impact power (HIP) index as described in Part 2.[11] These varied according to linear/angular head acceleration and velocity. Values in the range of 1.04–1.58 kW were predicted.

The ball incoming and outgoing velocity ratios are provided, demonstrating small differences between tests. The ball speed increase is a result of the head approach velocity which was constant throughout the tests. The differences can be attributed to the variations in ball response.

Table 3 Results from the numerical simulations

Type	Max. linear accel. (m/s^2)	Max. angular accel. (rad/s^2)	Max. total power (kW)	Max. neck shear (N)	Max. neck axial (N)	Ball exit velocity ratio
Baseline Tri-lance	156	−374	1.44	362	−570	1.68
Tri-lance (0.4 bar)	107	−286	1.04	329	−539	1.71
Tri-lance (0.6 bar)	150	−363	1.38	359	−566	1.69
Tri-lance (1.1 bar)	170	−402	1.53	368	−574	1.68
Ball Weight (290 g)	130	−293	1.21	279	−497	1.71
Ball Weight (350 g)	140	−330	1.30	307	−532	1.70
Tri-lance Wet	173	−410	1.55	376	−578	1.67
Mitre White Dry	132	−330	1.24	341	−551	1.70
Mitre White Wet	174	−446	1.55	433	−608	1.65
adidas World Cup 1974 Dry	180	−419	1.58	369	−574	1.67
adidas World Cup 1974 Wet	158	−402	1.44	398	−582	1.67
adidas Orange Dry	164	−390	1.49	366	−573	1.68
adidas Orange Wet	170	−404	1.53	376	−579	1.67
adidas Brown Dry	161	−379	1.46	348	−561	1.68
adidas Brown Wet	168	−439	1.51	429	−608	1.65

Accel, acceleration; Max., maximum.

Table 4 Relative differences from the numerical simulations

Type	Max. linear accel.	Max. angular accel.	Max. total power	Max. neck shear	Max. neck axial	Ball exit velocity ratio
Baseline Tri-lance	0%	0%	0%	0%	0%	0%
Tri-lance (0.4 bar)	−31%	−24%	−28%	−9%	−5%	1%
Tri-lance (0.6 bar)	−4%	−3%	−4%	−1%	−1%	0%
Tri-lance (1.1 bar)	9%	7%	6%	2%	1%	−1%
Ball Weight (290 g)	−17%	−22%	−16%	−23%	−13%	1%
Ball Weight (350 g)	−10%	−12%	−9%	−15%	−7%	1%
Tri-lance Wet	11%	10%	8%	4%	2%	−1%
Mitre White Dry	−15%	−12%	−13%	−6%	−3%	1%
Mitre White Wet	11%	19%	8%	20%	7%	−2%
adidas World Cup 1974 Dry	15%	12%	10%	2%	1%	−1%
adidas World Cup 1974 Wet	1%	7%	0%	10%	2%	−1%
adidas Orange Dry	5%	4%	4%	1%	1%	−0%
adidas Orange Wet	9%	8%	7%	4%	2%	−1%
adidas Brown Dry	3%	1%	2%	−4%	−2%	−0%
adidas Brown Wet	8%	17%	5%	19%	7%	−2%

Accel, acceleration; Max., maximum.

Table 5 Summary of kinematic data from human subject trials

Heading scenario	Head angle (deg)	Head velocity angle (deg)	Ball velocity angle (deg)	Head/ball velocity angle (deg)	Head velocity (m/s)	Head/ball velocity (m/s)
LS2	−18	−7	58	−65	3.2	8.0
LS2-M2	−16	−7	58	−64	2.5	7.6
LS2-M3	−20	−13	59	−72	2.7	7.2
LS2-P2	−23	−10	55	−65	2.8	7.9
LS2-P3	−19	−10	56	−66	3.0	7.8

A comparison of the head responses is presented in table 4 relative to the baseline heading scenario, LS2 with the adidas Fevernova Tri-Lance ball. This will be the same approach taken with the subject trials where intersubject variations were taken relative to a baseline test (LS2).

The relative changes for the linear acceleration head responses range from a decrease of 31% to an increase of 15% for all ball types and conditions. Similarly, the angular acceleration ranges from a decrease of 24% to an increase of 19%. For head impact power a 28% decrease to a 10% increase was observed. Neck loads in fore–aft shear or axial compression were predicted to decrease by up to 23% and to increase by 20% under the various conditions. Ball velocity ratios of incoming and outgoing speeds varied minimally from the baseline test with a decrease of 2% to an increase of 1%.

