Musculoskeletal Research Center

Summer Research Program

Department of Bioengineering

University of Pittsburgh

MSRC

2011

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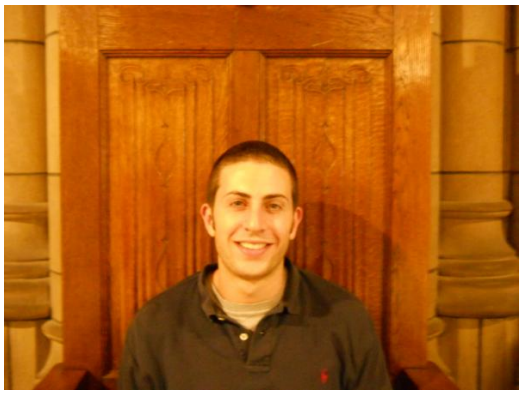
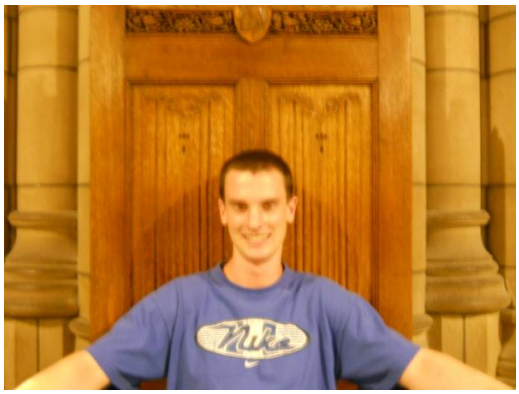
2011 Abstract Book Committee



Brad Edelman



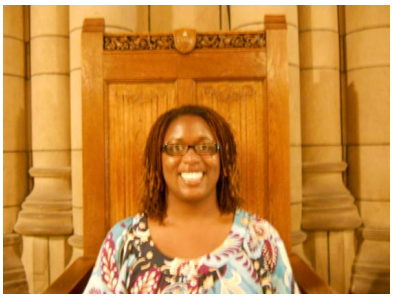
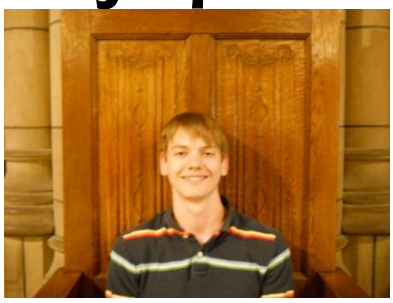
Zachary Merrill

The work presented in this abstract book represents the labors and efforts of the class of 2011 summer interns. I think I can speak for all the summer interns when I say that this summer has been one of the most rewarding experiences of our lives thus far. We have all come to greatly respect the passion and dedication that the members of the MSRC put in everyday to create a successful research institution and we are all very thankful that they have taken the time this summer to pass along some of their skills to us. In addition to what we have learned about our respected studies, the amazing University of Pittsburgh faculty has guided us in gaining valuable experience in abstract writing and presentations as well. The following abstracts are the perfect representation of the hard work and commitment that each of us has put in throughout the course of the summer.

On behalf of my fellow summer interns, I would like to thank the MSRC for providing us with such a friendly and patient environment to learn and be a part of. In addition, we would like to especially thank Dr. Woo, Dr. Debski, and Dr. Abramowitch for allowing us the opportunity to be a part of their work. We will take the knowledge that we have gained here at the MSRC throughout both the rest of our careers and lives.

- Brad Edelman, Editor

2011 Summer Symposium Committee


Jonquil Flowers, Daniel Browe, Aimee Pickering, Joseph Kromka, Hunter Eason

The symposium showcased all of the hard work in testing and research we've performed over the ten to twelve weeks during the MSRC Summer Research Program. Congratulations to all participants for doing a wonderful job in presenting his or her work. Even though the success of the presenters cannot be conveyed on paper, the following abstracts provide a great representation of the work that was put in by each student this summer

Each and every one of us in this Summer Research Program would like to thank the faculty, staff, fellows, and graduate students of the MSRC. The knowledge and skills gained over the weeks, as well as the success of this symposium could not have been possible without your mentorship and feedback.

Finally, we would like to thank Dr. Woo. Without your guidance, and vast knowledge and experience in the field of research science, the MSRC would not be the place it is today.

- Jonquil Flowers, B.S., Symposium Committee Chairman

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I was born on March 4, 1990 in the town of Indiana, Pennsylvania. For the first 18 years of my life, I spent in Indiana except for one six-month stint in New Mexico when I was six years old. I have an older brother, Matt, a younger brother, Jon, and a younger sister, Katie. My older brother works for the U.S. Department of Defense as a chemical engineer. My younger brother will go to college this year at the Indiana University of Pennsylvania, and my younger sister is entering her junior year in high school. When I was in high school, I ran for the cross country and track teams, and I still run today (though not competitively) when I can find the time. I also enjoy pick-up games of basketball, football, and ultimate frisbee.

My time in the Bioengineering Department at the University of Pittsburgh has been both interesting and challenging. I was drawn to Bioengineering because I was interested in how the human body works, and I have always been good at math. The last three years have validated my choice of major, and I hope to attend graduate school in Bioengineering (concentrating in biomechanics) after I graduate from the University of Pittsburgh in the spring of 2012.

This was my second summer at the Musculoskeletal Research Center, and it has been a great experience. Learning about biomechanics research has solidified my choice of the biomechanics concentration within the bioengineering major. Thanks to all the professors, graduate students, fellows, and other summer students in the lab for making this summer so much fun. I'd like to especially thank Dr. Debski for his guidance and Dr. Voycheck for her patience as a teacher throughout my time with the MSRC. Finally, I'd like to give a special thank you to Dr. Woo for being a mentor to us all.

THE MECHANICAL PROPERTIES OF THE GLENOHUMERAL CAPSULE: CHANGES INDUCED BY ANTERIOR DISLOCATION

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INTRODUCTION

The glenohumeral joint is the most frequently dislocated major joint in the body with about 2% of the population dislocating their shoulders between the ages of 18 and 70 [1]. About 80% of these shoulder dislocations occur in the anterior direction, and they most commonly occur in the apprehension position, which is characterized by 60° of glenohumeral abduction and 60° of external rotation [2]. The most common pathology associated with dislocation is instability due to permanent deformation of the glenohumeral capsule [3]. The glenohumeral capsule is a continuous sheet of tissue that connects the humeral head to the glenoid of the scapula and stabilizes the joint in end ranges of motion. When the shoulder is dislocated, the glenohumeral capsule becomes stretched-out and is no longer able to stabilize the glenohumeral joint. This may occur because the mechanical properties of the glenohumeral capsule change as a result of dislocation. Current surgical repair techniques for shoulder dislocations typically involve plicating and suturing the glenohumeral capsule in order to reduce redundancy in the tissue; however, up to 25% of patients still experience pain and instability after surgery [4]. Surgeons may benefit from knowing the location and extent of altered mechanical properties in the glenohumeral capsule. Therefore, the objective of this work was to determine how the mechanical properties of the glenohumeral capsule change in response to anterior glenohumeral dislocation.

MATERIALS AND METHODS

The procedure that was used had three stages, two experimental and one computational. The first experimental stage dislocated human cadaveric glenohumeral joints in the anterior direction. The second experimental stage tested the mechanical properties of the injured glenohumeral capsules from stage one. Finally, the computational protocol determined the material coefficients of the samples of glenohumeral capsule from the load-elongation curves from stage two.

For the first experimental stage, the following procedure was developed based on the work of Moore and coworkers [5]. Six fresh-frozen cadaveric shoulders were dissected down to the glenohumeral capsule. The humerus and the scapula were then fixed in epoxy putty, and the shoulder was then mounted onto a robotic/Universal Force-Moment Sensor testing system such that the humerus was fixed in space, and the scapula was attached to the end of the robotic arm. The shoulder was then taken to the apprehension position, 60º of glenohumeral abduction and 60º of external rotation, and dislocated in the anterior direction by incrementally loading the joint in the anterior direction until the humeral head translated half the largest anterior/posterior width of the glenoid in the anterior direction plus 3 mm. Following dislocation, the glenohumeral capsule was excised from the specimen and frozen for future mechanical testing.

For the second experimental stage and the computational stage, the following procedure is based on the work of Rainis and coworkers [6]. Five 25 mm by 25 mm sections are cut from each of the glenohumeral capsules of the injured shoulders. One section each came from the Anterior Band of the Inferior Glenohumeral Ligament (AB-IGHL), the Posterior Band of the Inferior Glenohumeral Ligament (PB-IGHL), the Axillary Pouch of the Inferior Glenohumeral Ligament (AP-IGHL), the Posterior Region (PR), and the Anterior-superior Region (AS). The tissue is hydrated with saline solution and placed in custom soft tissue clamps. A grid of 3 x 3 strain markers is adhered to the tissue using super glue such that the markers are at least 1 mm apart and at least 1 mm from the edges of the clamps. The tissue was then placed in the materials testing machine (EnduraTEC, Elf 3200, Bose), and a 0.5 N preload was applied. Measurements of the tissue geometry were taken using a ruler for the clamp-to-clamp distance and the width of the tissue and digital calipers for the thickness of the tissue. The tissue was then preconditioned using 10 cycles of 1.5 mm elongation. Following the preconditioning, the 0.5 N preload was reapplied. The non-destructive loading consisted of a 2.25 mm elongation for the tensile loading, and 0.4 of the clamp-to-clamp distance for the shear loading. After testing was completed in the longitudinal direction, the tissue was removed from the clamps, rotated 90 degrees, and the shear and tensile testing was repeated in the transverse direction.

The computational protocol involved using the load-elongation curves and tissue sample geometry as boundary conditions to determine the material coefficients using an inverse finite-element optimization technique [6]. An isotropic hyperelastic strain energy function, based on the form originally used by Veronda and Westman, was assumed to describe the tissue behavior [7]. The strain energy (W) is described as follows:

$$W = C_1[e^{C_2(\tilde{I}_1-3)} - 1] - \frac{C_1C_2}{2}(\tilde{I}_2 - 3) + \frac{1}{2}K[\ln(J)]^2 \qquad (1)$$

where $\hat{I}_1$ and $\hat{I}_2$ are the deviatoric invariants of the right Cauchy-Green deformation tensor **C**, $1/2K[\ln(J)]^2$ governs the dilatational response of the tissue (where J is the volume ratio), and C_1 and C_2 are the material coefficients that were determined using the inverse finite-element optimization routine, where C_1 scales the magnitude of the stress-stretch curve and C_2 governs the magnitude and nonlinearity of the stress-stretch curve [6]. The material coefficients of the average stress-stretch curves were compared using the results of the sensitivity analysis, 0.30 MPa for C_1 and 3.0 for C_2, for clinical significance.

