Study on Impact Loading and Humerus Injury for Baseball *

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Abstract

In the United States and Japan, baseball is a very popular sport played by many people. However, the ball used is hard and moves fast. A professional baseball pitcher in good form can throw a ball at speeds upwards of 41.7m/s (150km/hr). If a ball at this speed hits the batter, serious injury can occur. In this paper we will describe our investigations on the impact of a baseball with living tissues by finite element analysis. Baseballs were projected at a load cell plate using a specialized pitching machine. The dynamic properties of the baseball were determined by comparing the wall-ball collision experimentally measuring the time history of the force and the displacement using dynamic finite element analysis software (ANSYS/ LS-DYNA). The finite element model representing a human humerus and its surrounding tissue was simulated for balls pitched at variable speeds and pitch types (knuckle and fastball). In so doing, the stress distribution and stress wave in the bone and soft tissue were obtained. From the results, the peak stress of the bone nearly yielded to the stress caused by a high fast ball. If the collision position or direction is moved from the center of the upper arm, it is assumed that the stress exuded on the humerus will be reduced. Some methods to reduce the severity of the injury which can be applied in actual baseball games are also discussed.

Key words: Baseball, Impact Strength, Stress Wave, Biomechanics, Finite Element Method, Sports Injury, Damage Evaluation

1. Introduction

With the recent increase in spare time, the increased popularity of sports is a phenomenon involving enthusiasts from every generation. As a result, sports related injuries have also increased. Next to rugby and soccer, baseball has the high rate of injury incidence⁽¹⁾. The ball used in baseball is hard and moves fast. Professionals, and some senior high school baseball pitchers in good form, can throw a ball at upwards of 41.7m/s (150km/hr). If a ball at this speed hits the batter, serious injury is quite likely. Thus, baseball has been categorized as a dangerous sport. It is therefore hoped that the mechanisms of

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causing injuries and suitable preventative measures in baseball can be made clear. Among injuries in baseball, there were comparatively many studies concerning head wounds conducted in the field of biomechanics ⁽²⁾⁻⁽⁴⁾. However, there have not been many studies, which examine a reaction load with a baseball or any examinations on injuries except for the head ⁽⁵⁾.

In this study, a collision experiment involving a wall and a baseball were performed, and the dynamic characteristics of the ball were measured. We then made finite element models that simulated these characteristics. With the models, a collision simulation involving the upper arm of a human is analyzed. The reaction force of the ball elucidates the force exerted on the humerus and soft tissues. Additionally, preventative measures that seem to reduce the incidence of injury or damage caused by a ball hitting the batter are examined.

2. Finite element model of a baseball

2.1 Static compression experiment of a baseball

The material composition and dimensions of the cross section of a baseball are shown in Fig. 1. From the figure, the baseball consists of four materials which are cork, rubber core, woolen yarn (Gray and white wool) and cowhide. However, the thing which acts as unified dynamic behavior that is when external force acted on ball oneself of a ball is important. The baseball is compressed in a static compression examination as shown in Fig. 2(a), and load-displacement curves are obtained. On the other hand, a finite element model of the ball is made, and compulsion displacement is given to the ball. In addition, the same pressure as in an actual compression examination is analyzed. Commercial finite element analysis software (ANSYS, ver.9.0) was used for this analysis. Here, the material properties (equivalent Young's modulus E_s and Poisson's ratio ν_s) of the ball model an unknown parameter and agreed with the load-displacement curves best. We changed two unknown parameters, and the material properties, E_s and ν_s were identified.

Figure 1 Material composition of a baseball

(a) Experiment

(b) FEM

Figure 2 Compression test of a baseball

Figure 3 Load-displacement curves of a baseball

These experiments and finite element analysis are shown in Fig. 3. From the results, it can be seen that the analysis with the utility is sufficient even if the baseball is generally regarded as an objective consisting of just one material. The estimate values by analysis were $E_s = 23\text{MPa}$ and $\nu_s = 0.3$. Analysis afterward showed that these values were used for material properties of the hard-baseball (baseball) which was a static.