Subject trials

Kinematics

We measured subject kinematics with reference markers on the head and torso. This provided us the ability to measure the orientation of the head relative to the laboratory reference as well as the relative angles between the head and ball. A single heading scenario was chosen for the analysis, yet some variation in the kinematics was noted.

The average head kinematics are detailed in table 5 showing a 7° difference in average head orientation between the ball trials for the three subjects. This resulted in different approaches with the ball and when combined with the ball trajectory variations, differences of 8° between the head and ball were observed. The speed of approach with the ball was found to be in the range of 7.2–8.0 m/s.

Kinetics

The results of the numerical study showed that the ball mass, pressure, and condition can potentially decrease the head/neck responses from 23% to 31%. Under the investigated conditions, head response increases of 10% to 20% were also predicted. These findings warranted verification with the human subjects under similar conditions.

Verification of the numerical simulation findings was accomplished with a small sample of subjects (n = 3). Ball mass and pressure effects on head response were measured with instrumented subjects. Results of the trials are presented in table 6. The average peak head linear accelerations measured at the intraoral device ranged from 14 g to 18 g (140–175 m/s^2) and the average peak angular acceleration ranged from 1.23 krad/s^2 to 1.61 krad/s^2. The values represent the accelerations seen by the head including the approach kinematics. The summation of linear and angular head impact power terms resulted in a range of 216–412 W. Power was based on the added power to the head from the point of impact. The subject minima and maxima values as well as the sample standard deviations are presented in table 6. Relative comparisons of the head response with the baseline response (LS2) are presented in table 7. These are based on the average differences in each of the subjects. Average differences for the ball conditions ranged from a 10% decrease to a 12% increase in linear acceleration, an 11% decrease to 15% increase in angular acceleration, and decreases of 0% to 43% in HIP.

DISCUSSION

Ball impact characteristics

Impact characteristics of the ball were first established with ball impacts against an anvil approximating the average shape of the forehead. The force–deflection characteristics were used to provide dynamic stiffness values for use with the numerical model to establish the differences in ball mass and pressure, and to investigate the differences in response across balls of different vintage and hence construction.

Comparison of our results with previously published data is not possible due to differences in ball model, impact speed, and impact anvil surface shape. Previous studies presented

Table 6 Summary of human subject data

No of subjects	Scenario	Peak linear acceleration (m/s^2)				Peak angular acceleration (krad/s^2)				Head impact power (W)			
		Min.	Max.	Avg.	SD	Min.	Max.	Avg.	SD	Min.	Max.	Avg.	SD
3	LS2	155	158	156	2	1.24	1.77	1.47	0.27	274	502	412	121
3	LS2-M2	117	159	140	21	1.02	1.58	1.23	0.31	206	233	216	14
3	LS2-M3	124	157	142	17	1.46	1.50	1.47	0.03	167	279	217	57
3	LS2-P2	128	166	141	21	1.09	2.33	1.60	0.65	221	324	261	55
3	LS2-P3	139	216	175	38	1.08	2.58	1.61	0.84	240	560	380	164

Min., minimum; Max., maximum; Avg., average.

Table 7 Relative differences between subject trials

No of subjects	Scenario	Peak linear acceleration				Peak angular acceleration				Head impact power (W)			
		Min.	Max.	Avg.	SD	Min.	Max.	Avg.	SD	Min.	Max.	Avg.	SD
3	LS2	0%	0%	0%	0%	0%	0%	0%	0%	0%	0%	0%	0%
3	LS2-M2	−25%	3%	−10%	14%	−40%	29%	−11%	36%	−58%	−15%	−43%	24%
3	LS2-M3	−21%	−1%	−9%	10%	−17%	21%	3%	19%	−56%	−25%	−40%	15%
3	LS2-P2	−19%	7%	−10%	15%	−23%	88%	15%	64%	−53%	18%	−29%	41%
3	LS2-P3	−11%	36%	12%	23%	−31%	86%	13%	63%	−52%	24%	−2%	43%

See table 6 for abbreviations.