RESULTS

The material coefficients, C_1 and C_2, calculated for the injured axillary pouch and posterior region (n = 1) are shown

in table 1 and compared to the average coefficients for the healthy tissue [8]. The coefficients for the healthy tissue are averaged across many specimens, but the coefficients for the injured tissue are from only one specimen. Data are shown for the axillary pouch (AP) and the posterior region (PR). A sensitivity analysis has determined that differences of 0.30 MPa for C_1 and 3.0 for C_2 are clinically significant [8].

Table 1. Comparison of C_1 and C_2 for healthy tissue and injured glenohumeral capsule.

	Healthy	**Injured**	**Healthy**	**Injured**
	C_1 (MPa)	**C_1 (MPa)**	**C_2**	**C_2**
AP	**0.16**	**0.61**	**9.3**	**2.9**
PR	**0.34**	**0.34**	**8.5**	**3.0**

These results are compared using only the findings of the sensitivity analysis. In the axillary pouch, C_1 is greater for the injured tissue than the healthy tissue, and C_2 is smaller for the injured tissue than the healthy tissue. In the posterior region, there is no difference in C_1 between the healthy and injured tissue, and C_2 is smaller for the injured tissue than the healthy tissue.

DISCUSSION

This study determined the material coefficients C_1 and C_2 for injured glenohumeral capsule in a single specimen and compared these coefficients to healthy glenohumeral capsule. It was found that there were clinically significant differences in the axillary pouch for both C_1 and C_2 between the healthy and injured glenohumeral capsule; although, only one specimen was tested in the injured state. The material coefficients C_1 and C_2 have no direct physical meaning, but some inferences can be made concerning the shape of the corresponding stress-stretch curves. If the results from the single injured glenohumeral capsule are representative of all injured glenohumeral capsule, the higher C_1 value would indicate a larger initial stiffness of the injured tissue, and the lower C_2 value indicates a lower ultimate load of the injured tissue. The lower ultimate load would suggest that the injured tissue has sustained permanent damage as a result of dislocation. The higher initial stiffness of the injured tissue may reflect a mechanism of compensating for injury in the dislocated shoulder. Again, more specimens need to be tested before drawing significant conclusions.

In the future, this work will perform mechanical testing on an additional five injured shoulder specimens. Statistical significance can then be determined when comparing the normal and injured material coefficients of the glenohumeral capsule. In addition, more comparisons will be made between the healthy and injured material coefficients. For instance, the average stress-stretch curves can be compared at specific levels of elongation of the tissue. Finally, the last goal of this project is to create a finite element model of the injured glenohumeral capsule using the material coefficients determined in this work. By comparing the finite element models of the healthy and injured glenohumeral capsules, we hope to gain insight into how to better diagnose and treat shoulder dislocations.

ACKNOWLEDGEMENTS

I would like to thank Dr. Carrie Voycheck for all of her guidance and support throughout my project. I would also like to thank my faculty mentor, Dr. Richard Debski, for helping me earn this unique opportunity and for guiding me along the way. I would also like to acknowledge the Beckman Foundation for funding this research and the Swanson School of Engineering at the University of Pittsburgh.

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I was born in Orlando, Florida on November 9, 1990 to Timothy and Kathy Eason, and come from a large family of five children. I moved around a lot during my childhood because my dad was in the navy. My family finally ended up in Lancaster, Pennsylvania when I was fourteen and that is where I graduated from Manheim Township High School. I ran track throughout high school and participated in numerous clubs such as the school newspaper and National Honors Society. I went on two trips to China and one to Ireland while in high school.

I choose to attend the University of Pittsburgh because I loved the city when I visited and could picture myself spending four years in it. The bioengineering department was a huge draw as well. I was excited to take part in such an interesting field where I could actively participate in research. The school has lived up to the expectations I had and in many ways surpasses them. I did not expect to find such socially outgoing people amongst my engineering classes, but I have made some truly amazing friends in them.

My current plans for the future are to attend medical school. I am very unsure of where I want to go at the moment, but I would say that my two top choices, at the moment, would be to remain here at Pitt or attend Penn State's medical school. I also am unsure of what type of doctor I want to become, but I figure I have some time to figure that out.

I would like to thank Dr. Woo for the opportunity to work for him this summer. It has been an amazing experience and I plan on continuing my research through the school year. Also I would like to thank Matteo Tei for performing the surgeries needed for my testing to take place. Without him my summer project could not have occurred. Finally, I would like to thank my mentor Katie Farraro for working with me this summer. I have learned so much and look forward to working with everyone in the future.

IN VITRO EVALUATION OF THE EFFECTIVENESS OF A MAGNESIUM-BASED RING IN BRIDGING A TRANSECTED GOAT ACL

[1]Hunter Eason, [1]Katie Farraro, Ph.D. student, [1]Savio L-Y. Woo, Ph.D., D.Sc., D.Eng.
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INTRODUCTION

The anterior cruciate ligament (ACL) is injured in sports and work-associated activities with over 100,000 injuries occurring in the United States annually (Beaty). Midsubstance ruptures have a low capacity for healing, so reconstruction with an allograft or autograft has become the gold standard in treatment of ACL injury. However, due to unsatisfactory long-term results of reconstruction, research in promoting healing of the ACL is being conducted. The preservation of the native ACL would be advantageous because it would preserve proprioceptive nerve endings, anatomy of the ACL insertion sites, and double-bundle structure of the ACL (Murray 2005, Schutte, Woo). Current research has shown that scaffolds (Murray 2001, 2006, 2007, 2009, Wiig), cells (Agung), and growth factors (Joshi, Kobayashi, Murray 2009) can be used to increase the healing capacity of the ACL; thus, healing may soon be a viable treatment option for patients.

A previous study at our lab has shown that using an ECM bioscaffold can improve the healing response in the ACL. The experiment used small intestine submucosa (SIS) from genetically engineered pigs to prevent an immunological response and consisted of transection of the ACL and suture repair with addition of the SIS sheet wrapped around the injury site and SIS hydrogel injected into the wound (Liang). The results of the experiment after 12 weeks of healing showed that the ACL had continuous healing tissue without hypertrophy, and the stiffness of the healing FATC was superior to suture repair alone by 2.5 times. The ultimate load was also increased by 90%. However, the SIS-treated FATC had a stiffness that reached only 48% of sham-operated controls, and values for anterior-posterior tibial translation that were two to three times higher, suggesting insufficient initial stability.

These results led to the development of another study to determine the effects of a suture augmentation technique in which the sutures pass directly through a tibial and femoral bone tunnel to provide immediate mechanical support after injury (Fisher 2010, 2011). A time zero study showed a 76% reduction in APTT and in-vivo results after 12 weeks of healing showed a 22% reduction with a stiffness 2 times greater than suture repair alone.

However, despite these promising results, ACL healing is very slow, and a treatment is needed that allows for immediate mechanical loading of the ACL to prevent degradation of the insertion sites during the healing process. Thus, we believe that the addition of a magnesium (Mg)-based scaffold coated in ECM and used in combination with our suture augmentation technique could lead to superior healing. The Mg-based scaffold is cylindrical in shape and bridges the gap between the two ruptured stumps of ligament tissue. The degradation of Mg alloys can be controlled so that the healing tissue bears a load over a time scale of 6-12 weeks (Witte 2005, 2006, 2007). The Mg-based ring will provide initial mechanical support to the joint while the ACL heals but then degrades so the ACL bears the load over time.

OBJECTIVE

The goal of this study is to evaluate the mechanical properties of the injured FATC after suture augmentation supplemented with Mg-based ring repair, and compare the mechanical properties to those of suture augmentation alone at time zero in the goat stifle joint. The structural properties are also compared with those of a bone-patellar-tendon-bone (BPTB) graft.

MATERIALS AND METHODS

The physical design of the Mg-based ring was based on the geometry and available working length of the goat ACL (Table 1). The structural and mechanical properties of the Mg-based ring were designed so that they met or exceed the properties of the normal ACL (Table 2).

Plastic prototypes were created in accordance with the Mg-based ring design criteria and were used to determine the feasibility of suturing a ring structure to the ACL. Dr. Matteo Tei then performed surgeries on cadaveric goat stifle joints to evaluate the feasibility of attaching the ring structure to the ACL and gave his recommendation on design modifications. This led to the development of the first generation of Mg-based rings.

The surgical technique used for the Mg-based ring addition was to initially make a medial incision and remove the fat pad to expose the ACL. Next the ACL was transected and the ring was sutured to the ACL using PDS II #0 sutures. Suture augmentation was then performed by drilling bone tunnels by each insertion site of the ACL. A TI-CRON #2 suture was then passed through the bone tunnels and fixed using titanium buttons.

Next, an in-vitro evaluation of the biomechanical function of the first generation Mg-based ring was conducted. Cadaveric goat stifle joints were used in order to determine the structural properties of the femur-ACL-Mg-based ring-tibia complex (FAMTC).

The structural properties of the femur-ACL-tibia complex with suture augmentation supplemented by Mg-based ring addition was compared to suture augmentation alone using randomly-assigned paired hind limbs of skeletally immature goats. After surgical treatments, the specimens were clamped on a uniaxial materials testing machine and cyclically conditioned between 20 N and 70 N for 50 cycles at a displacement rate of 50 mm/min. The specimens were then allowed to rest for 60 min. A load-to-failure test was then performed at a displacement rate of 10 mm/min and the load-elongation curve was recorded to determine stiffness and ultimate load. Stiffness was determined by calculating the slope of the linear region of the load-elongation curve and ultimate load was the highest value obtained on the load-elongation curve.

RESULTS

The load-elongation curves obtained from the tensile testing can be seen in **Fig. 1** and the structural properties of the different repair techniques are listed in **Table 1**. Suture augmentation alone (n = 2) had a stiffness of 14.8±0.4 N/mm and an ultimate load of 149±3.5 N. Mg-based ring and suture augmentation (n = 2) had higher values for both stiffness and ultimate load, but no statistical significance could be achieved due to the small sample size. The stiffness of the Mg-based ring plus suture augmentation was 22.1±4.7 N/mm and the ultimate load was 211±37.5 N. The structural properties of the BPTB reconstruction were based off a previous study at our laboratory using Ti interference screws. The BPTB had a stiffness of 30.8±11.3 N/mm and an ultimate load of 210.6±52.4 N.