2.2 Collision experimental equipment and experimental method

A new type of "intelligent" pitching machine has already been developed that independently controls the three parameters of speed, pitch type and direction of the ball by the authors ⁽⁶⁾. The machine has three rollers and two actuators (horizontal and vertical angles), the motion of which is motion controlled by hierarchical neural network (NN).

A no turn ball (without a spin rate) was pitched from this machine as shown in Fig. 4, and the ball impacted the surface of a target equipped with measurement gauges at a right angle to measure impact force. The impact force measurement equipment consists of a 10mm thick square steel plate measuring 400mm. On the back side of this plate, three road cells are installed. This measurement equipment is connected to a personal computer (PC) over a dynamic strain measuring instrument, and the impact force is measured. The experiments for impact force employed both a standard hard-baseball (baseball) and a soft-baseball (rubber-baseball). Also, ball velocities of 19.4 and 30.6m/s for inbound velocity V_i were tested. The point of impact of the ball and the target were filmed using a high speed video camera (MEMRECAM fx-K3, made by Nak Co., Ltd.), which was focused at a vantage point just beside (X direction) the target at a frame rate of 5.0kHz. Image processing was subsequently done. From the images captured, the ball's inbound

Figure 4 Diagram of experimental set-up

(a) Baseball (Hard)

(b) Rubber-baseball (Soft)

Figure 5 Maximum deformation of baseballs at $V_i=19.4\text{m/s}$

velocity, V_i just before impacting on the target, the ball's outbound velocity, V_o after impact, contact time Δt of the ball and steel plate, and the maximum volume of deformation δ_{max} of the ball upon impact were all calculated. Also, the impact force F_z of Z direction in a collision was derived from the total value of the three road cells measured. Additionally, the coefficient of restitution (COR, e) of a ball was calculated using the equation below:

$$e = \left| \frac{V_o}{V_i} \right| \quad (1)$$

2.3 The Results of the collision experiment

The collision of a standard hard-baseball (baseball) with the steel plate (inbound velocity $V_i=19.4\text{m/s}$) is shown in Fig. 5(a). The results for a rubber type baseball (rubber-baseball) under the same conditions are shown in Fig. 5(b). From both figures, even if V_i for both a baseball and a rubber baseball is equal, it can be confirmed that the volumes of deformation and contact times differ greatly. For this reason, it is supposed that there are differences such as mass (baseball is 145g, rubber baseball is 135g), materials and coefficient of restitution (COR) in both balls, respectively.

The experimental and analytical results (mentioned later) of all baseballs are shown in Table 1. These experiment values show the mean of several experiments. In addition, a time history reply of the impact force, F_z of Z direction at $V_i=30.6\text{ m/s}$ is shown in Fig. 6. From these results, if the velocity of the ball is greater, the maximum volume of deformation and the impact force correspondingly become greater. On the other hand, contact time with the target and ball are shorter with the higher velocity. The impact force is greater for the baseball than the rubber-baseball when projected at the same velocity, but the rubber-baseball has a greater maximum deformation.

Figure 6 Time history of impact force, F_z of a baseball (Hard) and a rubber-baseball (Soft) at $V_i=30.6\text{m/s}$

Table 1 Experimental and analytical values of the baseball (Hard)

	Condition 1		Condition 2	
	Experiment	FEM	Experiment	FEM
Inbound velocity, V_i (m/s)	19.4	19.4	30.6	30.6
Outbound velocity, V_o (m/s)	-10.2	-9.23	-13.1	-12.64
Coefficient of restitution, e	0.526	0.476	0.428	0.413
Maximum deformation, δ_{max} (mm)	5.0	5.50	8.1	8.24
Contact time, Δt (ms)	0.81	0.80	0.69	0.71
Impact force, F_{zmax} (kN)	3.98	5.85	9.28	12.05

Figure 7 Viscoelastic model using three elements

Figure 8 FE models for dynamic analysis

2.4 Dynamic properties of a baseball

In this study, the dynamic properties of a baseball which express stress and relaxation are shown as a viscoelastic model consisting three elements (comprised of two springs and one dashpot) such as the one shown in Fig.7. In this viscoelastic model, a time dependent shear modulus $G(t)$ is formulated for time-dependant responses in the follownig equation ⁽⁷⁾.

$$G(t) = G_{\infty} + (G_0 - G_{\infty}) e^{-\beta t} \quad (2)$$

where G_{∞} is a relaxed shear modulus, G_0 is a instantaneous modulus, K is a bulk modulus and β is a decay constant. Here, G_{∞} can be calculated from E_s and ν_s (ref. section 2.1). However, G_0 and β are unknown parameters. Those G_0 and β that provided δ_{max} , contact time Δt and the outbound velocity V_o seem to agree, and are identified by the collision experiments.

