International Journal of Engineering Research & Technology (IJERT)
ISSN: 2278-0181
Vol. 2 Issue 9, September - 2013

Finite Element Analyses of ACL Interference Screws: A Review

Kevin M. Hawkins, M.S., Jonquil R. Flowers, M.S., Matthew B. McCullough, Ph.D.
Department of Chemical, Biological, and Bioengineering, College of Engineering,
North Carolina A&T State University, Greensboro, N.C.

Abstract

The anterior cruciate ligament (ACL) is the most often injured knee ligament with well over 100,000 injuries in the United States, annually. ACL reconstruction is the most common surgical treatment and involves replacing the damaged ACL with a soft tissue graft. ACL interference screws are often used to fix the graft into place. The early success of the reconstruction surgery depends on the graft fixation method. Several investigators have studied the effect of the properties of interference screws (i.e. screw design parameters) on their performance. Some have employed the use of finite element analysis (FEA), a computational method that can simulate loading conditions on differing screw designs. This review gives an overview of ACL interference screw usage and design as well as an in depth review of studies that have used FEA to assess ACL interference screw performance. Finally, future directions of how FEA can be used to optimize interference screw design and implementation are discussed.

1. Introduction

The anterior cruciate ligament (ACL) is located in the center of the tibiofemoral joint and is a major stabilizer of the knee [42]. Its primary function it to restrain anterior translation of the tibia with respect to the femur as well as aid in the control of the six degrees of freedom of knee motion [42, 43]. The ACL is the most often injured knee ligament which commonly occurrs during sporting and work related activities [37]. A ruptured ACL has limited healing potential and can lead to functional instability and long term complications [4]. Thus, surgical reconstruction is a common treatment where the damaged ACL is replaced with a soft tissue graft (i.e. BPTB, hamstring) which is fixed into bone tunnels with interference screws.

Interference screws have been described as the standard of graft fixation for ACL reconstruction. They act as a wedge to ensure fixation of the graft in the bone tunnels [18]. Many studies have investigated the effect of screw design on its performance [2, 5, 26, 29, 35]. Finite element analysis (FEA) has been employed for such investigation. This method has proved to be a cost effective and beneficial way to drive the design of ACL interference screws [16]. This review gives an overview of ACL interference screw usage and design as well as an in depth review of studies that have used FEA to assess ACL interference screw performance.

2. ACL reconstruction and graft fixation

There are well over 100,000 ACL injuries in the United Sates annually. These injuries can be caused by direct contact to the knee by an object or individual or by a non-contact mechanism such as an abrupt change in direction. Once the ACL is injured, there is limited potential for healing on its own.

There are several treatment options which include conservative treatments using bracing and rehabilitation and surgical treatments. The most common surgical treatment is ACL reconstruction which involved replacing the injured ACL with a soft tissue graft. Tibial and femoral bone tunnels are drilled through which the replacement graft is pulled and fixed into place. There are many surgical considerations that must be made to ensure surgical success including graft selection, tunnel size, tunnel placement, graft tensioning, and the graft fixation method.

The graft fixation method is the only entity that keeps the graft in place until there is biological incorporation of the graft. Graft fixation must withstand physiologic forces generated during activities of daily living and a progressive rehabilitation program. Additionally, the fixation method must be biocompatible. Thus, graft fixation is critical to the early success of ACL reconstruction. There are many commercially available graft fixation devices such as staples, screw posts, the EndoButton, press-fit bone plug, and washerplates. However, interference screws are the standard of graft fixation for ACL reconstruction.

Currently, there are two classes of biomaterials with which ACL interference screws are manufactured, metallic (i.e. titanium) and polymer such as poly-L-lactic acid (PLLA). Metallic interference screws have a high mechanical strength and provide good fixation strength. However, they are permanent implants and cases of graft laceration from sharp threading and MRI interference have been reported. The polymer interference screws gained popularity because they are designed to degrade over time and be replaced by the native tissue. But, the degradation is inconsistent and sometimes incomplete and breakage during and after implantation has been reported.

While metals and polymers have very different material properties, their fixation strengths have been found to be comparable. This can be contributed to the varying and strategic design of the interference screws. Thus, it has been shown that careful consideration must be given to the design parameters of interference screws for optimal performance.

3. Mechanics of screw function

3.1 Basics of screw design

The basic function of any orthpaedic screw is to provide interfragmentary compression. For example, for fractured bone injuries medical screws serve to compress bone fragments together thus increasing the friction between mating fragments, as well as, minimize micromotion[11]. This allows the fragments to rest relatively undisturbed so that the bones can efficiently heal together. In the case of interference screws for ACL reconstruction surgery the same concept applies except that interference screws behave more "wedge-like" in order to compress the biological tissue graft to the boney tunnel. However, this "wedge-like behavior still serves to provide increased friction and reduced motion within the tunnel where the graft is to heal and incorporate.