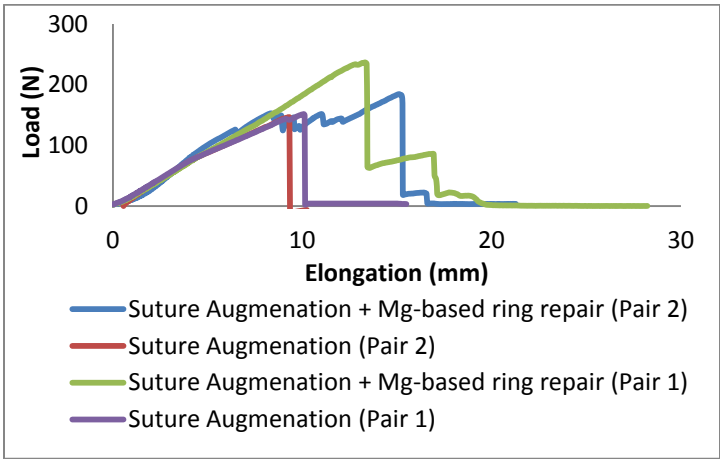

Figure 1. Load-Elongation curves of all four specimen tested .

TABLE 1. Mean values (± standard deviation) of stiffness and ultimate load for suture augmentation with Mg-based ring repair, suture augmentation alone, and BPTB graft.

	Stiffness (N/mm)	**Ultimate Load (N)**
Suture Aug + Mg-Based ring	22.1±4.7	211±37.5
Suture Aug	14.8±0.4	149±3.5
BPTB	30.8±11.3	210.6±52.4

DISCUSSION

The addition of the Mg-based ring repair to suture augmentation appears to increase both the stiffness and ultimate load of the femur-ACL-tibia complex (FATC). This shows that the Mg-based ring has allowed the load to be carried across the ACL during the testing, with the augmentation sutures and ring acting as parallel springs. In addition, the testing showed that the ultimate load of the FATC with suture augmentation and an Mg-based ring is comparable to that of the BPTB graft which is a current gold standard for restoring knee function after ACL injury.

This study is not yet complete because only two pairs of specimens have been tested so far. Further *in vitro* testing must be performed so that the data can be analyzed with independent t-tests to determine if the observed differences in structural properties are statistically significant. In addition to performing more uniaxial tensile tests, robotic testing will also be performed to determine the effects of the Mg-based ring on joint kinematics and in-situ forces in the ACL. Once the ring has been shown to be effective at time zero both in improving the structural and functional properties of the FATC, in-vivo studies will then be conducted to evaluate healing at several time points. For these studies, we will combine the Mg-based ring with suture augmentation and the SIS bioscaffold used in previous healing studies to determine how the combination of mechanical and biological augmentation of the stifle joint can lead to histomorphological and biomechanical properties closer to those of a normal ACL.

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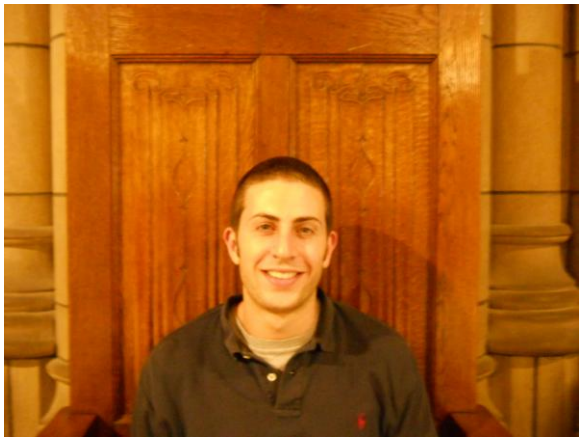

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I was born on September 3, 1990 in Denver, Colorado where I spent the first 18 years of my life. I graduated from Cherry Creek High School in May of 2008 where I developed a great interest in math and the sciences leading me to pursue further education in engineering. My great interest in cars growing up drove me to take an hour and half automotive technology class everyday at a local community college during my senior year of high school. It was this class that made up my mind about pursuing mechanical engineering. In addition to this, I was also active in many activities including NHS, Key Club, basketball and varsity soccer.

As I continued my college career, my interests expanded greatly and I began to integrate many sciences into my educational experience. I am currently obtaining my BS in mechanical and biomedical engineering with a minor in chemistry in hopes of obtaining an MS in biomedical engineering after graduation. At Carnegie Mellon University I play for the varsity soccer team (2009 conference champions!), and have also participated in the Society of Automotive Engineers (SAE), American Society of Mechanical Engineers (ASME) as well as various other clubs.

I was a member of the research team here at the MSRC during the summer and fall of 2010 under Dr. Abramowitch; the experience that I gained greatly improved my understanding of engineering and research. I was very excited to come back and continue my research this summer and gain even more understanding of soft tissue mechanics.

During my work at the MSRC this past summer I have learned so much about the process, problem solving and overall mindset that goes into successful research. The friendly and patient environment of the MSRC has allowed me to apply my classroom knowledge to real life situations and also learn much more. I would like to thank Dr. Abramowitch, Andrew Feola and Dr. woo for giving me this opportunity and guiding me through my project. I have been even further motivated by the knowledge that I have learned here at the MSRC to continue pursuing my passion of engineering.

THE REMOVAL OF PRESSURE FROM THE PELVIC FLOOR IN NULLIPAROUS RATS

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INTRODUCTION

Pelvic organ prolapse (POP) is defined as the downward decent of the pelvic organs into the vaginal canal due to inadequate pelvic support [1]. It is a prevalent condition, yet little is known of its origin due to the multifactorial nature of this disorder. In healthy women, the pelvic organs are bolstered by a structural support system composed of the levator ani muscles, connective tissues, and the vagina [2]. Vaginal childbirth is thought to be the primary risk factor for POP, however, there are many other factors which are believed to contribute to the weakening of the pelvic floor including age, menopause, obesity and chronically increased intra-abdominal pressure [3,4].

Studies in orthopedics have examined the application of Wolff's Law on the supportive tissue structures within the knee providing insight into reactions of soft tissue to varying loads. A study by Woo et al found that when the rabbit MCL was immobilized, its structural properties were inferior to those of naturally loaded tissues [5]. Furthermore, if these immobilized tissues were remobilized, the structural properties were once again found to be inferior to control groups. In addition, literature in urogynecology suggest that increases in IAP raise the incident rate of prolapse which leads us to believe that the mechanosensitivity of the pelvic floor plays a role in the mechanism of prolapse [6, 7]. Therefore, our first aim focuses on the removal of loading on the pelvic floor in order to examine the effects of decreased intra-abdominal pressure (IAP) on the vagina-supportive tissue complex (VSTC). We hypothesized that decreasing the IAP would results in inferior biomechanical properties of the VSTC.

METHODS AND DESIGN

In order to remove the loading normally placed on the pelvic floor, 5 nulliparous Long--Evans rats were hung by their tails in a custom-built hindlimb unloading cage. Following a previously developed model by NASA, a 0.3 m x 0.3 m cage was built with two sets of bearings which supported the hanging tail of the rat. This bearing system permitted movement in both the X and Y axes in order to reach food and water supply. The rats were held by their tails via Gorilla Tape in these cages so that the angle between the torso and the floor was 30 degrees placing normal weight on the front limbs (Fig. 1) [8].

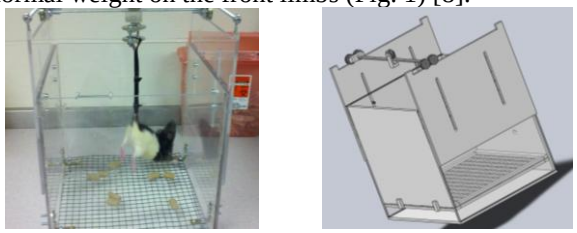

Figure1. (Left) A Long-Evans rat suspended in a hindlimb unloading cage per the parameters listed above. (Right) A SolidWorks model of the final hindlimb unloading cage design.

Rats had full mobility around their cage with food provided and water ad libitum. The tails were checked for injury and slippage, and tape was reapplied if necessary. Rats were suspended for a period of four weeks until testing was performed.This study utilized a previously established testing protocol which performed a uniaxial tensile test of the dissected vaginal-supportive tissue complex (VSTC) (Fig. 2) [1]. In order to obtain the VSTC, the skin, pelvic fat and rear limbs were removed. The spine was then cut just below the chest cavity; the remaining spine was cleaned and potted in PMMA. The potted spine is finally secured in a cylindrical clamp while the vagina is placed in a custom-designed soft tissue clamp and attached to the crosshead of a 1kn load cell with an accuracy of 0.5%.

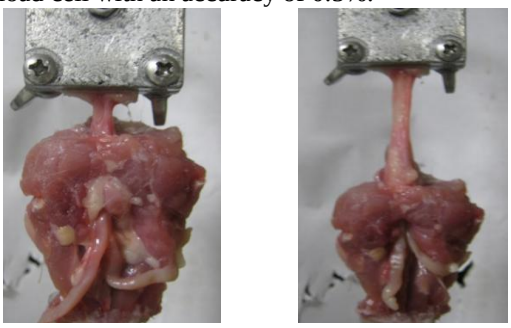

Figure 2. (Left) An example of a dissected VSTC attached to a uniaxial testing machine prior to being loaded to failure. (Right) The same VSCT after the load to failure testing.

Briefly, samples were preloaded (0.15N), preconditioned (0-2 mm, 10 cycles) and loaded to failure at a rate of 25 mm/min. Structural properties including the ultimate load (N), ultimate elongation (mm), stiffness (N/mm) and energy absorbed (J) were calculated from the load-elongation curves acquired from the uniaxial tensile test to failure. The results from the decreased IAP rats were then compared to the structural properties of the VSTC of control rodents (n=8) from a previous study.Statistical analysis was performed utilizing a Student's t-test ($p<0.05$) [1]. All data is represented as mean±standard deviation.