2.5 Collision simulation and the results

The target and baseball of the finite element models (FE models) are shown in Fig. 8. The analysis conditions are the same as the collision experiment. A ball with no spin is pitched at the steel plate. The inbound velocity of the hard-baseball (baseball) was $V_i = 19.4$ and 30.6 m/s. The target of the analysis models was arranged at the axis of three road cells installed on the back of the steel plate, and an axial backward end was completely restricted. The dynamic finite element analysis software (ANSYS/ LS-DYNA, ver.9.0) was used for this analysis.

The analytical and experimental results are shown in table 1. In addition, a time history reply of the impact force, F_z of Z direction at $V_i = 30.6$ m/s is shown in both results in Fig. 6, respectively. From table 1, δ_{max} and Δt agree well with $V_i = 19.4$ and 30.6 m/s. Conversely, V_o , e and the maximum impact force F_{zmax} were different from the experimental and analytical

values. Additionally, from the time-history reply for the impact force of a baseball (see Fig. 6), the first peak value and its time of compression impact force occurs early in the collision, about -10kN and 0.85ms are agree with the experiment and analysis values, respectively. Furthermore, the time that changes in the tensile wave occurs by a reflection from the compression shock wave (period), and it is characteristic to see the second peak of the tensile wave at about 2.6ms. These phenomena are recognized both together.

As for the results of the collision experiment and analysis, we think that finite element models for the baseball are sufficient enough for only a certain number of qualitative analyses. Finally, the estimated material properties in the baseball models were $G_0 = 46.15\text{MPa}$ and $\beta = 7000\text{s}^{-1}$.

3. Simulation of collision with the human body

With the material properties of the baseball estimated in the previous chapter, the collision with the living body (the human body) and the baseball was analyzed. In the sport of baseball, the head and upper arms are the locations where injury and bone fracture most commonly occur ⁽¹⁾. Thus, we have decided to analyze the upper arm in this study. To remain objective, a finite element model of the upper arm is made, and bone fracture risks are examined by analysis.

3.1 Finite element models of upper arm

In this study, the finite element model of the upper arm were based on the man walker model that Toyota Central R&D Labs., Inc. developed as a human body model (Total Human Model for Safety, THUMS). THUMS is the computer simulation model that can be seen in Fig. 9(a), and shows simulated dynamics characteristics of the human body such as

Figure 9 Finite element models of upper arm and human body

Table 2 Material properties of upper arm and baseball

Material properties	Cancellous bone (Elasticity)	Cortical bone (Elastoplasticity)	Muscle and Fat (Viscoelasticity)	Skin (Shell element) (Elastoplasticity)	Baseball (Viscoelasticity)
Density, ρ (kg/m ³)	1000	2000	1000	1000	757
Young's modulus, E (MPa)	40.0	8 000	-	2.2	-
Poisson's ratio, ν	0.3	0.3	-	0.4	-
Tangent modulus, E_t (MPa)	-	4 400	-	0.01	-
Yield stress, σ_y (MPa)	-	84	-	6.86	-
Instantaneous modulus, G_0 (MPa)	-	-	7.012	-	46.15
Relaxed shear modulus, G_∞ (MPa)	-	-	2.337	-	8.85
Bulk modulus, K (MPa)	-	-	45.92	-	100
Decay constant, β (s ⁻¹)	-	-	100	-	7000

