Finite Element Simulation of the Ulnar Collateral Ligament Under Various Valgus Torques to Investigate Injury to the Ligament in Baseball Pitchers

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Fall 2023 g

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Summary

In Major League Baseball (MLB), injuries to the ulnar collateral ligament (UCL), are an ongoing epidemic [1]. The main injury mechanism has to do with the excessive valgus loading placed on the UCL during the end of the arm cocking phase/beginning of the arm acceleration phase of a pitch [1][2][3]. During this pitching phase, the elbow is subjected to an external valgus torque, which contributes to external rotation at the shoulder, and hence the arm, imparting a massive tensile force on the medial side of the elbow. The UCL plays an essential role in resisting this external valgus torque, but like all soft tissue, has its limits [3]. In 2018, 13% of MLB players and their minor league affiliates underwent UCL reconstruction, for a total of 637 athletes, with pitchers occupying a large majority of the affected population [2]. Additionally, 75 of 700 MLB pitchers (11%) had Ulnar Collateral Ligament Reconstruction (commonly known as "Tommy John" Surgery) [1]. There has been a large motivation in the baseball community to gain a better understanding of the injury mechanics, as they relate to pitching biomechanics, for UCL injuries, since the Tommy John surgery keeps players out of the game for at least 9 months, costing major league teams hundreds of millions of dollars on athletes who cannot participate [1].

In this project, the effect of valgus torques on the UCL during pitching will be investigated. A variety of induced valgus torques will be simulated in LS-Dyna, on two different models of varying fidelity. The valgus torques are correlated with pitching speeds, as seen in the literature, and the results of the simulations will be used see whether an injury would be sustained based on existing injury thresholds [1]. A total of 6 simulations will be conducted, across two models, each with three different loading patterns based on the pitch speed. The post-processing features of LS-Dyna were used to extract the response at the elbow and compared to the experimentally extract in-vitro strain threshold of the UCL (11.96%), to predict whether an injury has occurred [1].

First, the LSTC NCAC 50th percentile Hybrid III (H3) crash dummy was first used to investigate the loading conditions on a simplified model. The results of the simulation were unphysical in nature due to the limitations of the model's applications, indicating a need for a more biofidelic model to be implemented to better encapsulate the loading behaviours. The biofidelic model used was the THUMS full-body model developed by Toyota, specifically, the Version 7 AM50 Occupant model. A direct comparison of the two models was not possible since the H3 model did not include any ligaments. However, an attempt was made which showed that the H3 model was extremely limited, and produced unphysical responses to valgus torques, which is likely due to constraints on the joints' range of motion and the use of simplified hinges and levers to represent joints. Unlike the H3 model, the THUMS model has very anatomically detailed joint and tissue definitions, making it better for this application. The results will be presented later in detail, but it is important to note that both models have many limitations.

Introduction

In Major League Baseball (MLB), injuries to the ulnar collateral ligament (UCL), are an ongoing epidemic [1]. Such injuries may include partial tears or complete ruptures [1]. In 2018, 13% of MLB players and their minor league affiliates underwent UCL reconstruction, for a total of 637 athletes, with pitchers occupying a large majority of the affected population [2]. To put that into context, between 2013-2018, these athletes collectively missed 14,232 regular season days while teams have spent approximately $193.5 million on those pitchers' salaries, who are not even participating in practice or play [1]. For comparison, The National Diabetes Association has reported that in the US, for those over the age of twenty, 11.3% have diabetes [1].

The main injury mechanism is complex and varies between different sources of literature [1]. However, many sources agree that a major mechanism of the injury in pitchers has to do with the excessive valgus loading on the UCL during the end of the arm cocking phase/beginning of the arm acceleration phase of a pitch [1][2][3]. This is exemplified in Figure 1 [1]. During this pitching phase, the elbow is subjected to an external valgus torque, during the external rotation of the shoulder, imparting a massive tensile force on the medial side of the elbow. The UCL plays an essential role in resisting this external valgus torque, but like all soft tissue, has its limits [3].

Figure 1: Visualized Valgus Torque and Tensile Force on the Medial Elbow during the Late Arm Cocking-Early Arm Acceleration Phase of Baseball Pitches [1].

In-vitro studies conducted on cadavers reported that the UCL would rupture after an ultimate external valgus torque of 34 Nm was applied to the proximal end of the ligamentous insertion [3]. Other studies suggested that an exhibited strain of 11.96% would also lead to tears in the UCL [1]. These metrics seem to conflict with the data that suggests that MLB pitchers may experience peak valgus torques in the range of 40 to 120 Nm [3]. The injury thresholds mentioned were extracted from experiments on the isolated ligament and do not consider the stabilizing behaviour of the surrounding structures, such as the joint capsule, other ligaments and even muscles that span the elbow joint [4]. Studies have shown that the musculature in the elbow alone produces a maximal counteracting varus torque of 71 Nm, contributing to the stabilization of the elbow, and reducing the load on the UCL [4].