RESULTS

The ultimate load of the control animals was 14.04±1.44 N whereas the value for the decreased IAP animals was 10.27±1.54 N showing a 27% decrease ($p<0.001$) (Fig. 3). The stiffness decreased 52% in hindlimb unloaded animals compared to control animals with values of 1.75±0.38 N/mm and 3.30±0.63 N/mm respectively ($p<0.001$) (Fig. 4). The ultimate elongation of the control animals was 8.50±1.90 mm whereas the value of the decreased IAP animals was 10.3±0.92 mm displaying a 25% increase in the decreased IAP group ($p=0.04$) (Fig. 5). When looking at the energy absorbed for the two groups, the control group had an average value of 52.5±16.37 J,

whereas the average value for the hindlimb unloaded group was 47.52±10.14 J (p=0.55).

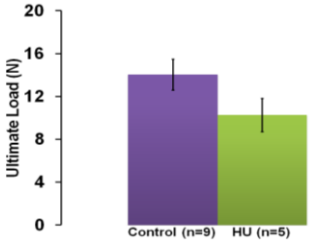

Figure 3. A graph comparing the ultimate load (N) between the control and hindlimb unloaded animals (P<0.001)

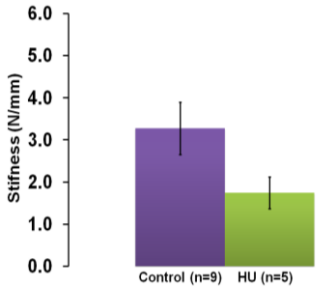

Figure 4. A graph comparing the stiffness (N/mm) between the control and hindlimb unloaded animals (P<0.001)

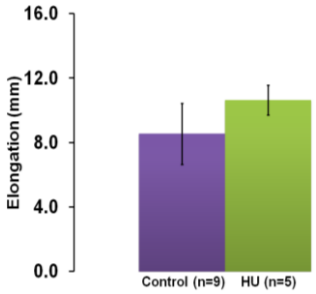

Figure 5. A graph comparing the ultimate elongation (mm) between the control and hindlimb unloaded animals (P=0.04)

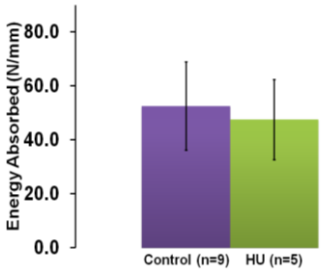

Figure 6. A graph comparing the energy absorbed (J) between the control and hindlimb unloaded animals (P=0.55)

DISCUSSION

The removal of pressure from the pelvic floor relieves the VSTC of the native physiological loading, and these results show that decreasing the loading on the pelvic floor leads to inferior structural properties of the VSTC. Due to this relationship between physiological loading and structural properties, we can begin to quantify the mechanosensitivity of the VSTC and understand how varying degrees of force due to IAP levels can affect the supportive tissue. In this sense, the behavior of the pelvic floor can be characterized, which can lead to theoretical models involving realistic conditions.

In this study there is one major limitation that cannot be overlooked. In the rat, the levator ani muscles primarily function to wag the tail whereas in humans these muscles provide major support for the pelvic organs [2]. Hanging the rats secures their tails in a fixed, vertical position limiting mobility and use of the levator ani muscles. Therefore, a possible reason for the inferior structural properties of the VSTC could be due to a minor case of levator ani and connective tissue atrophy that occurred during the four week hindlimb unloading experiment. A possible solution to this would be to create a type of harness similar to one used by Morey et al that can secure the rat at a 30 degree angle but still allows mobility of the tail [9].

Future studies will aim at exerting excess loads on the VSTC by implanting an abdominal balloon into the peritoneal cavity to determine the effects of an increase of IAP. Additionally, a control group for this study will be tested to ensure that the experimental techniques and results are a valid comparison to the decreased IAP and future increased IAP animals. Additionally, other risk factors including parity, maternal birth injury and simulated menopause can be included in repeat studies in order to isolate and quantify each event and its effects on pelvic support.

ACKNOWLEDGEMENTS

I would like to thank Caroline Evans and Katrina Knight for their assistance with this study. Additionally, I would like to thank Dr. Woo, Dr. Debski and the rest of the MSRC for all of their guidance throughout the summer.

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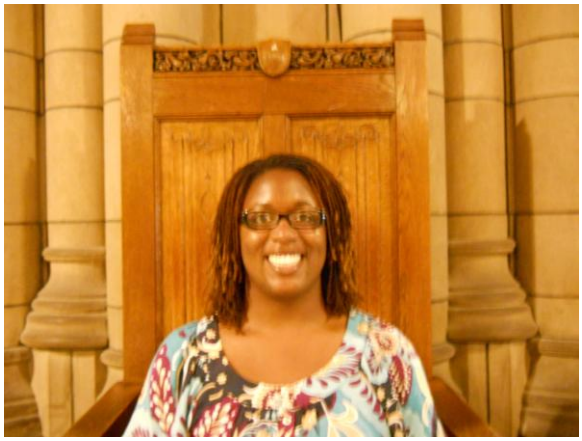

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I am from Atlanta, Georgia where I attended Westlake High School and The Georgia Institute of Technology (Georgia Tech). I have always had an affinity for and excelled in my mathematics and science courses. In high school, I had broad academic interests. For example, I learned French and was a chief officer for *X-Entertainment, Inc.*, the business formed by my entrepreneurship class which served the school and surrounding community. Outside of the classroom, I participated in numerous extracurricular activities. The most memorable was being a member of the City of Atlanta Dolphins swim team, with whom I maintained Georgia "All Star" status.

After graduating salutatorian of my high school class, I attended Georgia Tech. I received my Bachelor of Science in Biomedical Engineering, a major which allowed me to apply mathematics and science to create and improve medical devices and processes. A highlight of my undergraduate academic career was my senior capstone project entitled *A Novel Device for the Reconstitution and Administration of Lyophilized Drug Products*. It was a team-oriented design project involving medical device development processes including prior art research, determination of engineering design specifications, consideration of regulatory compliance, 510(k) composition, and prototype development and testing. In order to supplement my academics, I had opportunities to do undergraduate research at Georgia Tech in the Laboratory for Neuroengineering and the Muscle Physiology Lab. I also participated in several student organizations such as the Women, Science, and Technology Learning Community and was involved with numerous community service initiatives including tutoring at the Bellwood Boys and Girls Club of Metro Atlanta. It was and continues to be my goal to excel academically and positively impact the community.

Upon graduation from Georgia Tech, my desire was to pursue a graduate degree in Biomedical Engineering. Starting this fall (2011), I will begin the masters program at North Carolina Agriculture and Technical State University (NCAT) in Greensboro, NC. This summer research experience at the University of Pittsburgh's MSRC has been a perfect fit for me and my interest in biomedical engineering research, product development, and biomechanics. It has also served as a great introduction to the subject area that will be the focus of my studies at NCAT. Not only have I been exposed to the subject matter, I have learned to ask questions, pay attention to details, and be proactive in considering research ethics. I would like to thank Dr. Woo, Dr. McCullough, and Kwang Kim for sharing their knowledge and research expertise. I am so appreciative to Dr. Woo and my MSRC family for their support.

UNDERSTANDING STRIPPING OF THE MAGNESIUM BASED INTERFERENCE SCREW FOR ACL RECONSTRUCTION USING FINITE ELEMENT ANALYSIS

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INTRODUCTION

The anterior cruciate ligament (ACL) is the most often injured knee ligament. There are over 100,000 ACL injures in the US annually. An injured ACL has a limited potential for healing and leads to functional instability and long term complications such as meniscal injuries, failure of secondary stabilizers, and the early onset of osteoarthritis. Surgical reconstruction is a common treatment and can succeed in restoring the stability of the knee joint. The ACL reconstruction surgery is an arthroscopic surgery which removes the injured ACL and replaces it with an autograft such as the bone patellar tendon bone graft or an allograft. Interference screws are used to ensure secure graft fixation. The two commonly used types of interference screws are the titanium interference screws and the bioabsorbable polymer interference screws. While they both have their strengths and weaknesses, the ideal interference screw can be described as "best of both worlds" displaying a high mechanical strength, secure initial fixation, degradability, biocompatibility, allows for osseointegration and bone healing, and does not interfere with imaging. Magnesium has been shown to be a promising biomaterial for orthopedic applications because of its potential to meet all of the criteria for the ideal interference screw. As a result, a magnesium based interference screw for ACL reconstruction was developed using similar design parameters to that of titanium screws (See Figure 1). Since magnesium differs from titanium, namely magnesium is a softer metal, the magnesium interference screw didn't function as well as titanium screws of the same design. During the biomechanical testing, several issues were observed. There was stripping in the head of the screw at the interface of the screw driver and the screw. With stripping, the screw driver goes around and around without advancing the screw. Stripping prevents full insertion of the screw into the bone tunnels. The in vitro test results show unsatisfactory ultimate load and graft slippage. There was also graft laceration during screw insertion. Based on ACL interference screw studies, it has been determined that screw design can directly affect the screw resistance to failure and its ability to provide secure fixation.

OBJECTIVE

The objective of this preliminary study was to use finite element analysis to understand the stripping mechanism. ANSYS Workbench software was used to simulate the torque and axial load that act upon the drive design of the interference screw during insertion in order to observe the stress distribution therein. The results will be used to propose and justify potential design modifications to the interference screw which will ultimately lead to improved insertion success.

Figure 1: Design Parameters of the First Generation Mg-based Interference Screw
Mg Alloy AZ31, 15mm length, 5mm diameter, 1.73 mm inner diameter (cannulated), 2.51 mm hexagonal drive

MATERIALS AND METHODS

The finite element analysis process required the use of the computer model of the geometry, which was development in SolidWorks. The 3D geometry was imported into ANSYS Workbench. Within ANSYS Workbench, the model was auto-meshed and four simulations were designed and run in order to understand the stripping mechanism that occurred within the hexagonal drive design of the magnesium interference screw.

In all four simulations a fixed geometry was applied to the outer surface of the screw to restrain the motion in the shank portion of the screw and to observe the effects of the loads within the drive design of the screw head. Simulation 1 was run on the screw geometry with titanium material properties with a 2.5 Nm torque applied within the screw's drive design. Similarly, Simulation 2 was run on the screw geometry with magnesium alloy material properties also with a 2.5Nm torque applied within the screw's drive design. Simulations 3 and 4 were run on models of the magnesium screw in contact with a stainless steel screw driver. Simulation 3 applied a 2.5 Nm torque to the screw driver to be applied to the screw. And Simulation 4 applied a 2.5Nm torque and a 10 N axial load. Simulation 4 is the closest to what actually occurs in the clinical setting. The surgeon applies both a torque and axial load in order to screw into the bone tunnel during the reconstruction surgery. The desired output variables included deformation, shear stress and von Mises' stress.