Table 8 Published ball impact response characteristics

Reference	Test condition	Maximum force (N)	Maximum impulse (Ns)	Contact time (ms)
Armstrong *et al*[2]	Rigid flat plate 9.6–9.7 m/s Dry, wet 0.4, 0.6, 0.8 bar for dry 6, 9 m/s for wet 20 ball models	512–558*	8.3–8.6*	11.6–12.4*
Levendusky *et al*[3]	Rigid flat plate 17–18 m/s 0.6 bar for dry 20 ball models	851–912*	12.4–13.7*	10.2–10.8*
Naunheim *et al*[12]	Rigid head form on neck 9, 12, 15 m/s Dry 0.4, 0.6 bar 1 ball model	637–1212†	N/A	N/A

*Avergae values of ball groupings
†Calculated from head mass and acceleration.

data as the loads transmitted to either a rigid support surface or a partially constrained head form (table 8). This is in contrast with the current study; we have reported the loads transmitted to the subject's head (simulated and human) to assess the effect of ball characteristics.

Ball mass and head response

We investigated the effects of ball mass on head and neck responses with the new generation balls, Fevernova Tri-lance Size 5 and two lighter balls (the Junior 290 and Junior 350). Although we investigated ball mass, it is likely differences in other dynamic properties, such as construction and stiffness, may affect head impact severity. Dynamic tests have shown that the Junior 350 and Junior 290 are 3.5% and 6.5% stiffer than the baseline ball, respectively. The corresponding mass decreases are 21% and 35% from the baseline ball having a mass of 0.444 kg.

The results of the simulations are depicted in fig 1 for the head peak resultant linear acceleration and linear power term. A linear relation can be observed for both measures of impact severity. In addition, the head angular acceleration and neck shear/axial loads showed similar trends indicating that a lighter ball mass provides a net benefit for all measures of head/neck response (see table 3). A reduction up to 23% is possible for all head/neck responses with the lightest ball but it must be cautioned that the mass is not the only influential parameter due to the different balls used. A comparison of the mass effects for different ball models is shown in fig 2 for all balls tested, including old generation balls both in the dry and wet conditions at standard pressure (0.8 bar). It is

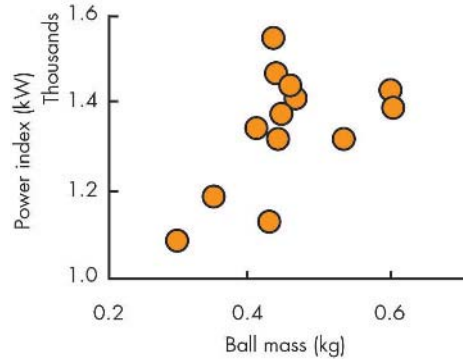

Figure 2 Effect of ball mass for old and new generation balls.

apparent that there is some but limited correspondence between mass and head response.

For the human trials, the peak linear accelerations decreased by 9% to 10% with a decreased ball mass of 20% to 32%, respectively. Analysis of the kinematic data shows an overall decrease in magnitude of the acceleration response without significant difference in the contact duration. When combined with the observed reduction in head velocity a net reduction in the power index of 2–29% was observed. The effect of ball mass on peak angular acceleration is not clear due to contradictory results obtained with the two lighter balls. Analysis of the angular acceleration response in general showed that the measurement was susceptible to high variability.

Ball pressure and head response

We used the Fevernova Tri-lance Size 5 ball to investigate the effects of ball pressure on head and neck responses. A single ball type was chosen to reduce the influence of other factors such as ball construction. Four ball pressures were chosen representing a 25% and 50% decrease (0.6 and 0.4 bar, respectively) and an increased level commensurate with that used in competition matches, an increase of 38% (1.1 bar).

The simulations were conducted with the baseline heading configuration and the relative changes in responses are depicted in fig 3. A net benefit in head and neck responses occurred for the lower ball pressures. A corresponding detriment was noted for the higher ball pressure and can be attributed to the increased ball stiffness. For the range of ball pressures changes the corresponding change in dynamic stiffness of the ball was −47% to +16% for the lowest and highest pressures.

The head responses were reduced up to 31% with a ball pressure reduction of 50%. Little reduction (<5%) was noted for ball pressure reductions less than 25% where a 1% reduction in head response was noted for a 4% reduction in

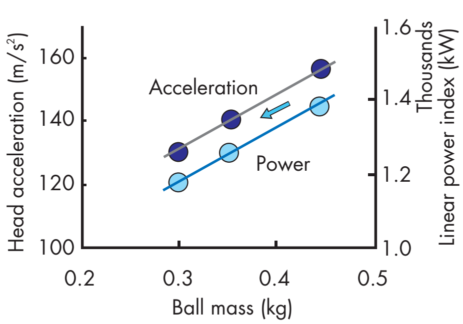

Figure 1 Effect of ball mass on predicted head response.