Figure 10 FE models on Y-Z cross section and points A, B and C

strength of bone, flexibility of skin, in addition to the shape of the whole human body⁽⁸⁾. Fig. 9(b) shows, the upper arm segment of THUMS consisting of a humerus, soft tissue and skin. However, some elements of the bone are actually coarse in the model in order to evaluate the risk of bone fracture of the humerus in detail. Thus, in Fig. 9(C), a new finite element model, which consisted of the smaller element of the bone (cancellous and cortical bones), soft tissue comprising of muscle and fat, and skin, was made and attempted to show the humerus. Here, it was thought that the measurement ratio and materials structure of the bone and soft tissue are important in order to create a general model of the upper arm in this collision analysis. This analysis model measures each lap length from the top to the bottom end of the upper arm, and can be seen in Fig. 9(b). Using both sections of the humerus and soft tissue, the circumference and lap length seem to be similar.

Next, the material properties of the made upper arm model will be described. The humerus consists of the cancellous and cortical bones. The cancellous and cortical bones are modeled for elasticity and elastoplasticity, respectively. Furthermore, the muscle and fat are modeled in a viscoelasticity of three elements, the same as the baseball. These are all the material properties using values of THMUS. The skin, on the other hand, used a shell contact element of elastoplasticity, that is necessary to consider contact. The material properties of the skin used values of a study by Tanaka et al.⁽⁹⁾. Each material property of the upper arm and the baseball used for the analysis are shown in Table 2.

Figure 10 shows the Y-Z cross section in the analysis model of the upper arm and the baseball. The points A, B and C are analyzed in the figure. As for the boundary conditions of the upper arm, when they were taken without just evading the baseball, the top edge surface into the humerus was completely restricted. As for the contact conditions, the dynamic coefficient of friction, μ between the baseball and skin was had the general value, 0.4 of skin in a dry state. In addition, the baseball's velocity, V_i before the impact had three distinct conditions ($V_i=19.4, 30.6$ and 41.7 m/s). The baseball collided at a right angle and at the center of the upper arm model with no-spin (without turn). This analysis model was the standard model, Model-I.

3.2 The analysis results and consideration in Model-I

One example of the analysis results, a compression principal stress distribution and a contact stress distribution of the soft tissue when a maximum compression principal stress (minimum principal stress) occurred in a Y - Z cross section at $V_i=41.7\text{m/s}$ in Model-I, and is shown in Figs. 11(a) and (b), respectively. Additionally, Fig. 12 shows the time-history reply of the maximum tensile principal stress (maximum principal stress, $S1$) and maximum compression principal stress (minimum principal stress, $S3$) occurring at analysis point A, B and C. The first part of the impact, is understood to have a compression stress wave ($S3_C$) that occurs at point C, and the compression stress wave ($S3_B$) occurring at point B occurs a little slower (about 0.4ms). The next impact state, being slightly slower (about 0.1ms), the tensile principal stress ($S1_A$) occurred at point A. From these results, it is predicted that the central part of the upper arm around the collision side occurs at the stress state which is said to be near the bending stress of a beam. After impact, it is understood that the stress wave becomes the tensile or compression wave within reflects an end, that spreads while damping in the whole upper arm repeatedly.

In the collision analysis, the maximum principle stress occurs at the other side of the collision and in the circumference neighborhood of the cortical bone (point A). The value of this is 84.2MPa . This value is equivalent to yield stress σ_y of a humerus (an adult man is about 84MPa), and it may be said that this presents the danger of a bone fracture⁽⁹⁾. However, we think that naturally the analysis value changes into the greatest stress value possible if the boundary condition changes. From point C in Fig. 12, the damping of the stress wave of muscle and soft tissue shows that the tensile wave does not appear as strong in comparison with the bone. It is still unclear whether soft tissue is an important factor showing that the peak value and continuance time for the first of the compression stress wave are related to injuries.

The maximum and minimum principal stress values of the humerus that occurred at the points A and B in Model-I for each velocity of each ball are shown in Fig. 13. In this figure, the analysis models (Model-II, III) are shown simultaneously. It is understood that the absolute value of each stress value increases with the increase in the velocity.