RESULTS

In simulations 1 and 2, they experienced similar shear stress distibutions. The shear stress range was -62.731 MPa to 54.337 MPa. The maximum shear stress was seen at the bottom of the hex drive. The von Mises' stress distribution was also similar. The von Mises' stress ranged from zero to 158.35 MPa. The maximum von Mises stress was seen in the corners of the hexagonal drive of the screw. The deformation patterns in simulations 1 and 2 were similar. However the

numerical values showed the magnesium screw with more deformation than the titanium screw.

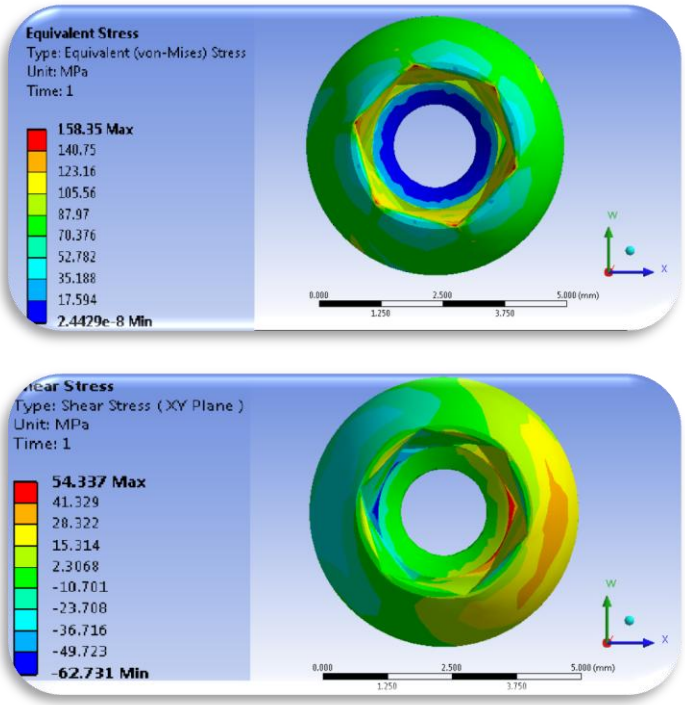

Figure 2: Simulation 1 Results
Von Mises Stress and Shear Stress Distribution

In simulation 3 the maximum von Mises' stress was in the corners of the drive design. The maximum von Mises stress was 31.672 MPa. The maximum shear stress was also seen in the corners of the hex drive. The maximum shear stress was 17.98 MPa. In simulation 4 the maximum von Mises' stress was in the corners of the drive design. The maximum von Mises stress was 1878.9 MPa. The maximum shear stress was also seen in the corners of the hex drive. The maximum shear stress was 1086.4 MPa. The magnitude of the stress in the maximum stress areas must be reduced in order to prevent stripping.

DISCUSSION

There was a common pattern of stress distributon throughout all four simulations. The shear stress was mostly observed at the bottom and in the corners of the hexagonal drive design of the screw. The von Mises' stress was also maximal in the corners of the hexagonal drive design. These areas of high stress must be modified in order to reduce the stress concentrations and ultimately prevent the stripping on the magnesium based screw during insertion.

Based on the results obtained from the finite element analysis and the knowledge gained from previous studies, modifications to the screw design which have the potential to reduce stripping in the interference screw, are needed. Potential modifications to the screw head include increasing the drive depth or changing the drive design all together. There are several drive designs used for ACL interference screws. There are drive designs that have been determined to be superior to the hexagonal design of the magnesium screw. A potential modification to the screw threading is a more gradual tapering which may increase the ease of insertion. Lastly, since the screw driver used for insertion of the polymer screws was also used for the insertion of the magnesium insertion screws, a screw driver specifically designed to be used with the magnesium interference screw could be a possible solution as well.

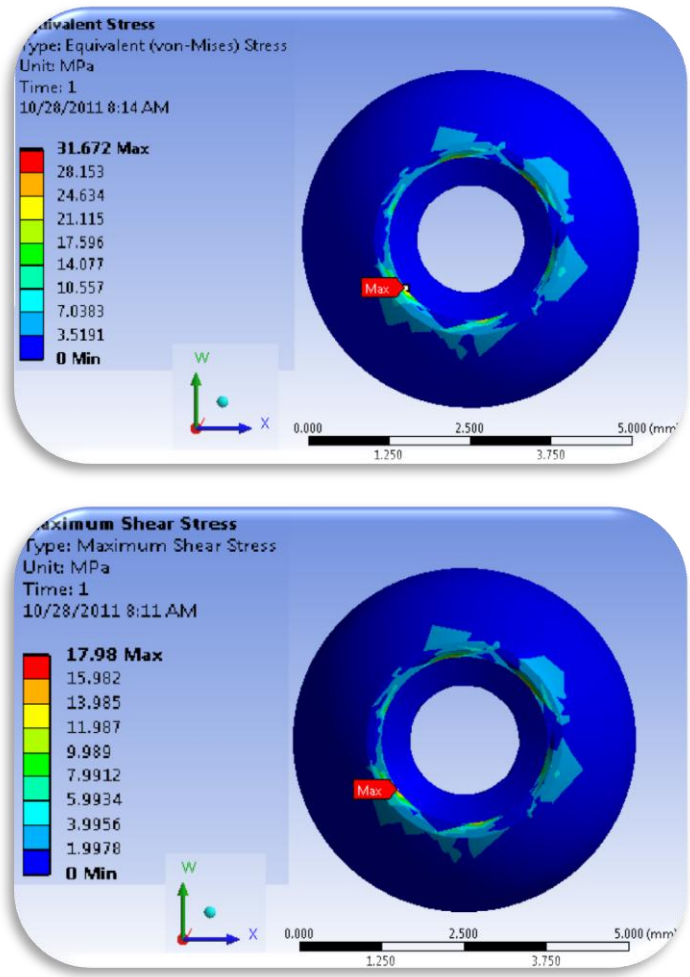

Figure 3: Simulation 3 Results
Von Mises Stress and Shear Stress Distribution

While working in SolidWorks and ANSYS Workbench, it is important to understand how changing certain software paramenters affects the accuracy of the simulations. It was noticed that SolidWorks assembly mates affect the range of motion allowed in the simulations. Also, understanding and accurately assigning the contact definitions is important. There is a large difference in stress values in Simulations 3 and 4. While they weren't expected to be similar because there were different loads applied to each one, the difference appeared to be large and may be due to how the contact region was defined.

The design of the magnesium based interference screw is based off of the design of titanium interference screws. It has been noticed that the polymer screw design possesses some of the proposed design modifications, including a deeper drive design and more gradual tapering. A part of future work is to consider the design of the polymer screw with the magnesium based alloy and asses the performance of that screw. More specific future work includes the experimental validation of the FE model. Secondly, use the validated model to analyze the screw design in reference to the specific problems of stripping, initial fixation strength and graft laceration. Then, use SolidWorks to implement design changes. Lastly, perform FEA on the modified screw design to confirm the superior screw design. Once this process is complete, the second generation magnesium based interference screw can be manufactured.

ACKNOWLEDGEMENTS

Thank you Dr. Woo, Dr. McCullough, and Kwang Kim for sharing you knowledge and research expertise. I am so appreciative to Dr. Woo and my MSRC family for their support.

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I was born on September 9, 1989 in Idaho Falls, Idaho. My family is originally from western Pennsylvania, so we moved back to the Pittsburgh Area when I was 2 years old. I grew up in Monroeville with my 5 brothers and 2 sisters. Consequently, I attended Gateway High School and graduated in 2008. Throughout my time at Gateway, I participated in National Honor Society, earned a letter in cross country, and captained the varsity basketball team.

When it came time to select a college, I chose Carnegie Mellon University for its reputable engineering program, proximity to home, and the opportunity to continue playing basketball at the collegiate level. In fact, this year I will be a captain for the Tartans. When I'm not studying or on the basketball court, I participate in activities for Tau Beta Pi, spend time with family, and enjoy the great outdoors.

During the summer of 2010, I worked in industry doing mechanical and manufacturing engineering work for the Westinghouse Electric Company. However, I felt called to utilize my biomedical training in research and development. Before my time at the Musculoskeletal Research Center, my lab experience was limited to lab classes and I had done no research work. Additionally, my knowledge of anterior cruciate ligament (ACL) reconstruction and repair were limited and I knew nothing about the medial patellofemoral ligament (MPFL). In my internship at MSRC, though, I gained dissection and mechanical testing experience and learned much about the anatomy, injury, and treatment of the ACL and MPFL. I truly have learned a great deal in only a few months time.

Looking back on the summer, I would like to thank my mentor Kwang Kim for introducing me to many facets of tissue engineering and giving me an interesting project to work on. Also, I would like to thank Matteo Tei for his careful instruction on dissection and for giving me his perspective on clinical medicine. I also want to express my appreciation to the Pittsburgh Tissue Engineering Initiative for funding my summer work and my faculty advisor Dr. Savio Woo for giving me the opportunity to perform research in his lab. My experience at MSRC has helped me to decipher what ultimate career path is best for me.

DESIGN OF A CUSTOM PATELLA CLAMP FOR TENSILE TESTING OF THE MEDIAL PATELLOFEMORAL LIGAMENT

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INTRODUCTION

The medial patellofemoral ligament (MPFL) is a fan-shaped ligament that spans from the adductor tubercle of the femur to the superior medial border of the patella and is the primary passive restraint to lateral movement in the range of 0° to 30° of knee flexion [1,2]. It has been found that up to 90% of cases of first time patellar dislocations are accompanied by MPFL disruption [3]. For this reason, surgical reconstruction of the MPFL has become the treatment of choice for recurrent patellar dislocations. However, a significant number of patients experience unwanted side effects such as quadriceps dysfunction, positive apprehension, and decreased range of motion [4]. Furthermore, redislocation rates after surgical reconstruction are still as high as 10-30% [5].