The complex anatomy and structural contribution of the elbow joint make predicting UCL injuries quite difficult [3][4]. Coupling this with the variability induced by fatigue and pitch counts makes UCL injuries nearly impossible to predict, using modern methods [3][4]. In this project, the effect of valgus torques on the UCL during pitching will be investigated. A variety of induced valgus torques will be simulated in LS-Dyna, which correlate with pitching speeds, as seen in the literature, to see whether an injury would be sustained based on existing injury thresholds [1]. To do so, the moment arm between the elbow joint centre of rotation and the top of the arm, where the ball is held, will be approximated from the model and multiplied by the induced valgus torque. This will yield a compressive force at the top of the model, which will serve to externally rotate the shoulder, which is fixed rigidly in place, effectively straining the UCL. First, the H3 crash dummy was first used to investigate the loading conditions on a simplified model. The results of the simulation were unphysical in nature, indicating a need for a more biofidelic model to be implemented to better encapsulate the loading behaviours. The biofidelic model used was the THUMS full-body model. Initially, the entire bony anatomy of the throwing arm, including the humerus, radius, ulna, and all bones of the hand were to be included in the biofidelic model. The idea was to apply the compressive force at the centre of the hand, to simulate a real throw, as was done in the H3 model. After some trial and error, the bones of the hand and the radius were difficult to implement in the THUMS model, since their contact force definitions were not well applied, causing the model to continuously fail. As such, the model was reduced to only include the absolute necessary anatomy for the problem, which was the humerus, ulna and UCL. The post-processing features of LS-Dyna will then be used to extract the strain at the elbow and a measured strain greater than 11.96%, anywhere on the UCL will indicate that an injury has occurred [1].

Background

UCL injuries can be simplified greatly as a unassuming physics problem. As mentioned earlier, the valgus torque at the elbow applies a tensile load on the UCL, and this torque quite often exceeds the in vitro UCL failure threshold. Thinking of this through the lens of Newton's second law, the greater the mass of the pitchers throwing segment, the greater the tensile force on the insertions of the UCL would be [3]. Furthermore, for faster throws, a greater acceleration of the upper arm would occur, increasing the tensile load on the UCL insertions [3]. When you begin to incorporate the angles of shoulder external rotation, and elbow flexion, then the 3-dimensional force contributions must also be considered. For example, pitchers achieve between 150-180° of external shoulder rotation during the late arm cocking-early arm acceleration phase of the pitch, with between 70-90° of elbow flexion [5]. Typically, it is considered that the ideal pitching motion achieves 180° of external shoulder rotation with 90° of elbow flexion and shoulder abduction [5]. Ideally, it would be best to simulate the differences in UCL strain at different joint angles. Due to issues in the joint positioning functionality of the THUMS model, changes to the joint angles were not possible and the default position of the model was used since it best replicates the ideal conditions out of all the available biofidelic models.

The main anatomy that is relevant to this injury, as it pertains to this project, includes, the bones of the arm that attach to the ligament (i.e. humerus and ulna) and the UCL. In reality, all of the components of the elbow joint and arm/hand, including muscles, ligaments, joint capsule, synovial fluid etc. would need to be considered in a complete evaluation of the injury. Additionally, as mentioned earlier, the bones of the hand and radius were also removed from the model, to avoid some fatal errors in the model caused by inappropriate contact definitions in the hand, and for the sake of simplicity, only the humerus and ulna were included, as they directly attach to the UCL.

In terms of trauma criteria, only in vitro studies have been conducted [1][3]. As with all ligaments, different grades of trauma exist [6]. Grade 1 is excessive ligament laxity due to stretching; grade 2 is a partial ligament tear, and grade 3 is a complete ligament tear, otherwise known as a rupture [6]. As it relates to the injury thresholds, literature suggested that an in vitro valgus torque on the UCL of 34Nm would cause a rupture, indicating that this would be a grade 3 injury. At the same time, a strain of 11.96% or greater would initiate ligament tearing implying this would be a grade 2 injury threshold [1][3].

Regarding the relevant material properties, the crucial element to model properly would be the UCL. For the UCL properties, the THUMS full-body model developed by Toyota was used as a reference, since the H3 model did not contain any ligaments [7]. Specifically, the Version 7 AM50 Occupant model was used [7]. The AM50 model was selected since it contains the anatomy of the 50th percentile adult male population and most closely fits with the build of an average MLB pitcher, out of all the models available [5][7]. In the documentation for this model, ligaments and other soft tissues were supposedly designed as hyperelastic materials, as highlighted in the model documentation [7]. The documentation justified this decision by stating that ligaments generally have low stiffness for small elongations and higher stiffness for greater elongation [7]. However, upon close inspection of the keyword input file, it is evident that a piecewise linear plasticity material model (MAT_024) was selected [7]. The use of MAT_024 is a direct contradiction to the claims that ligaments were modelled as hyperelastic, since MAT_024 is an elastic-plastic model. Despite this contradiction, the piecewise linear plasticity material model that was the default for ligament material definitions in the keyword file was left as is, to not overcomplicate the implementation of the model or cause possible errors. However, it is important to note that this is a key limitation of the biofidelic model. In terms of the key material properties that were implemented into the piecewise linear plasticity model, the values of interest are the density, Young's modulus, Poisson's ratio, and yield stress, which had default values of 1.1e-9 ton/mm^3, 1000 MPa, 0.49, and 9 MPa respectively. A thorough literature review confirmed that the selected values for density, Poisson's ratio and yield stress were very good approximations of the literature values, and as such, were left as is [1][8][9]. However, when it came time to select Young's modulus for the simulation, there were conflicting values in the literature [1][10][11]. In three sources, the linear region of the stress-strain conditions for the UCL were reported, but their methodologies and objectives differed significantly, which is likely what led to the differences in reported results [1][10][11]. Additionally, all three sources relied on cadaveric UCL specimens, and depending on the quality of that specimen (which has material properties that vary based on specimen geometry, age etc.), different outcomes may occur. Hansen reported a modulus of 68.6 MPa, Smith reported a modulus of 13.77 MPa and Huang reported a modulus of 23.2 MPa [1][10][11]. After much consideration, Hansen's value for Young's Modulus (68.6 MPa) was selected [1]. The main reason for this is that Hansen's work most closely aligned with the objectives of this project [1]. Additionally, much of the data used in this project (i.e. valgus torque values and strain threshold) came from Hansen's paper. By consistently using data from the same source, a cohesive set of data could be implemented, instead of picking and choosing parameters from various sources, that each vary in application [1]. Finally, Hansen included the most detailed account of how the modulus was calculated [1]. The final material properties selected for the UCL can be summarized in Table 1.