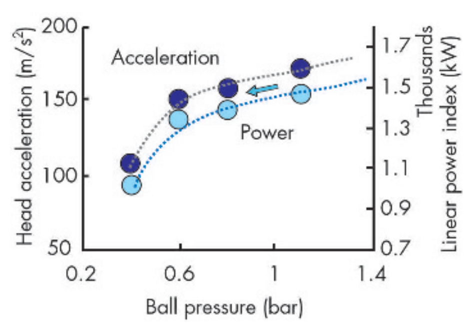

Figure 3 Effect of ball pressure on predicted head response.

ball pressure. For ball pressure reductions greater than this, a 1% reduction in head response is noted for a 1.2% reduction in ball pressure indicating that there is some threshold after which greater benefits are realised. Similar observations were noted for head angular accelerations and head impact power calculations involving angular terms. This suggests that both linear and rotationally induced loading mechanisms are also addressed with reductions in ball pressures.

Reductions in the axial neck compression responses were noted for changes in ball pressure reductions less than 25% and much greater benefits for reductions above this. Similar trends were observed for neck shear loads in the fore–aft direction suggesting that the overall loading to the neck structure is reduced for reductions in ball pressure.

In the human subject trials, decreases in peak linear accelerations up to 10% were observed for a ball pressure decrease of 50% (see table 7). Analysis of the kinematic data showed an overall decrease in linear acceleration response with an increase in the contact duration. This resulted in similar, but slightly lower, head velocity change. Both of these factors have the net effect of reducing the head impact power by 29%. The angular accelerations were increased in all cases of change to the ball pressures and may be due to the associated variability with angular acceleration measurements.

Ball stiffness and head response

A preliminary investigation of dynamic ball stiffness was carried out as it was expected that an increase in stiffness would correspond to an increase in head response for equal ball mass. The predictions from the simulations are presented in fig 4 for new and old generation balls in the dry and wet condition at standard pressure (0.8 bar). A moderate increase was found with increasing stiffness although other ball characteristics likely resulted in the poor correlation.

Ball condition on head response

A series of simulations were conducted to investigate the relative effects of ball condition on head and neck loading for a series of old and new generation balls. Conditioning of the wet balls was carried out by submersing the ball for three hours and letting them drip dry for three minutes prior to testing. The percentage of mass increase due to water absorption varied considerably and, in some cases, exceeded the current FQC specifications (table 9). Note that some balls were clearly worn and the conditioning method deviated from the FQC specification.

Overall, the lighter balls result in lower neck loads (see table 4), however, a corresponding change in the head responses with the ball mass was not apparent. This is probably due to differences in ball construction and dynamic behaviour under impact. The overall differences in neck responses were in the order of 5% and those for the head were in the 10–15% range. All changes in head responses were consistent whether linear/rotational accelerations or head impact power were used.

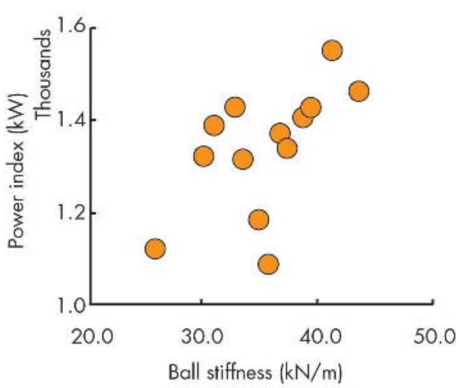

Figure 4 Effect of ball stiffness on head response.

Table 9 Water absorption of different ball models

Ball type	Condition	Pressure (bar)	Mass (kg)	Water uptake (%)
Fevernova Tri-lance	Dry	0.8	0.444	3
	Wet	0.8	0.458	
Mitre White	Dry	0.8	0.428	41
	Wet	0.8	0.603	
adidas World Cup 1974	Dry	0.8	0.439	22
	Wet	0.8	0.534	
adidas Santiago Orange	Dry	0.8	0.445	4
	Wet	0.8	0.463	
adidas Santiago Brown	Dry	0.8	0.412	47
	Wet	0.8	0.604	

The head and neck responses increased in the wet condition due to the water uptake but again there was no direct correlation of the responses with either ball mass or stiffness. Some balls increased in stiffness when wet whereas others became less stiff. As with the dry balls, it can be noted that ball construction, materials, and dynamic properties all have an effect on the responses. The changes in neck and head responses with wet balls were less than 20%. In terms of head responses, both linear accelerations and power rankings were consistent with each other. The relative change in ball condition (dry *v* wet) varied considerably among ball types. Some of those that resulted in the lowest head response in the dry condition had the highest response in the wet condition.