In order to improve surgical outcomes, it is important to first verify that the replacement graft has similar properties to the native MPFL to reproduce normal ligament function. A pilot study at the Musculoskeletal Research Center (MSRC) on the structural properties of the MPFL in a porcine model showed that maintaining the patella's anatomical orientation in the patellofemoral groove during testing is essential in measuring the stiffness of the MPFL. Also, exploratory testing suggested that insufficient preservation of the femoral condyle around the attachment of the MPFL could result in decreased ultimate load. However, the clamp previously used to hold the patella could not fulfill both of these requirements due to its bulky patellar attachment mechanism. Moreover, the clamp lacked fine tune adjustability for patellar tilt as well as for MPFL fiber alignment which is paramount in ensuring uniform fiber loading. Therefore, a new patella clamp was designed to be used in a human study on the MPFL.

EXPLORATORY OBSERVATIONS

Two exploratory tests on porcine MPFL specimens were performed, and no data was officially collected. However, two key observations were made that paralleled the results from a previous MSRC study on the structural properties of the porcine MPFL. The first specimen was tested in a non-anatomical orientation using the old patella clamp and a small femoral bone block potted in a rectangular clamp. When this specimen underwent tensile testing, failure actually occurred by the cortical bone peeling away from the cancellous bone at the site of cutting the femoral bone block. Figure 1 shows a similar failure mode observed in the previous MSRC study. This suggested that a larger piece of femoral bone needed to be used in order to achieve failures at the patella or femoral attachment or at the mid-substance of the ligament.

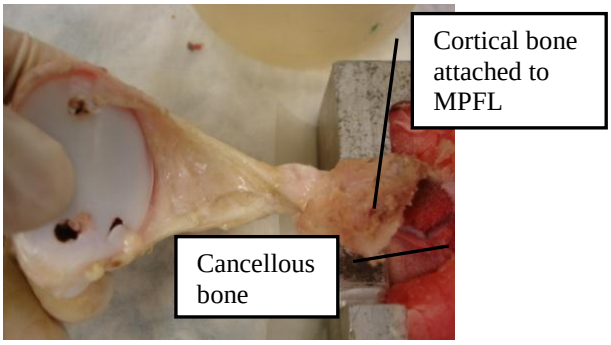

Figure 1: Tensile test of an MPFL specimen with failure due to cortical bone peeling (Previous MSRC study)

The second exploratory test used a specimen oriented in non-anatomical orientation, but used a full femoral clamp that preserved the entire femoral condyle. In this test, the ultimate load of the specimen was significantly higher than the ultimate loads observed in the previous study that used only small femoral bone blocks. With only two specimens tested, no definite conclusions could be drawn. Nonetheless, the observations from exploratory testing did suggest that a new patellar clamp capable of preserving more femoral bone would be beneficial for future MPFL studies.

OBJECTIVE

Therefore, the goal of this summer project was to design an improved patellar clamp for MPFL tensile testing in a human study. This clamp needed to be able to test in "anatomical" and "non-anatomical" orientation—Figure 2 depicts these two patellar positions. Additionally, the patella attachment mechanism needed to be reworked in order to allow preservation of a larger femoral bone block or even an intact femur. Moreover, the new clamp design needed to include a means to adjust the patella tilt and MPFL fiber orientation. These adjustments are essential, especially if a specimen is not potted or drilled in perfect alignment. Finally, the deflection of the clamp during testing needed to be insignificant when compared to the elongation of the MPFL. All of these requirements were taken into consideration during the design process.

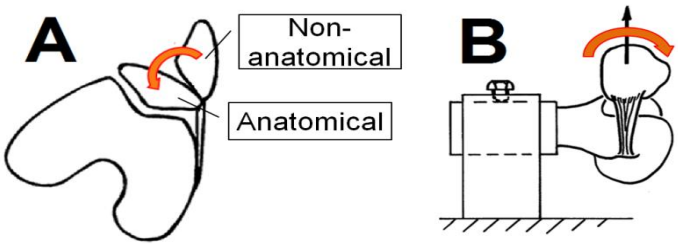

Figure 2: Patella tilt (A); MPFL fiber orientation (B) [6]

DESIGN

A design was formulated using conceptual sketches, simple models, and finally SolidWorks analysis. In this design, two concentric circular plates oriented at right angles. These two plates were fixed using lip-in-groove and a pin-in – slot designs to give infinite adjustability—the old clamp used fixation holes with a finite number of positions. The larger top plate provided adjustment of patella tilt. In order to keep the clamp small enough to fit in a saline bath, two interchangeable top plates were designed—one for anatomical orientation and one for non-anatomical orientation (Figure 3A, #1). The smaller bottom plate offered adjustment for MPFL fiber alignment to guarantee that uniform fiber loading is possible (Figure 3A, #2). Additionally, the clamp was given a more streamlined attachment to the patella than the old clamp. Stainless steel was used instead of aluminum, resulting in a thinner piece (Figure 3A, #3). This made it possible to preserve more bone around the femoral origin of the MPFL, ensuring the integrity of the attachment site. Since the clamp will be used on an Instron material testing machine, it was designed to keep the MPFL aligned such that the fixtures only experienced axial loading. A slotted attachment to the larger circular piece allowed fine tuning of this alignment (Figure 3A, #4). Furthermore, the MPFL's location in the clamp at the center of rotation of both circular pieces allowed the setup to naturally keep the MPFL in the desired alignment, regardless of rotation. Finally, the clamp was designed such that its deflection during testing was insignificant in comparison to the elongation of the MPFL. The clamp will be machined out of 6061 aluminum, other than the stainless steel plate mentioned above, and use stainless steel hardware to avoid corrosion when the clamp is used in a saline bath.

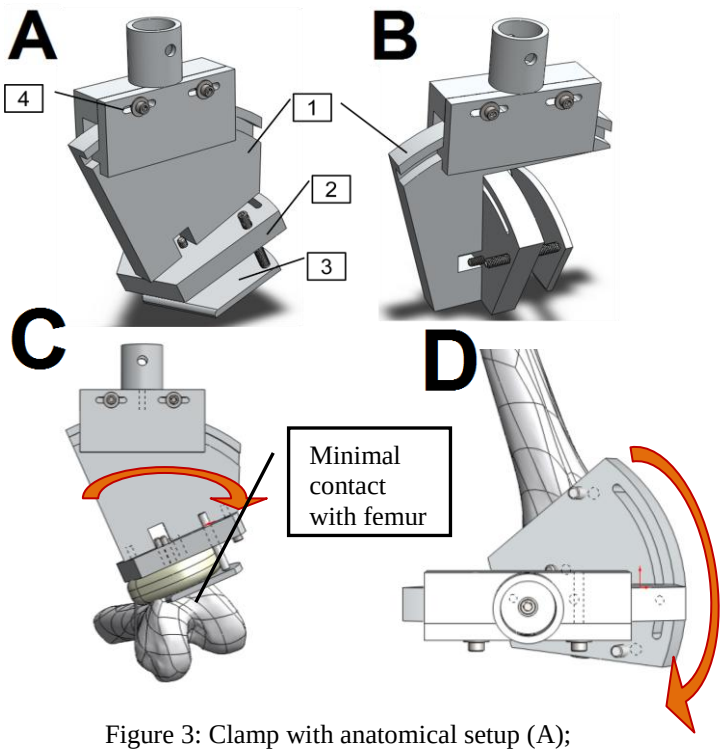

Figure 3: Clamp with anatomical setup (A); Non-anatomical setup (B); Front view of testing setup, patella tilt adjustment (C); Top view of testing setup, MPFL fiber alignment adjustment (D)

DISCUSSION

Exploratory tests were performed in stifle joints from skeletally immature pigs. Therefore, the integrity of the ligament attachment sites may have been relatively weaker than the mature specimens that will be used in the human study [7]. For this reason, it is possible that a small bone block would be sufficient in skeletally mature specimens. However, using a larger femoral bone block will only lend credibility to future studies. However, the clamp that was used in the exploratory test was a large femoral clamp normally used for ACL testing. This clamp is too big to fit in a saline bath, which was part of the testing protocol for past MPFL testing. Therefore, a new femoral clamp may need to be designed to accommodate a larger bone block or intact femur. Overall, this new clamp design addressed the inadequacies of the previous clamp and will allow for more accurate testing in future human MPFL studies.

ACKNOWLEDGEMENTS

Thank you to Dr. Woo to giving me the opportunity to work in his lab and to the Pittsburgh Tissue Engineering Initiative and NSF for funding me. I would also like to thank Kwang Kim and Matteo Tei for their guidance in performing research and dissection. Finally, I would like to express gratitude to everyone in the MSRC as my family for their dependable support throughout the summer.

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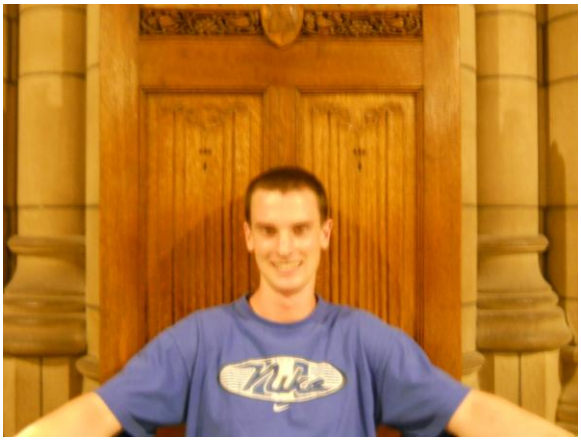

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Shoulder Group
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I was born on August 12, 1990 in Bremerton, Washington, where I remained for about two months before moving to Monterey, California. With a father in the Navy, I then lived in Massachusetts, Monterey (again), and Washington DC, before settling in Burke, Virginia. I have lived in Virginia since I was 10 years old. I attended Lake Braddock High School, where I played baseball and participated in NHS and the Young Conservatives Club. I also worked part time during my junior and senior years as a rehabilitation assistant at Patriot Sportsmedicine, a physical therapy and sports medicine clinic. It was working there that I really became interested in orthopedics and biomechanics.

About half way through my junior year, I made my first visit to Pitt and decided right away that it was one of the top three schools I wanted to attend. After hearing about the engineering program and research opportunities after a couple more visits, I decided that this is where I wanted to go to college. I am very happy with my decision to attend Pitt, as I have had the most fun three years of my life here.

While at Pitt, I have been involved in a handful of organizations, including College Libertarians and NSCS, however I have probably spent most of my free time with my fraternity, Sigma Alpha Mu, where I served as president during my junior year. I am looking forward to another year of having fun while succeeding in the classroom.

