Head impact response to simulated ball-to-head collisions

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Abstract

Softball is a relatively safe sport, but, as with all sports, the risk of injury remains. Many of the rare severe injuries which occur on the softball field involve ball-player contact. To mitigate the risk of these injuries, modified softballs have been developed with greater compliance than their standard counterparts. The purpose of this work was to investigate the difference between standard softball models approved for use in slowpitch play and two modified softball models in ball-to-head collisions. A finite element model of ball-to-head impacts was developed from a leading softball model and the Total Human Model for Safety (THUMS) developed by Toyota. Simulated softball models included a standard slowpitch softball, a cork-core softball and Reduced Injury Factor (RIF) softball. Both frontal and lateral impacts were simulated for initial ball velocities ranging from 26.8 m/s to 53.6 m/s. The stiffness of cork-core and RIF softballs were 39\% and 23\% of the standard softball model stiffness. In the case of lateral impacts, differences in maximum bone stress were as small as 16\% and 30\% on average for cork-core and RIF balls, respectively, while in the case of frontal impacts differences in maximum bone stress were as large as 57\% and 74\% on average. These results suggest that modified balls may be a viable approach for limiting skull fracture injuries for ball-to-head impacts in some regions, but that these balls may be less effective in others.

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1. Introduction

Serious injuries on the softball field are rare, but, nonetheless do occur (Tolbert et al. 2011). These injuries often occur when other players contact other players, bases or the ball. Between the 1992-1993 and 2003-2004 seasons,
ball-player contact accounted for 23.5% of all injuries sustained in U.S. collegiate fastpitch play (Marshall et al. 2007). Similarly, a retrospective study of slowpitch play in the U.S. Air Force reported that ball-player contact was responsible for 20% of recorded injuries (Burnham et al. 2010).

For collegiate softball, as well as many amateur leagues, the performance of approved softballs is regulated. These regulations are designed to preserve the integrity of the game by maintaining the balance between defense and offense and ensuring consistency in play and player experience over time. Different ball types are available for use in slow and fastpitch leagues, but still other ball types are available which are designed to be more compliant than standard ball types.

These “safety-modified” balls have been created for use by younger athletes who may be more susceptible to injury, those still be developing the skills associated with the game, or other groups interested in reducing injury risk during play. It has been suggested that such balls may reduce the risk of catastrophic injury in games like softball (Link et al. 2002; Burnham et al. 2010). While these modified balls are more compliant than the standard softballs used in league play, the magnitude of the effect this difference has on player safety is unclear. Understanding how safety-modified balls perform in ball-player collisions is important for guiding future safety-modified ball designs.

Recently, sophisticated finite element models of balls used in various sports have been developed. For example, a model of golf ball-club collisions was developed by Tanaka et al. using hyperelastic and viscoelastic material models (Tanaka et al. 2012). Similarly, finite element models have been developed to capture the dynamic performance of hockey and soccer balls (Ranga & Strangwood 2010; Price et al. 2006). Models of dynamic softball performance are particularly challenging due to the large magnitude of energy dissipated during collisions occurring under play-like conditions (Smith & Duris 2009).

To investigate the relative performance of safety-modified softballs in ball-player collisions a finite element model of ball-to-head impacts was constructed. Head impacts were chosen to simulate a worst-case scenario. Simulations consisted of a leading softball model (Burbank & Smith 2012) and the Total Human Model for Safety (THUMS) developed by Toyota. Head impacts with a softball were simulated for both frontal and lateral impact locations, and over a range of velocities. These simulations were conducted with models of high (H; standard slowpitch softball), medium (M; cork-core softball), and low (L; reduced injury factor softball) stiffness ball types. Estimates of peak force (PF) and maximum stress in the skull were obtained from the simulations and used to evaluate the relative performance of the safety-modified balls.

2. Methods

2.1. Ball modeling

Three softballs, one approved for adult slowpitch play and two safety-modified balls, were tested according to ASTM F2845. The procedure described in the ASTM standard measures PF, dynamic ball stiffness (DS) and the cylindrical coefficient of restitution (CCOR) at deformation rates and magnitudes representative of play. Data collected from these tests were used to develop a finite element model of each ball.

Ball models were developed in the LS-DYNA finite element code (Version 974m LSTC, Livermore, CA) using the low density foam material model (Mat #57). The behavior of this material model is characterized by the compressive response loading curve and parameters which control the unloading behavior and viscous effects.

The stress-strain relationship in loading for the standard slowpitch ball was determined from dynamic tests performed on core samples of the polyurethane ball, and unloading parameters were selected based-on dynamic whole-ball tests conducted in accordance with ASTM F2845. Models of these tests were developed in LS-DYNA and used to perform parameter identification. For each safety-modified ball model the loading curve and unloading parameters were adjusted to achieve best fit of force-displacement data obtained from whole ball tests. Fitting was performed over speeds of 26.8 m/s, 42.5 m/s and 53.6 m/s to account for rate effects (Burbank & Smith 2012).
2.2. Head modeling

Human head response to impact with each ball type was simulated using the 50th percentile adult male Total Human Model for Safety (THUMS AM50), a finite element model developed by Toyota and implemented in LS-DYNA. THUMS is an anatomically detailed model which includes the scalp, facial and masseter muscles, component boney structures of the skull, and both white and grey matter within the brain among many other structures. Cortical and cancellous components of the skeletal structures were represented in THUMS using shell and solid elements, respectively (Kimpara et al. 2006).

While validation of such a complex human model is difficult, biofidelity of the THUMS head and neck response to frontal impacts was assessed by comparing simulation results to data from several cadaveric studies, including three head impact tests and two flexion and cervical axial compression tests. THUMS displayed good agreement with experimental data in most comparisons (Kimpara et al. 2006).