Table 1: Final UCL Piecewise Linear Plasticity Model Material Properties

UCL Material Property (MID: 85100301)	**Value**
Density	1.1e-9 ton/mm^3 [8]
Young's Modulus	68.6 MPa [1]
Poisson's Ratio	0.49 [9]
Yield Stress	9 MPa [8][9]

Various experimental data exist which are pertinent to this project. First, recall that the ideal pitching motion achieves 180° of external shoulder rotation with 90° of elbow flexion and shoulder abduction [5]. Ideally, the project aims to examine the strain on the UCL at these ideal joint positions. As mentioned earlier, this is not possible and the standard positions of the right arm of the V7 AM50 model were used. Next, determining the moment arm will be essential to converting the valgus torques into an approximated compressive force on the palm which externally rotates the rigid shoulder. However, the most essential piece of experimental data comes from the valgus torques associated with different pitching speeds. Hansen's paper conducted a literature review which correlated pitching speeds with valgus torques [1]. This experimental data will serve as the input for the valgus torque in the project's simulation, which when multiplied by the moment arm, will produce the compressive forces on the palm [1]. A summary of the data selected and implemented from Hansen's review can be seen in Table 2 [1]. Note that these values were not measured directly and were extrapolated from different results in their respective literature. Additionally, these values were extracted from a motion analysis of a pitcher during a throw, including the ball, meaning that the segment masses and mass of the ball are considered in the valgus torque values [1].

Table 2: Pitching Speeds and Associated Valgus Torques from Literature [1].

Source	**MCW Database (2011)**	**Aguilando (2009)**	**Chu (2009)**
Mean Pitching Speed (Mph)	85	75.8	60
Mean Valgus Torque (Nm)	54.8	50	46

Methods

As mentioned previously, the H3 model was first used to gain a better insight into the loading response of a valgus torque in baseball pitches, on a lower fidelity model, compared to the higher fidelity THUMS model. To begin, the model was downloaded, and the keyword file was used to set the joint positions on the graphical interface. The only joints that were altered were the shoulder and the elbow joints. The shoulder was abducted to 90°, the elbow flexed to 90° and the shoulder externally rotated to -40°. These were the values that were deemed best for a pitcher's form in the literature, except for the shoulder external rotation [5]. Ideally, the shoulder should have been externally rotated to -90°, but the maximum permissible range in this model was -40°. This is a major limitation of the model since it would be best to have the shoulder externally rotated to the ideal position taught to the pitcher (90°) [5]. The 'flesh' of the arm was also cleared from the part viewer to get a clear sight of the inner workings of the dummy since this is most representative of the anatomy being used in the higher fidelity THUMS model and the more superficial soft tissues of the arm are not pertinent to the simulations. An image of the model positioning being used can be seen in Figure 2. The contact definitions between various segments in this model were left unchanged.

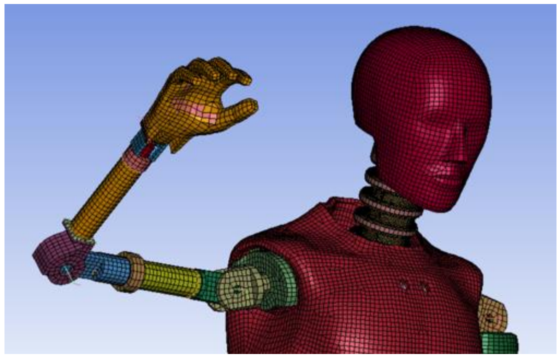

Figure 2: Hybrid III model used for simulation with the throwing arm joints positioned.