Before I joined the Shoulder Group at the beginning of my junior year, I did not have any experience working in a lab. I was very interested in studying biomechanics at the joint and computational level, and I have had a great time working here while obtaining a lot of valuable lab experience. I would like to thank Dr. Debski and Dr. Woo for the opportunity to work at the MSRC for the past year, and Dr. Abramowitch for greatly increasing my interest in biomechanics during Biomechanics I. I would also like to thank Carrie Voycheck and all of my shoulder group co-workers for working with me both in the lab and in the classroom.

IN VIVO ANALYSIS OF OSTEOCHONDRAL DEFECTS AND SHOULDER INSTABILITY

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INTRODUCTION

The glenohumeral joint is the most commonly dislocated joint in the human body, with occurrences as high as 4% in athletic and military populations, and 2% in the rest of the population [1]. The glenohumeral joint also has the widest range of motion in the body, due to the relative sizes of the humeral head and the glenoid fossa, which is located on the superior later portion of the scapula. The humeral head is much larger than the glenoid, in which it rests, making the joint analogous to a golf ball on a tee. While this provides a wide range of motion, the joint is also inherently unstable. Some stability is provided by the capsule, which is essentially a continuous sheet of ligaments, as well as the four rotator cuff muscles; the supraspinatus, infraspinatus, subscapularis, and teres minor. Although these stabilizers help keep the joint intact, the humeral head is often dislocated in the anterior direction. After dislocation, the humeral head is pulled back toward the rim of the glenoid, which can result in lesions being formed on the humeral head, the edge of the glenoid, or both. If these lesions are large enough or located on the right part of the humeral head, they can cause the shoulder to lose stability due to how the lesions move relative to the joint during normal shoulder motion [2].

After dislocation, the injured shoulders are often surgically repaired, especially if a bony (osteochondral) defect results on one of the bones. This surgical repair usually involves using a graft over the defect, but even after surgical repair, the repair can fail and dislocation can still occur. Other methods of failure as defined for this work involve the more objective feelings of instability, pain, numbness, or a reduced activity level.

The effects of shoulder dislocation, subsequent surgical repair, and failed surgery can all be studied by measuring glenohumeral kinematics. Glenohumeral kinematics involves quantifying the relative motions of the humerus and scapula, including rotation angles and translations. The relative rotations were expressed as a sequence of three rotation angles, known as Euler angles, and the translations were expressed in the anterior-posterior, superior-inferior, and medial-lateral directions.

In order to determine the effects of dislocation on shoulder kinematics, the rotations and translations of the humeral head relative to the scapula had to be compared in patients with one previously dislocated shoulder, and one healthy shoulder. The design criteria for calculating the rotations and translations first involved developing reproducible coordinate systems on the humerus and scapula from selected anatomical landmarks. The objective of this work was to design a method to accurately create reproducible coordinate systems on the humerus and scapula of each sample patient from three dimensional computed tomography (3D CT) models, and then use a Matlab script to analyze the relative motions of injured and uninjured shoulders, and finally compare the changes in kinematics in each patient.

METHODS

This study was designed to compare the changes in shoulder kinematics across various groups of patients with dislocated shoulders. The categories of patients observed were patients with no surgical repair, successful surgical repair, and failed surgical repair. Among each of these three categories, the patients were further divided into sub categories, based on if they had an osteochondral defect, and what type of surgical repair was performed (i.e. grafts to the humeral head, glenoid, or both, or a Latarjet procedure).

The first step of this process was to establish coordinate systems on each humerus and scapula from selected landmarks. These landmarks were chosen from 3D CT models collected of both shoulders in each patient in three positions; 0° abduction/0° external rotation (equivalent to the arm resting at the side), 30° abduction/30^oexternal rotation, and with the forearm over the head. These positions were all measured with the upper arm relative to the torso.

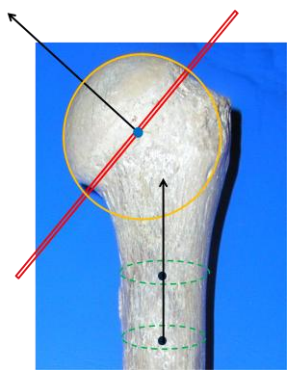

Figure 1: Anatomical landmarks chosen for the humerus. The Y axis was established from the two centroids of the humeral shaft.

The 3D models of the humerus and scapula were created from the CT images using the modeling software Mimics (*Materialise Medical Software*, Leuven, Belgium). The first models created were of the uninjured shoulder in the 0/0 position, and once the anatomic landmarks were selected, these original models were shape matched to the uninjured models in the other two positions. The purpose of shape matching was to keep the landmarks on each bone in the same position relative to each other, and to avoid having to pick the landmarks for each position. Because bony defects on the injured shoulder could alter the selection of the landmarks, the 0/0 uninjured models were first mirrored (with the landmark points), then shape matched to the injured side in the same manner as shape matching the uninjured models.

Using the anatomical landmarks on each of the 3D CT models and the locations of these landmarks in the CT global coordinate system, the humeral coordinate system was established as shown in Figure 1. The points used were two centroids on the humeral shaft, the center of a sphere fit to the humeral head, and a vector normal to the plane formed by the anatomical neck. The Y axis was established in the superior-

inferior direction from the two centroids, then using the normal vector, the X axis was established in the anterior posterior direction, and the Z axis in the medial-lateral direction. The origin was set as the point in the center of the humeral head.

For the scapula, the coordinate system was constructed using the most superior, inferior, anterior, and posterior points on the glenoid, and well as the centroid of the glenoid, which was used as the origin. The X axis was created from the most anterior and posterior points, then the Y axis was set in the superior-inferior direction, and the Z axis in the medial-lateral direction.

The next step was to flip the coordinate systems of the left shoulders so that they would be oriented in the same directions are the right shoulders for easier comparison, as well as maintaining a right handed coordinate system for each shoulder. Flipping the left shoulders was done by multiplying the medial-lateral (Z) axes of the humerus and scapula by -1.

In order to measure the rotations of the humerus, a sequence of three rotations were used, rotating about the Y axis of the humerus, then the humeral X axis, and again about the humeral Y axis. This rotation sequence corresponds to defining the plane of elevation, angle of abduction, and external rotation [3], with both Y axis rotations occurring about the long axis of the humerus. The final step of comparison was done by comparing the elevation angles and translations between the injured and uninjured shoulders of each patient. Establishing the coordinate systems, as well as calculating and comparing the rotation angles were done using a Matlab script.

To determine if creating the coordinate systems used was repeatable, a repeatability analysis was performed using two observers, with four trials each of analyzing the same shoulder. The results showed the intra- and inter-observer standard deviations were less than or equal to 2° for elevation and 0.7 mm for translations, indicating that the coordinate systems can be accurately established for the purpose of this experiment.

RESULTS

For the category of patients with an osteochondral defect and failed repair (N=6), 4 patients had a 4-5 mm anterior shift in the injured shoulder. One patient had a significantly smaller elevation (20 degree difference), while another patient had a 16 mm anterior shift, with the CT model indicating that the humeral head is not resting in the glenoid. A statistically significant difference ($p<0.05$) was observed for elevation angle in the overhead position, with a 7.5% decrease in the injured shoulder.

Of the patients with failed repair and no defect (N=4) and no surgical repair (N=8), two from each category showed an anterior translation of 4-5 mm, while none of the patients showed any major differences in elevation angles.

In the final category of patients with successful repair (N=6), two each have undergone soft tissue capsule repair, glenoid and humeral allograft, and Laterjet repair. 5 of the 6 patients did not show any major changes in elevation or translations, while one had a 15 degree smaller translation and 5 mm anterior translation in the overhead position. The results for this patient are similar to those found for the failed surgical repair patients.

DISCUSSION

Of the 24 patients examined so far, only six patients have had a successful surgical repair. Five of these six did not show any major difference between elevation angles or translations in the injured and repaired shoulders. Having no kinematic differences between the injured and repaired shoulders indicates that the surgeries, which in two cases involved a graft on the humerus and glenoid, two Latarjet procedures, and two soft tissue repairs, truly were successful and that these patients have regained a normal level of mobility in the repaired shoulder.

The other 18 patients tested so far have had failed repair or no repair, and most have not shown any difference in elevation angles, however eight of the eighteen showed a 3-4 mm or greater anterior shift in the injured shoulder in the $30^o/30^o$ position, overhead position, or both. A shift of this size is significant due to the glenoid having a width of only about 2-2.5 cm. The humerus shifting this far forward in the glenoid can greatly reduce the portion of the humeral head that remains within the glenoid, and subsequently reduce the stability of the shoulder in certain positions, leading to a greater chance of dislocation.

CONCLUSION

The design criteria were met by establishing a system to create reproducible coordinate systems on the humerus and scapula in order to calculate glenohumeral kinematics; however, there were some limitations as far as looking at changes in kinematics resulting from dislocation, mostly due to the small sample size of the patients observed. Because there have only been five patients tested so far, there has not been enough data collected yet to draw any conclusions based on the changes in glenohumeral kinematics in the different categories of surgical repair. The data collected so far indicate that in the patient with a successful surgical repair, the repair was truly successful, as the patient has the same kinematics in both shoulders. As for the patients tested with failed surgical repair, none of them showed differences in rotations, but three of the four have shown a trend of increased anterior translations in the injured shoulder. The future plans for this project involve analyzing more patients in each of the categories, and eventually determine from the data how large lesions can be detrimental to glenohumeral kinematics.

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I was born July 15, 1991 and have lived my entire life in Irwin, Pennsylvania, located 45 minutes southeast of Pittsburgh, with my parents and two brothers. I graduated from Hempfield Area High School, where I participated in various clubs and activities such as National Honor Society, Spanish Honor Society, and Interact Club. In addition, I spent a great deal of time volunteering at Excela Health, a hospital system in my area, and I continue to do so whenever time allows. While my extracurricular activities gave me an insight into my future, the stimulating classes I took throughout high school, especially in math and science, truly sparked my interest in engineering and medicine.

I chose the University of Pittsburgh because of its strong bioengineering program and proximity to an advanced medical community. Also, I love living just a bus ride away from the diversity that Pittsburgh offers. In the small amount of free time that the bioengineering curriculum affords, I actively participate in the Society of Women Engineers. As the Outreach Chair, I work with organizations such as Girl Scouts of America to promote women in engineering. I also enjoy working in the community and spending time with friends. Last summer, I received my EMT certification, which proved to be one of the most challenging, yet gratifying, experiences I have had since starting college.