The rather unique design of screws, referring to the threads, allows them to be efficiently inserted and torqued to provide varying levels of compression and position within the body. The threads of a screw are particularly important to screw design because they provide axial resistance to prevent various extraction forces to remove the screw once it has been inserted. There are generally two different types of screw threads for bone screws: cortical which are used for fixation within the stiffer cortical bone and cancellous, used for fixation within the more brittle and porous cancellous bone. The primary difference in the two types of thread are that cancellous threads tend to be deeper, smaller in thickness, and the pitch is often larger[20]. The reason for this is that the porous structure of cancellous bone is broken up during insertion and this design allows more space for the broken particles to fall in. This occurance alludes to another important property the threads of bone screws in particular provide. As mentioned, as the threads are advanced fragments of bone are broken up, filled into the porous structure, and compressed thus producing a stiffer more dense material to hold the screw. This is important because cancellous bone is very brittle and does not provide very good structural support, but, because of this phenomenon a better structural support is created for the screws. It is also for this reason that pre-tapping of cancellous bone is not recommended. Pre-tapping creates the threads in the bone for the screw and removes much of the debris. This generally enlarges the hole and has been shown detrimental to cancellous bone insertion. As a result many cancellous bone screws are self-tapping, distinguished by a flute on the tip which is designed to cut away bone and allow for material to be removed. Alternatively cortical bone screws generally require pre-tapping in order to prevent cracking in the bone[10, 12]. Another property to note is that many screws must cross through both cancellous and cortical bone. For these cases these screws are designed with an unthreaded portion that can pass through without interacting with one section of bone and provide fixation within another.

Because medical bone screws are inserted into small holes in the body where visiblility and control is minimal, an importan design aspect is cannulation, a hollow core travelling along the screw's axis. This is very common in orthopaedic screws, especially interference screws for ACL reconstruction because it allows for maximum accuracy during insertion of the screw asit can guided along a wire.

3.2 Specific parameters and their functions

3.2.1 Screw design. As mentioned the primary purpose of bone screws is to provide interfragmentary compression, however there are many parameters by which screws are measured and characterized. The end of the screw that is generally inserted first is referred to as the tip while the other end is known as the head. The thread is characterized by the remaining material after a helical groove has been removed from a cylindrical shaft. This groove travels from the tip of the screw to a prescribed distance along the shaft referred to as thread runout where it gradually merges into the outer surface of the shaft. When seen from profile the threads appear as a uniform, repeating wavy or crimp-like pattern. The outermost surface of a screw thread profile is known as the crest with the root being its counterpart at the bottom most surface. The distance between two consecutive crests of a screw thread is known as the pitch. The profile width a single thread with one crest is known as the thread width. Typically, depending on the shape that will be discussed later, there is a straight line connecting the crest and the root called the flank. There are two flanks to either side of the thread profile. Adjacent flanks are characterized by the respective angulation between them, however, as the specific angulation of the flanks between consecutive crests are not always equal; these flanks are often characterized by their respective flank angle. The specific flank angles between threads are commonly known as the proximal and distal half angles. The distal half angle is the angle of the flank between two threads that is closest to the tip while the proximal half angle is the one closest to the head.

3.2.2 Pullout Factors. The parameters discussed above describe the general thread shape of a screw, however; screws are also made up of several specific parameters primarily looked at to characterize performance which includes the lengths, diameters, pitches and depths, thread shapes and tapers. One of the key factors to screw performance in medical bone screws is its ability to maintain a rigid fixation until proper heal is able to take place. The greatest contributing factor to screw performance, especially within cancellous bone, is the apparent density and shear strength of the bone [10, 17, 24]. The reason for this is because bone, in most cases, is the weaker material, when compared to the polymer and metallic screws, and is therefore the material expected to fail first. However, the properties of bone between different individuals vary greatly, and these properties cannot be easily controlled. Because of these uncontrollable properties, design parameters of the screw must be utilized to increase hold and performance within the bone. In fact a commonly used formula to predict shear failure force in polyurethane foam, commonly used to ideally mimic human bone, assumes the bone as the failing material in the shape of a cylindrical section of the bone the approximate major diameter of the screw and is therefore based on the product of the materials ultimate shear and the screw's thread shear area. The formula, seen below, was based upon three screw design factors that are widely considered contributors to pullout, diameter, length of engagement, and a thread shape factor (TSF) [1, 6, 10, 23, 24, 36, 39, 41].