With the joints in position, the distance between the centre of rotation of the elbow joint and the centre of the palm was approximated, to represent the moment arm which converts the elbow valgus torque to the applied force. The applied force would be evenly distributed among nodes about the size of a baseball's diameter (about 73 mm), in a circular distribution originating at a central node of the palm [3]. That means that the moment arm was approximated and measured from the central distal node of the elbow hinge opening (purple part in Figure 2) and the most central node of the palm (NID: 52491064). This was done using the ruler tool in the interface, and the distance was approximated to be around 300 mm. Next, the simulation time of the experiment was set to 0.5 seconds, which will be justified shortly, by configuring the CONTROL_TERMINATION keyword, while the simulation timesteps were set to 0.1 seconds,

It was now time to begin defining the boundary conditions for the various parts of the model. To begin, a node-set was defined by selecting all the elements for the humerus-like part, and then the SPC_SET boundary condition for that set was created, which did not permit any motion to the part (linear and rotational). This was done to keep the part fixed in place. Similarly, a node-set was created for the forearm-like part, and then the SPC_SET boundary condition for that set of nodes was created which only permitted rotation about the y-axis and limited any linear or rotational movements in all other directions. This was done to simulate the valgus torques imparted at the elbow by having the forearm rotated around the fixed humerus, like a damped lever system rotation around a fixed beam.

Afterwards, the nodes of the palm, corresponding to a circular distribution of a baseball's diameter, originating at a central node, were set and titled load_nodes. These nodes were selected as the nodes that would be impacted by the compressive force and were used since this is where the ball would be held. Since multiple nodes were selected, the loading values defined later had to be divided by the number of nodes, to ensure that the sum of those force contributions by each node was equal to the total desired force.

It was now time to begin defining the load curves to be imparted on the load_nodes. Three simulations were to be conducted, one for each valgus torque reported in the literature, as seen in Table 2. To determine the force which would be imparted on the palm, the valgus torque for each test case was multiplied by the measured moment arm (~300mm), which produced the desired force. The force was evenly distributed across all the nodes of the baseball's geometry. These forces had to be formatted in the form of a curve, with timesteps and associated loading values. Recall previously that the simulation time was set to last 0.5 seconds with 0.1-second timesteps. Literature states that during the late arm cocking phase, it takes major league players approximately 0.5 seconds to reach maximum shoulder external rotation, which is assumed to correspond to the 180° of shoulder external rotation expected from these players [1][3][4]. The simulation time was set to 0.5 seconds, to theoretically provide the valgus torque with sufficient time to externally rotate the throwing arm to the desired position. Furthermore, the time steps of 0.1 seconds were arbitrarily selected. The curve was defined by increasing the force from 0 to the maximum values calculated, in a linear fashion. This means that from 0 to 0.5 seconds, with a timestep of 0.1 seconds, 6 load data points were created (since the first point was 0 seconds with 0 load), which were linearly spaced until the maximum force (corresponding to the max valgus force) was reached. This linear loading pattern was arbitrarily selected since there is no literature which describes the loading patterns needed. For each pitch speed, a .csv file was created in excel in this manner, each with 6 data points of the time (0-0.5 second) and the load at that time, which was then used as the input for the DEFINE_CURVE keyword in the software. For each case, the input curve was assigned to the LOAD_NODE keyword. The force was applied to the node-set corresponding to the palm.

With all this done, the model was ready to be simulated. 3 simulations were conducted, pertaining to each of the pitching speeds and associated loadings. Once the simulations were complete, some analysis on the stress distributions on the elbow hinge was conducted, since the model did not have any ligaments to investigate [1]. In the results section, these simulations will prove to be inaccurate, since they exhibited very odd behaviours, with the mechanical responses at the elbow being very unphysical. This will be discussed thoroughly in the results sections and some hypotheses as to why this occurred will be considered. Regardless, there was a need for a more biofidelic model to better encapsulate the loading mechanics.

The THUMS Version 7 AM50 Occupant model was used as the biofidelic model for this project [7]. All edits to the model were made in the LS Pre-Post graphical interface. To begin, the full body model was stripped to the desired bones of the throwing arm (humerus and ulna) and the UCL by deleting the unnecessary parts from the keyword file. The justification for this revolved around the idea that the inclusion of the bones of the hand and the ulna was not pertinent to the simulation based on the insertion of the UCL to only the humerus and ulna, and caused some fatal errors when simulations were attempted with these bones included. A copy of the entire arm and final, implemented biofidelic model can be seen in Figure 3. The removal of the hand bones and radius may have made the model more manageable, but ultimately, reduced the accuracy of the biofidelic model which was a key limitation in this project.

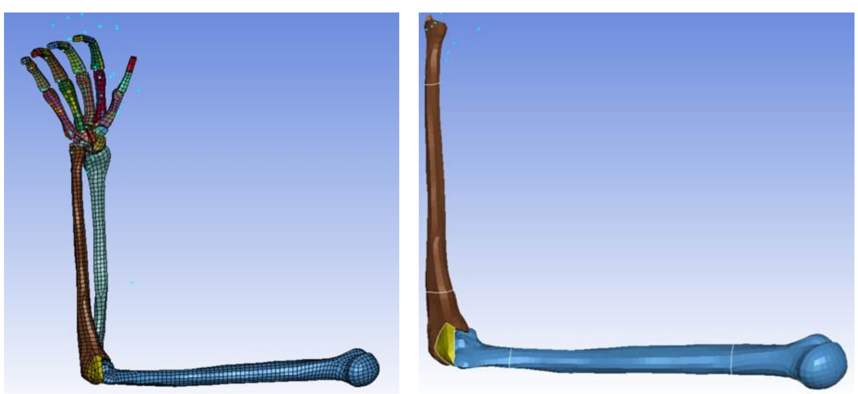

Figure 3: Whole Throwing Arm model (left) and Implemented Biofidelic Model (right) of the Humerus (blue), Ulna (brown) and UCL (yellow), to be used in the Simulation.