I began volunteering at the MSRC last semester and am thankful to have had the opportunity to stay this summer. Previously, I had no research experience, but my time here has taught me so much. Not only have I learned about ACL reconstruction, the pros/cons of various interference screws, and biomechanical testing, but I have also gained a better understanding of the research process in general. I would like to thank Dr. Woo for welcoming me into his lab, sharing his talents, and providing words of wisdom that extend far beyond ACL research. Also, I want to thank Kwang Kim for his guidance and for teaching me to become a more independent thinker. Lastly, I express my gratitude to Dr. Matteo Tei for performing the reconstructions and helping me with the biomechanical testing.

IN VITRO BIOMECHANICAL EVALUATION OF SECOND GENERATION MAGNESIUM SCREWS

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INTRODUCTION

The anterior cruciate ligament (ACL) is one of the most commonly injured ligaments in the knee. Over 100,000 cases occur annually in the United States alone and most are due to work and sports-related activities [1]. ACL injuries cause knee instability which may lead to damage in surrounding soft tissue, especially the menisci [2]. Furthermore, midsubstance tears have a low capacity for healing [3]. In order to restore knee joint stability, reconstructions using the patellar tendon or hamstring tendon autograft are performed on the majority of patients [4]. Interference screws are used to firmly affix the graft within the bone tunnel.

Currently, surgeons use interference screws comprised primarily of metals (stainless steel and titanium alloys) or polymers (polyesters). Metallic screws are widely used because they provide secure graft fixation and have a low occurrence of breakage and acceptable biocompatibility [5, 6]. However, because metallic screws are permanent, they complicate revision surgeries, and their removal requires additional surgeries [7]. In addition, metallic implants often distort images on MRI scans, which impede a physician's ability to administer follow-up care [8].

In an attempt to overcome the complications associated with metallic screws, bioabsorbable polymer screws have been implemented. Unlike their metallic counterparts, the screws degrade in the body and have no effect on MRI images [9]. However, their degradation rate is not well-controlled, with some screws showing limited degradation after 3 years of insertion [10]. Additionally, polymer screws exhibit an increased occurrence of breakage, resulting in risk to patients [11]. They also have poor biocompatibility which can adversely affect osseo-integration and result in bone tunnel widening [9].

As a result of problems with current interference screws, the advantages of bioabsorbable magnesium (Mg)-based screws are being explored. Through the use of alloying and coating, Mg screws can be designed to degrade at a controlled rate. Upon degradation, the screws are replaced by native tissue, eliminating the need for additional surgeries to remove the implants [5]. Additionally, in comparison to polymers, Mg alloys have superior mechanical properties, such as greater tensile strength, Young's modulus, fracture toughness, and ductility, resulting in a reduction of surgical complications associated with breakage [5, 12]. Mg is also already present in the body so the screw's degradation products can be safely absorbed or excreted [13].

In order to assess Mg-based screws, our research center has performed in-vitro and in-vivo studies on prototypes composed of pure Mg and alloy AZ31. However, problems such as stripping of the screw and unsatisfactory ultimate load and graft slippage were observed. Therefore, 2nd generation Mg-based screws must be developed with improved design, material, and mechanical properties.

OBJECTIVE

The objective of this study is to evaluate 2nd generation Mg screws in terms of initial fixation strength and resistance to graft slippage and compare those results to commercially-available polymer screws (Milagro®). The first phase of this study involved performing an ACL reconstruction on a goat cadaver stifle joint using a bone-patellar tendon-bone autograft and Milagro screws and evaluating initial fixation and graft slippage of the femur-ACL graft-tibia complex (FATC) by performing a tensile test. The Milagro screw data was then compared to data obtained during a preliminary study in which titanium, pure Mg, and Mg alloy AZ31 screws were utilized.

MATERIALS AND METHODS

A 5-mm-wide bone-patellar tendon-bone (BPTB) graft was harvested from a cadaveric goat stifle joint. Then the ACL was surgically removed, and two tunnels (6 mm in diameter) were drilled at the tibial and femoral footprints. Bone blocks (5 mm in diameter) were pulled through the tunnels and affixed with Milagro® (Johnson & Johnson) interference screws. Maximum insertion torque (N-m) was measured using a custom torque driver.

All the soft tissue was dissected from the joint, leaving the FATC, and a uniaxial materials testing machine was used to determine its structural properties. The FATC was secured to clamps and anatomically aligned so that a 45° flexion angle existed between the femur and tibia. A 3N preload was applied and then three cyclic creep tests were performed. For the first and third tests, the grafts experienced loads between 20 and 70 N. In the second test, loads between 20 and 105 N were used. Each test consisted of 100 cycles performed at a rate of 50 mm/min, followed by a 60-minute resting period before the next cyclic test. Residual elongation (mm), the permanent elongation of the FATC as a result of cyclic loading, was determined after each resting period in order to evaluate the graft's slippage. The gauge length after the initial 3N preload was set to 0 mm of elongation. Then after each test and 60-minute recovery, the 3N preload was reapplied, and the gauge length (residual elongation) of the FATC was measured. After cyclic loading, a load-to-failure test was performed at a 5 mm/min elongation rate. From the load elongation curve, the stiffness (N/mm), ultimate load (N), and ultimate elongation (mm) of the FATC were determined. The stiffness was determined by calculating the slope of the linear region on the curve from 50 N to 100 N. Additionally, the mode of failure of each FATC was noted. Independent t-tests will be used to compare the test parameters from the Milagro screw group and 2nd generation screw group ($p<0.05$).

RESULTS

Thus far, successful BPTB reconstructions using Milagro screws and the entire tensile testing protocol have been performed on four (4) cadaveric goat stifle joints. Biomechanical data for these specimens (*n=4*) was collected and can be seen in **Table 1**. Upon completion of three cyclic creep tests, the total residual elongation of the FATC was 1.9 ± 1.1 mm. During the load-to-failure test, the stiffness of the FATC was 49.4 ± 4.3 N/mm. Additionally, the ultimate load was 235 ± 71 N and the ultimate elongation was 9.7 ± 1.6 mm. Two of the specimens failed as a result of tibial screw breakage. In the remaining two specimens, the mode of failure was graft slippage past the screw. This occurred on the tibial side in one case and the femoral side in the other.

Table 2 compares the total residual elongation, stiffness, and ultimate load of the FATCs reconstructed with Milagro screws to data collected during previous studies in which titanium *(n=4)*, pure Mg *(n=4)*, and Mg alloy AZ31 *(n=4)* screws were utilized. The total residual elongation of the FATCs reconstructed with titanium, pure Mg, and AZ31 screws were 1.4 ± 0.4, 6.7 ± 5.0, and 4.0 ± 1.1 mm, respectively. During the load-to-failure test, the stiffness of the titanium, pure Mg, and AZ31 groups was 30.8 ± 11.3, 30.6 ± 18.6, and 44.6 ± 7.3 N/mm, respectively, and the ultimate load was 211 ± 52, 173 ± 108, and 203 ± 18 N, respectively.

Table 1 Summary of data on 4 specimens reconstructed with Milagro screws. Total residual elongation, stiffness, ultimate load, ultimate elongation and failure modes are listed.

	Residual Elongation (mm)	**Stiffness (N/mm)**	**Ultimate Load (N)**	**Ultimate Elongation (mm)**	**Failure Mode**
1	2.3	53.5	330	10.9	Tibial screw breakage
2	1.0	49.6	159	10.6	Tibial pull out
3	3.3	43.5	224	9.9	Femoral pull out
4	0.9	51.2	227	7.4	Tibial screw breakage

Table 2 Total residual elongation, stiffness, and ultimate load in Milagro, titanium, pure Mg, and alloy AZ31 groups.

	Milagro (n=4)	**Titanium (n=4)**	**Pure Mg (n=4)**	**AZ31 alloy (n=4)**
Total Residual Elongation (mm)	1.9 ± 1.1	1.4 ± 0.4	6.7 ± 5.0	4.0 ± 1.1
Stiffness (N/mm)	49.5 ± 4.3	30.8 ± 11.3	30.6 ± 18.6	44.6 ± 7.3
Ultimate Load (N)	235 ± 71	211 ± 52	173 ± 108	203 ± 18

DISCUSSION

The purpose of this study is to compare the initial fixation strength and resistance to graft slippage of 2nd generation Mg-based interference screws to that of Milagro interference screws. Each screw type is used to affix a BPTB autograft during an ACL reconstruction in a cadaveric goat stifle joint. First, a biomechanical evaluation of the Milagro interference screws was performed. Three cyclic creep tests were used to determine total residual elongation, which is a way to quantify graft slippage past the interference screw in the bone tunnel. Additionally, during a load-to-failure test, stiffness, ultimate load, and ultimate elongation were determined. This provided data to help make design improvements to the 2nd generation Mg screws and will allow a future in-vivo study to be performed.

During the reconstructions, a noted problem was breakage of the Milagro screws, and this limited the number of specimens tested. Since biomechanical data was only collected from four samples, statistical significance could not be achieved. However, thus far, it appears as though Milagro screws result in comparable or superior graft slippage and initial fixation strength when compared to titanium, pure Mg, and alloy AZ31 screws. The total residual elongation, or slippage, of the FATCs affixed with Milagro screws and titanium screws was similar. Additionally, Milagro screws resulted in a higher stiffness and ultimate load than any of the other screw types. These results suggest that the Milagro screws have a superior design on which that of the 2nd generation Mg screws can be based. Milagro screws are longer than titanium screws or either of the Mg protypes, which could have contributed to the increased stiffness and ultimate load [14]. Additionally, the thread design of the Milagro screws, such as the thread height, may have played a role in the comparable slippage and superior stiffness and ultimate load [15].

Future work will include completing the in-vitro biomechanical testing of the Milagro and 2nd generation Mg screws with an appropriate number of samples. Additionally, an in-vivo test comparing the graft fixation and graft slippage of both screw types in a goat model will be performed. Lastly, based on the results of the in-vitro and in-vivo tests, design improvements will be made to the Mg screws so that they can eventually become an effective alternative to current interference screw options.

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2011 Summer Student Symposium

Hunter presenting to the summer Students

The director of the symposium, Jonquil, introducing Joe to give his presentation

Dr. Woo's closing remarks

Musculoskeletal Research Center
Summer 2011

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