3.3 The analysis results and consideration in Model-II and III

The throw of an actual baseball has various kinds of breaking balls. In order to simulate this, the ball collides with the upper arm of a standard model (Model-I) with a spin (clock turns direction in Fig. 10). This model the same as Model-II, and the spin rate ω of the ball is supposed to be 1800 min^{-1} relating to the general value of a fastball in baseball⁽¹⁰⁾. Additionally, other analysis conditions are similar to Model-I.

One example of the analysis results in Model-II, a compression principal stress distribution and a contact stress distribution of the soft tissue when a maximum compression principal stress (minimum principal stress) occurred in a Y - Z cross section at $V_i=41.7\text{m/s}$ in Model-I is shown in Fig. 14(a) and (b), respectively. When Model-II is compared with Model-I without a spin (see Fig. 11(a)), there is no significant difference in figures. However, the whole stress distribution becomes the form that moved slightly above. As for the contact stress distribution in Fig. 11(b), the peak stress value does not change. The first occurrence of peak stress, and subsequent high stress, is concentrated on a center around the contact surface in Model-I. On the other hand, in Model-II, the second occurrence of peak stress, and subsequent high stress, is distributed over the whole contact surface in depth.

The maximum and minimum principal stress values of the humerus that occurred at the points A and B for each velocity of each ball in Model-II (with Model-I) are shown in Fig. 13. Model-I and Model-II are almost equal in value in several velocities. From these results, it is understood that the impact stress depends on the collision velocity of a ball, while the

Figure 11 Stress distribution in Y-Z cross section at $V_f=41.7$ m/s in Model-I (without spin ball)

Figure 12 Time-history of maximum and minimum principal stresses (S1, S3) at points A, B and C at $V_f=41.7$ m/s in Model-I

Figure 13 Relation between principal stress of the humerus and velocity at points A and B in Models-I, II and III

Figure 14 Stress distribution in Y-Z cross section at $V_f=41.7$ m/s in Model-II (with spin ball)

Figure 15 Top view of FE models in Model-III

Figure 16 Principal stress distribution in X - Z cross section at $V_i=41.7\text{m/s}$ in Model-III

impact stress does not depend on the spin of the ball. In the case of a ball with a spin, the value of peak stress does not change without spin, while the high area of the stress occurs widely on a contact surface. From these results, it can be assumed that when a wound occurs to the soft tissue, the area of the wound becomes wider.

Furthermore, in Model-III, as shown in Fig. 15, collision of a ball when moved 50mm up the X -axis from the center of the upper arm, is shown. This collision model is equivalent to the upper arm, and the ball collided with only half of the one side. This case is compared to when the upper arm is hit directly by a ball (in Model-I). Figure 16 shows one example of the analysis results, a compression principal stress distribution when a maximum compression principal stress occurred in a Y - Z cross section at $V_i=41.7\text{m/s}$ in Model-III. From this figure, the maximum compression stress changes because of the ball and the collision angle with the upper arm part is understood. In addition, the maximum tensile stress occurred in the humerus circumference of the other collision side (results not shown). This is the same result as the collision in the center of the upper arm (Model-I). The stress distribution spreads through the whole upper arm model. Additionally, when Model-III is compared with Model-I, it is understood that about 30% of the values of the maximum and minimum principle stress are decreasing in Fig. 13.

By the analysis results in Model-III, if a collision position and direction can be moved from the center of the upper arm a little, it may be said that the humerus can restrain impact stress in a collision. In other words, a pitch which hits the batter is not just taken to decrease impact stress, like in Model-I, but it is necessary to avoid right, left or backward stress like in Model-III. This is one method that reduces the damage of the soft tissue and fracture risk of the bone of the upper arm.

4. Conclusion

In this study, collision experiments and dynamic finite element analysis of a baseball were performed, and the finite element model of a ball was suggested. In addition, with this ball and human body model (THUMS), collision simulations to the living body (an upper arm) were analyzed, and the injury to the upper arm was examined. The following data were conclusively obtained:

- (1) A finite element model showing the dynamic property of a hard-baseball was suggested.
- (2) The bone damage and effect of soft tissue on the upper arm of the human body in a ball collision were examined.
- (3) If the position and direction of the ball's collision can be kept away from the center of the bone even slightly, the risk of humerus injury and damage of the soft tissue is reduced.

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