With the relevant anatomy now isolated, a new coordinate system was defined, using the *DEFINE_COORDINATE_SYSTEM keyword, which aligned the longitudinal axis of the humerus with the x-axes, with the y-axes pointing up and the z-axis pointing anteriorly. This would make prescribing loads easier later. Once that was done, the distance between the centre of rotation of the elbow joint and the top of the ulna was approximated, to represent the moment arm which converts the elbow valgus torque to the applied force. This was done using the measurement tool in the interface, and the distance was approximated to be around 240 mm. The correct material properties, as seen in Table 1, were then added to the keyword file for the UCL material definition. Next, the simulation time of the experiment was set to 0.5 seconds by configuring the CONTROL_TERMINATION keyword, while the simulation timesteps were set to 0.1 seconds, as was justified in the methods of the H3 model.

It was now time to begin defining the boundary conditions for the various parts of the model. The contact force definitions of the arm, including the interfaces of the humerus, ulna and UCL were sufficient in the original model, and as such, no changes were made there. To begin, a node set was defined by selecting all the elements for the humerus, and then the SPC_SET boundary condition for that set was created, which did not permit any motion to the part (linear and rotational). This was done to keep the humerus fixed in place. Similarly, a node set was created for the ulna, and then the SPC_SET boundary condition for that set of nodes was created which only permitted rotation about the x-axis and limited any linear or rotational movements in all other directions. This was done to simulate the valgus torques imparted at the elbow. The decision to do so essentially made this complex anatomy into a 2D lever, where the humerus stays stationary (x-axis), while the ulna rotates around it (z-axis, with rotation around x). While this happens, the UCL, which attaches to both bones, will be stretched, causing changes to the tissue.

Afterwards, the top-most node of the ulna was put into a set, titled load_node. This node was selected as the node that would be imparted by the compressive force and was used because it is closest to the hand, which, would be the best area to impart the force since this is where the ball would be held. Ultimately, the decision to impart the entire force on one node, specifically on the ulna, was a decision made to further simplify the simulation. As mentioned previously, it would have been better to impart the force on the centre of mass of the hand, but also in a more uniform manner, with the force distributed along the palm. This simplification made the biofidelic more manageable but again is a big limitation to the model's accuracy. It is generally not recommended to apply a load on one node, since this would put an immense amount of pressure on the node, however, due to the small magnitude of the forces being used, which were scaled down by the moment arm, it was deemed acceptable.

It was now time to begin defining the load curves to be imparted on the load_node. Recall that three simulations were to be conducted, one for each valgus torque reported in the literature, as seen in Table 2. To determine the force which would be imparted on the ulna, the valgus torque for each test case was multiplied by the measured moment arm (~240mm), which produced the desired force. These values were 13.152 N, 12 N and 11.04 N for the pitching speed of 85 Mph, 75.8 Mph, and 60 Mph, respectively. These forces had to be formatted in the form of a curve, with timesteps and associated loading values. Recall previously that the simulation time was set to last 0.5 seconds with 0.1 second timesteps, which has the same justification and implementation as the H3 model. For each pitch speed, a .csv file was created in this manner, each with 6 data points of the time (0-0.5 second) and the load at that time, which was then used as the input for the DEFINE_CURVE keyword in the software. For each case, the input curve was assigned to the LOAD_NODE keyword, which was constrained to loading along the negative z-axis. This was done by selecting the z-axis as a single degree of freedom with a negative one scaling factor, to account for the compressive nature of the force, required to rotate the throwing arm externally. The force was applied to the node-set corresponding to the top of the ulna.

With all this done, the model was ready to be simulated. 3 simulations were conducted, pertaining to each of the pitching speeds and associated loadings. Once the simulations were complete, the engineering strain of the UCL was extracted from the d3plot file and reported for comparison to the injury threshold from the literature [1].

An important aspect of the project was validating my biofidelic model, namely, the mechanical response of the UCL under such loading conditions. As such, before the simulations were conducted, it was essential to undergo some model assessment to determine the initial validity of the model. Hansen's work referenced throughout this report conducted a very similar investigation, where he developed an isolated UCL model, based on geometry and material properties from literature [1]. The model was subjected to various loading conditions, like this project, and the various ligaments responses were measured and reported [1]. This paper has been cited extensively and used for a variety of publications relating to UCL injuries, making it an excellent source for comparison to this project. In Hansen's work, the maximum longitudinal stress on the UCL was reported for elbow valgus torques of 40, 60 and 80 Nm on the UCL. Hansen converted these torques into tensile loads applied uniformly to the distal end of the UCL, with the proximal end being fixed. For the sake of validation, a separate iteration of the THUMS model consisting of only the UCL was created, which was loaded in the same way, with the same values as described in Hansen's paper. The maximum longitudinal stress (z-axis in this case) was reported and compared to Hansen's values as seen in Table 3.