SUMMARY

The effects of ball mass and pressure were investigated under a single heading scenario. A qualitative summary of the impact severity measures is provided in table 10 relative to the baseline low speed configuration. A large effect (indicated by "+++" or "−−−" for positive or negative benefits, respectively) is approximately related to a relative change of 20% or greater.

A reduced ball mass resulted in decreases for peak linear acceleration and head impact power. This can be attributed to the lower energy transferred to the head. The angular acceleration response changes were not consistent for the various ball masses and may be attributed to variability in the ball speeds, subject techniques, and sensitivity of the measurement method. However, a mild reduction is observed for the lowest ball mass.

Ball pressure reductions provided lower peak linear accelerations and head impact power values. The lower pressure resulted in larger contact times with the head providing reduced peak accelerations and velocity change.

Table 10 Summary of head responses for various ball characteristics

Ball type	Linear acceleration	Angular acceleration	Power index
Baseline mass, pressure	=	=	=
Lowest mass	++	++	+++
Low mass	++	−	+++
Low pressure	++	−−	++
High pressure	−−	−−	+*

*Data exhibited large test variability.

What is already known on this topic

It is known that ball mass, pressure, and other physical characteristics influence the severity of impact to the head as the ball is the source of impact. Previous studies have shown some benefit in reducing the mass and pressure of balls but biomechanical considerations are lacking.

What this study adds

The study has shown that ball mass and pressure reductions can reduce head impact severity. Other ball characteristics (stiffness and construction) also govern the ball's response and must be considered in conjunction with traditional parameters of mass and pressure. Changes in ball characteristics can have equal or greater effect than heading technique changes with potential for widespread implementation.

The moderate increase in angular acceleration response for the lower ball pressures was observed but again exhibited large fluctuations. An increase in ball pressure resulted in higher peak linear and angular accelerations. The small reductions in head impact power were inconsistent with the numerical model and are likely due to test variability.

For the human trials, differences in the speed of approach between the ball and head resulted in greater energy difference, and hence, change to the head response. Further, differences in head angle and ball trajectory at impact may result in head response changes due to the force application angle and perhaps contact point.

CONCLUSIONS

Our results present a number of options to reduce impact severity to the head, neck, and both. One alternative, noted in Part 1 of the series, is to improve the heading biomechanics through improved technique. This has the benefit of reducing some aspects of the head responses while providing greater player skills for controlling the ball, and perhaps greater appreciation of the game.

In the review of the ball characteristics and literature, a second alternative becomes evident. A reduction in ball mass and pressure results in improvements to both the peak linear accelerations and power index. The magnitudes of these changes are equal or greater to those observed with changes in heading technique. Tests conducted with various ball types led us to conclude that ball mass, pressure, and stiffness are insufficient to describe the head impact severity. It is likely that ball construction also plays an important role but this would need to be addressed in future studies.

The use of improved ball characteristics provides an attractive approach due to the immediate effect on all heading scenarios, regardless of the player's skills. The possible detriment in playability and handling characteristics of the balls with reduced mass and/or pressures will have to be weighed carefully against the possible safety benefits. The ultimate objective would be to preserve the ball playability while simultaneously reducing the impact severity during heading. Additional constraints provided by the FIFA *Laws of the Game* and the FQC for footballs will need to be reviewed before changes might be implemented for match balls.

The numerical simulations investigated changes to one ball parameter at a time and did not reflect potential heading technique changes that the player may employ to compensate for reduced ball mass or pressure. The potential changes require further investigation, but note that the rebound velocity from the simulation was within 2% for all cases suggesting that large changes to technique may not be required.

To conclude, this three part study has provided a greater understanding of heading biomechanics. Although several recommendations related to heading techniques and ball characteristics were identified, they will require thoughtful implementation. The inconsistencies in the results relating to modified heading techniques will make this a difficult approach to justify; however, we have shown that changes to the ball characteristics provide an overall benefit and can be effectively implemented. A more inclusive study with a larger sample, greater variety of heading scenarios and more comprehensive measurement capability will help increase our awareness and enhancement of heading biomechanics.

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