2.3. Initial conditions and simulation outputs

Both frontal and lateral ball impacts were simulated using the THUMS head model. For frontal impacts, the initial trajectory of the softball center of mass was normal to the coronal plane and 0.059 m superior to the nasion (51% of the vertical distance to the vertex). For lateral impacts, the initial trajectory of the softball center of mass was normal to the sagittal plane and passed through the anterior aspect of the temporal bone near the sphenoid bone (Figure 1). Impacts were simulated at both locations, for each ball type, and for initial ball velocities ranging from 26.8 m/s to 53.6 m/s in increments of 4.47 m/s. All musculature within THUMS was inactive, and the simulation results represented a completely passive response to impact.

For each simulation, impact force and the resulting von Mises stress in the cranial bone element were output as a function of time. From this data peak impact force and maximum bone stress for each collision was determined.

2.4. Data analysis and bone strength

For simulations of whole-ball testing, ball model performance was compared to experimentally observed PF and CCOR. For ball-player collision simulations, PF and maximum bone stress were compared to assess the effect of simulated ball compliance on head impact response.

Simulated head response metrics were related to bone strength data available in the literature. For cranial cortical bone under quasi-static loading conditions mean ultimate stresses of 73.8 to 96.5 MPa have been observed (McElhaney et al. 1970). Bone is a viscoelastic material, however, and when loaded dynamically (2.5 m/s) strengths of 123.12, 133.61 and 126.91 MPa have been observed for samples from the right and left parietal bones.
and frontal bone, respectively (Motherway et al. 2009). Therefore, simulation results were compared to a bone strength of 125 MPa.

3. Results

3.1. Ball model accuracy

The finite element model of ball H demonstrated excellent agreement with experimental force-displacement data and CCOR measures. The evaluation of this model is described in detail elsewhere (Burbank and Smith, 2012). Safety-modified ball models (balls M and L) also demonstrated excellent agreement with experimental force-displacement data (Figure 2), though agreement was more modest with respect CCOR (Table 1). All models were expected to demonstrate comparable fidelity in ball-player impact simulation, which is governed by contact forces.

![Figure 2. Comparison of force-displacement data from experimental observations (dashed) and simulation of ball tests (solid) for ball H (blue), M (green) and L (red) balls. The data depicted are from impacts onto a cylindrical impact surface at 42.5 m/s (95 mph).]

Table 1. Observed and simulated PF and CCOR for the H, M and L ball types for impacts at 42.5 m/s.

| Ball Type | Empirical Observation | Model Performance | % Difference |
|-----------|-----------------------|-------------------|--------------|
|           | PF (kN)  | CCOR  | PF (kN)  | CCOR  | PF  | CCOR  |
| H         | 18.4     | 0.376 | 18.0     | 0.381 | -2.1% | 1.1% |
| M         | 11.2     | 0.394 | 11.4     | 0.347 | 1.2%  | -8.5% |
| L         | 9.17     | 0.401 | 8.69     | 0.367 | -5.2% | -11.9% |

3.2. Peak impact force

For both impact locations and all ball types, peak impact force was directly proportional to initial ball velocity and relative ball compliance (Figure 3). Impact forces generated by the M and L ball models were between 31.3% to 38.9%, and 45.8% to 56.2% of those generated by ball H, respectively. Average reduction in PF across all impact locations and conditions was 32.5% and 51.5% for simulations with the M and L balls, respectively.
3.3. Bone stress

Maximal stresses were observed in the inner table of the frontal bone for frontal impacts, and at the medial aspect of the base of the zygomatic arch for lateral impacts. For both impact locations and across all initial ball velocities the M and L ball models produced bone stresses ranging from 12.3% to 60.1% and 21.1% to 76.5% of those resulting from contact with the H ball model, respectively (Figure 4). Average reductions of maximal bone stress across all impact conditions were observed to be 36.5% and 52.0% for the M and L ball models. For lateral impacts the average reductions was 16.1% and 31.1% for M and L balls, while the average reduction for frontal impacts was 56.9% and 73.9%.

Despite comparable impact forces between lateral and frontal impacts, for all initial ball velocities maximal stresses in lateral impacts were larger than those produced in frontal impacts. This is due to the lower head stiffness under lateral loading compared to frontal loading, and in line with prior findings of lower force at fracture in cadavers under lateral loading relative to frontal loading (Yoganandan et al. 1995; Yoganandan & Pintar 2004).

Balls are routinely hit at speeds exceeding 42 m/s. The results in Fig. 4 suggest that modified balls may be successful at preventing skull fraction for frontal, but not lateral impacts. Skull fracture is not the only criterion that is affected by ball compliance, however. After impact, compliant balls tend to take longer to return to their original
spherical shape. The persistent deformed shape of a compliant ball in flight can complicate successful fielding. Thus, non-standard balls have the potential of reducing injury severity, but may increase injury frequency. These risks should be carefully balanced before advocating changes in ball design.

4. Summary

This investigation compared PF and maximal bone stresses in ball-to-head impact simulations using one standard slowpitch and two compliant, safety-modified softball types. A decrease in ball DS of 61% (M) and 77% (L) resulted in up to a 39% and 56% reduction in PF, and up to a 60% and 77% reduction in maximal bone stress, respectively, for some impact conditions.

This work suggests that ball properties can have a large impact on player safety on the field. Further investigation of ball-to-head contact will be required to better understand the relationship between ball compliance and brain injuries. Additionally, more work is required to better understand how significant reductions in ball stiffness like those investigated here may influence the integrity of the game.

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