Table 3: Comparison of Maximum Longitudinal Stress in Hansen's model Versus the altered THUMS Isolated UCL model for Validation.

Valgus Torque (Nm)	**Max Longitudinal Stress from Hansen (MPa) [1]**	**Max Measured Longitudinal Stress (MPa)**	**Percent Difference**
40	4.44	3.81	15.27 %
60	6.65	5.93	11.45 %
80	8.88	8.47	4.73 %

Evidently from Table 3, the model used in this project produced some mixed results. The differences likely stem from the fact that the geometry used in Hansen's model differs significantly from the geometry of the UCL in the THUMS model. Hansen also used a very simplified material model which was linearly elastic, which may have also led to differences. For the smaller valgus torques, the percent difference between Hansen and the THUMS model was large. However, for the largest simulated torque, the percent difference between the two models was 4.73%. Considering that in this project, the selected valgus torques from the literature are on the higher end compared to the 40 and 60 Nm torque used for validation, it is sensible to assume that the implemented biofidelic model works well enough at higher torques. As such, since this report used generally higher torques than in the validation tests, it is reasonable to accept the implemented biofidelic model as being fairly accurate, when compared to the literature.

The original THUMS model has been well studied and considerable thought and effort was made to ensure its accuracy. It is among the gold standard models for human body finite element simulation. However, the extensive changes and simplification made to the final biofidelic model, as seen in Figure 3, have left some questions about the accuracy of the model. The changes to the selected anatomy, boundary conditions and material properties have not been thoroughly validated. For the sake of the project, these changes will be accepted, but it is important to note that all results and conclusions should be taken with a grain of salt.

Results and Discussion

The results of the H3 model were rather disappointing, but expected. Since there were no ligaments, it was not possible to do analysis on the mechanical response of the UCL, making direct comparison to the THUMS model impossible. Nonetheless, some mechanical analysis was done on all 3 simulations with the H3 model, namely on the elbow hinge joint of the throwing arm. The fringe component functionality in LS Pre-Post was used and the stress along all 3 axes were considered. The order of magnitude of the stresses, mainly in the area of the attachment of the hinge and forearm component were extremely unphysical. This region exhibited stresses nearly an order of magnitude larger than the THUMS simulation for the 85 mph throw and the discrepancy between the other two pitching speeds and their equivalents in the THUMS model also varied greatly. These values far exceeded the yield stresses of the ligament as highlighted in Table 1. Furthermore, upon animating the model, there was some unphysical behaviour at the elbow in which the elbow appeared to lock up and contort after a certain range of motion was reached.

Ultimately, the nature of the H3 model was not ideal for this sort of simulation. Recall that this model was designed for vehicle occupant frontal impact analysis, which differs significantly from its use here. It's inability to respond accurately to the load case prescribed in this project, combined with the anatomically inaccurate definitions of the joints (i.e. elbow is a mechanical hinge), may have made this model unable to handle the valgus loading experienced by baseball pitchers. Additionally, it is hypothesized that the deformation that occurred during animation was a product of the rigid constraints of the range of shoulder external rotation. If you recall, the shoulder was only rotated up to its maximum permissible range, as defined by the model. The force on the palm imparted during simulation would tend to further externally rotate the arm, but this may have been impacted by the rigid constraint on this type of motion. Evidently, the limitations of the lower fidelity H3 model were reflected by its mechanical response under these loading conditions, further justifying the need for the higher fidelity THUMS biofidelic model. So although the results were disappointing, implanting the H3 model was important in justifying the need for a better model.

With regards to the THUMS mode, it was initially thought that the UCL would exhibit plastic strain when being injured. However, a review of the literature on the yield stress implemented into the model confirmed that the yield stress of the UCL is roughly 9 MPa [9]. From the validation activity conducted, and reported in Table 3, the max longitudinal stress did not exceed the yield stress of the UCL under the most extreme loading condition (85 Mph throw). This makes sense since, throughout the duration of play, it is extremely unlikely that the athlete will reach the plastic response on their ligament every time they throw. As such, there was a need to calculate the engineering strain of the ligament under these conditions, instead of extracting the effective plastic strain directly. The engineering strain was selected because the degree of elongation was quite minimal. The strain was calculated by measuring the change in length in divided by the original length of the ligament. This was done with the ruler tool in the software. Note that while using the ruler tool, the longitudinal length (z-axis) was selected as this produced the largest change in length, and hence, the maximum strain, which is useful in assessing the chances of injury to the ligament. The engineering strain can be seen in Table 4 and an example of how it was measured is seen in Figure 4.

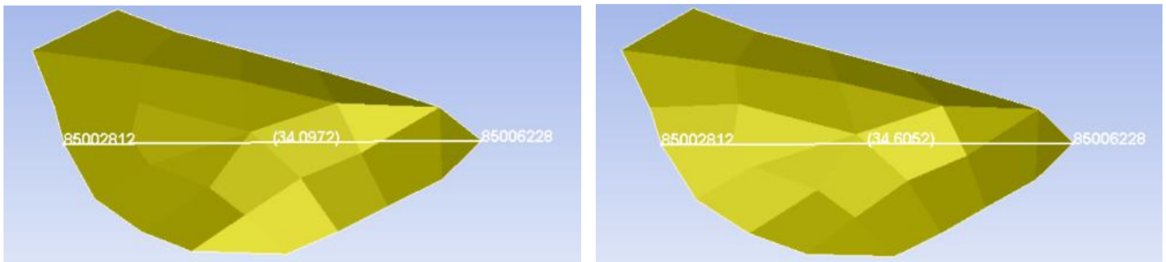

Figure 4: Sample Engineering Strain Measurement, using the Ruler Tule, for the 60 Mph Throw. Both the Initial Length (left) and Final Length (right) are Shown. The other parts of the model are hidden to highlight the changes.

Table 4: UCL True Strain from Simulation in Comparison to the Injury Threshold

Pitching Speed	Initial UCL Length (mm)	Final UCL Length (mm)	Engineering Strain (%)	Exceeds Threshold? (11.96%) [1]
60 Mph	34.0972	34.6052	1.489%	NO
75.8 Mph	34.0972	34.6093	1.502%	NO
85 Mph	34.0972	34.6147	1.518%	NO

For all 3 simulated test cases, it is evident that the UCL did not come close to the injury thresholds and likely did not sustain a ligament sprain or tear. As mentioned previously, the changes made to the original THUMS model, such as the selected anatomy, boundary conditions and material properties have not been thoroughly validated. For the sake of the project, these changes were accepted, but all results are questionable. All changes have been justified and explained in this project, but to say that the model used for simulation is appropriate for this loading case is not acceptable, in its current state. Even after the justifications, the THUMS model was designed for vehicle occupant impact analysis, which differs significantly from its use here [7]. Even though the results seem reasonable, the model was never truly configured for finite element simulation of the UCL under valgus loading, and as such, more work needs to be done to determine whether the model is appropriate for this type of loading scenario. However, logically, these values made sense, since a single loading scenario will likely not result in UCL injury, since soft tissue fatigue is a huge part of the injury mechanism, and will be discussed shortly [1-4].

Although the simulations did not exceed the injury thresholds, some key components of throwing biomechanics are not considered which further puts the results into question. In the literature, it was shown that the musculature in the elbow alone produces a maximal counteracting varus torque of 71 Nm, contributing to the stabilization of the elbow, and reducing the load on the UCL [4]. This metric would likely be greater if the other elements of the shoulder joint were included such as the joint capsule, flesh and skin and other ligaments. This contribution was not included in the model and makes the results questionable.

Furthermore, UCL injury in baseball is difficult to predict because of the variety that exists between the conditions of individual players [3]. For example, some key predictors of UCL injury based on literature include muscular fatigue, pitch count, pitch-type percentage, body weight and body height, which are not considered in this project [3]. Fatigue and pitch count have been studied extensively in baseball biomechanics, but there have yet to be any concrete predictors to injury that can be explained by these factors, due to their complexity [3]. It is assumed that excessive use of the throwing arm, as is the case for a pitcher during a game, would cause muscular fatigue, changing the loading patterns of the UCL, while also increasing the ligament's laxity [3]. These conditions may ultimately impact the material properties of the ligament, making it more susceptible to injury, which again, is not being considered in this project [3].

Recommendations

Throughout the project, many simplifying assumptions were made. However, each time an assumption was made, the model became more unrealistic and moved further away from real-life applications. The main recommendation to improve the accuracy of the model, and hence, its reliability, includes the selection of the THUMS material model, properties, and geometry for the UCL, the positioning of the joints, the included anatomy, moment arm measurements, and the range of motion of the upper extremities. Only improvements to the THUMS model are presented.

The assumption that is believed to have had the greatest impact on the accuracy of the analysis is the selection of the material model, properties, and geometry of the UCL in the biofidelic model. Recall that the original THUMS model labelled their ligaments as hyperelastic and strangely used a piecewise linear plasticity model for all the ligaments [7]. Not only that, but the material properties, such as density and Young's modulus had values that were not reflective of the true properties of the UCL. To address this, literature was consulted, and updates were made to the material properties, as mentioned earlier. However, if the aim was to model ligaments as a hyperelastic material, there are more reliable hyperelastic material models in LS-Dyna [7][12]. TO name a few, *MAT_HYPERELASTIC_RUBBER or *MAT_MOONEYRIVLIN_RUBBER/ *MAT_OGDEN_RUBBER as was done in previous course assignments. However, a crucial flaw of the hyperelastic and piecewise linear plasticity definition for the ligaments in the THUMS model is that ligaments, especially the UCL, are transversely isotropic and viscoelastic, which are behaviours not entirely encapsulated in the implemented model [7][13]. The assumption that the ligament does not exhibit isotropic, viscoelastic behaviour is a major one that likely skews the results. Since ligaments display time-dependent behaviour and have a nonlinear stress-strain response, they are inherently viscoelastic [13]. Furthermore, due to the orientation of the collagen fibers in ligaments, they display directionally dependent material properties, which can qualify them as transversely isotropic [13]. LS-Dyna has a material model that perfectly suits the behaviours of ligaments and is designed for soft tissues, like ligaments [12]. A key recommendation would be to implement this material model instead, which is *MAT_091-092 or *MAT_SOFT_TISSUE_VISCO (the viscoelastic option) [7]. This material is a transversely isotropic viscoelastic model for representing biological soft tissues such as ligaments, tendons, and fascia [7]. The viscoelasticity option relies on a a six-term Prony series kernel for the relaxation function which represents the elastic (long-time) response [7]. However, this model was not used in the project since its reliance relies on the use of a six-term Prony series kernel, where each term requires some analysis to identify the terms' coefficients [7]. The difficulty with this is that there was no reliable UCL material property experimental data to fit the coefficients of this model accurately, and as such, was not used.

Another issue with the UCL in the THUMS model is that its geometry and anatomy are not entirely accurate. Firstly, the UCL is comprised of 3 bands, the anterior, posterior, and transverse bands [2][3]. The THUMS model only considers one band, which appears to be the anterior band, which for the sake of this project, is acceptable, considering that an overwhelming majority of UCL injuries in baseball occur at this anterior band [2-4][7]. Additionally, the geometry of the single band in the THUMS model has not been justified entirely and using some basic preliminary analysis (measurement tool in LS-Dyna), appears to be wildly inaccurate [7]. Measuring the geometry of the UCL is no easy task and often requires detailed MRI data to achieve this successfully, which is beyond the scope of accuracy aimed to be achieved by the THUMS model [7]. To improve the biofidelic model used, it would be recommended that all three bands of the UCL be included, and the geometry of the bands, especially the anterior one, be more rigorously implemented, as seen in Hansen's work [1-3][7].

A key flaw in the implemented model was the inability to change the initial joint positions of the V7 AM50 Occupant model. Typically, when downloading the model from the THUMS site, a separate file is meant to be included which allows for the customization of initial joint angles. After extensive review, these files were unavailable, preventing any customization of the initial joint angles. As such, the joint positions in the biofidelic model were left unchanged. It would have been beneficial to change the initial joint positions of the elbow and shoulder joint to reflect the ideal conditions from the literature, using the built-in joint changing functionality of the model or even prescribing forces to alter the initial position of the joints.

Another crucial recommendation to improve this model would be to expand the selected anatomy that was included in the throwing extremity. Recall that the only included anatomy was the humerus, ulna and UCL [7]. In the literature, it was shown that the musculature in the elbow alone produces a maximal counteracting varus torque of 71 Nm, contributing to the stabilization of the elbow, and reducing the load on the UCL [4]. This metric would likely be greater if the other elements of the shoulder joint were included such as the joint capsule, flesh and skin and other ligaments. As such, it would have been useful to keep all the anatomic elements of the upper extremity to get a more realistic representation of the throwing arm, since their inclusion would have provided a stabilizing effect to the joint. However, the goal of the project was to examine the response of the UCL to valgus torques alone, partially justifying the decision to omit the other anatomical elements. The biofidelic model in its implemented state also greatly simplified the complexity of the model making it more manageable when it came time to make edits and to simulate it.

Next, the moment arm measurements conducted to convert the valgus force of the elbow to a compressive force at the palm were simplified to match the scope of the project. Recall that the ruler tool in LS-Dyna was used to get an approximate distance from the elbow joint centre of rotation and the distal extremity. However, the measurement used was a very rough number that did not accurately reach either end of the true joint centres and instead relied on human perception. Additionally, since the bones of the hands were omitted from the THUMS model, the moment arm was shortened. A better approach would have been to include the hand in order to represent the location of the ball during the throw, which would have increased the moment arm slightly, compared to the use of the top of the ulna. It would have been useful to identify the exact nodes or elements which most accurately represent the true joint centres, but this is no easy task and was beyond the scope of the project.

Finally, the last key recommendation for this project revolves around the true biomechanical ranges of motions of the elbow and shoulder joint during a pitch. During the late arm cocking-early arm acceleration phase of baseball pitches, which is the region of interest for this project, the pitcher should demonstrate a highly dynamic kinematic chain from lead knee extension, pelvis rotation, upper trunk rotation, elbow extension, to shoulder internal rotation [14]. During this kinetic chain, the angles of the joint of the throwing arm, namely the elbow and shoulder, are changing at incredibly high rates, with elbow extension and shoulder internal rotation reaching changes of up to 2700°/s and 7500°/s, respectively [14]. Obviously, during a highly dynamic movement such as a baseball pitch, the angles of the elbow flexion/extension and shoulder abduction/adduction would not be fixed but were constrained via boundary conditions to further simplify the analysis and implementation of the model. A recommendation to improve this model would be to allow more dynamic movement of elbow flexion/extension and shoulder abduction/adduction to represent the dynamic changes to joint angles that are present in the baseball pitch. This can be done by creating a model designed for pitching biomechanics, which is highly dynamic and encapsulated the full dynamic throwing behaviour.

Despite the fact that there are many more limitations and recommendation that can be used to improve this model, the project was still an excellent learning tool, which provided insight and a greater appreciation for the complexities of biomechanical finite element simulations. The results produced seem reasonable, but cannot be taken at face value, since the problem has been oversimplified. Hopefully, this analysis can be used as a foundation for further research into the difficulties associated with predicting UCL injuries in baseball pitchers.

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