$$F_s = \{S * L * D_{major}\} * TSF$$

where:

$$S = ultimate\ shear\ of\ bone$$
$$L = purchase\ length\ of\ screw$$
$$D_m = major\ diameter\ of\ screw$$
$$TSF = thread\ shape\ factor$$

3.2.3 Diameter. Hughes and Jordan et al. 1972 claimed that the holding power of a screw is not dependent on the screw's material strength but rather the shear strength of the bone and the diameter of the screw[24]. The two diameters most commonly used to characterize a screw are the major diameter and the minor, or root diameter. The major diameter is described as the largest diameter characterized by the distance from crest to crest on the profile view of a thread. Similarly the root diameter is characterized by the distance from root to root in the profile view. The specific effect of the diameters on screw performance has been somewhat of a debate as there are two of them to consider. The study by Hughes and Jordan found the root diameter to yield a more significant impact on performance especially regarding extraction resistance. This goes back the equation mentioned above which assumes failure as material being removed as a cylinder the exact diameter of the screw. While the major diameter decides the idealized diameter of the removed material, the root diameter determines the amount of support material existing between the threads to provide axial extraction resistance. However, Decoster et al 1990 found the major diameter to have a more linear effect, with a decreased minor diameter contributing to smaller but still significant increases in strength. Additionally, the ratio of major to root diameter was found to have a small but, significant effect. Having a larger difference between these two parameters would lead to a greater surface area between the bone and screw and therefore a greater amount of contact potentially providing a greater hold. This ratio is also representative of the thread depth, another parameter characterized by half of the difference between the major and root diameters. A larger depth allows a greater amount of material or broken bone fragments to rest between the bone and the threads. The importance of this ratio has also potentially been shown in studies where it has been discovered that bioabsorbable polymer screws which tend to have a larger depth than their metallic counterparts have often times performed better [3, 41]. The major diameter has a drawback tin that it has been found to be limited by the size of the bone, or screw tunnel in the case of an interference screw[17]. On the other hand, the root diameter is often limited by cannulation. This can be particularly critical when dealing with medical bone screws, especially interference screws, which need to be accurately guided and most often contain cannulations as mentioned earlier[10]. With respect to interference screw fixation it is believed that major diameter plays a more significant role as their fixation function is based upon compressing graft material against a tunnel wall, and as such a larger diameter will better enable these screws to provide a greater compressive load[27, 34, 41].

3.2.4 Length. The length of a screw is defined as the distance from the head to the tip screw. However, because the thread and its length of engagement into the bone is one of the most important aspects of a screw, purchase length is also characterized as the length of only the threaded portion of a screw. The biggest advantage of screw length, or more specifically purchase length is that it can directly increase the amount thread engagement with the bone. Several studies have observed the effect screw length on interference screw fixation [36, 41], and have shown that increasing the length of screws for soft tissue graft fixation, significantly increases the stiffness and pullout, even more so than the effect of the diameter. The reason for this is believed to be because longer screws can provide addition contact with cortical bone in addition to the traditionally seen cancellous bone contact and additionally provide interference compression over a larger area, thus greatly increasing the stiffness. While this is believed true for soft tissue grafts it has, been observed that increasing the length of screws used for tissue grafts with hard bone plugs on the end provides a much less significant effect[41].

3.2.5 Taper. Another parameter describing screw design, taper, refers to the narrowing of a screw diameter as it draws towards the tip. Generally shorter portions of the tips of most screws are tapered in order to allow for ease of insertion; however it is believed to have another purpose. Taper is a critic aspect of interference screw design and greatly plays into their "wedge-like" behavior. Screw taper is another factor believed to contribute to the seemingly slight superiority of polymer screws. Metallic screws are traditionally straighter screws with the exception of the tapered tips. Polymer screws are often fully tapered, from head to tip, because they, form a material standpoint, are mechanically weaker, and this allows more precise insertion while reducing the risk of breakage and insertion failure [36]. Tapers were initially thought to weaken pullout performance because they reduced the number of threads fully engaged and thus the effective purchase length of the screw. However, the tunnels were often pre-tapped to reduce the risk of screw breakage during insertion which also weakens fixation for the reasons discussed earlier [3]. There have been studies which have shown screw taper to increase the overall rigid fixation of interference screws [9, 33, 36]. It is believed this design playing into their wedge-like characteristics allows the screws to be inserted deeper and with more ease thus constantly increasing the friction and compression with every turn. A particular study by Mann et al. 2005 took carefull precautions to neglect the effects of enhanced insertion by not tightening the screw to the specified torques as traditionally done in practice, but inserting a tapered screw into a tapered tunnel the same depth as a non-tapered screw was inserted into a standard tunnel. In this study the tapered screw was still observed providing a superior fixation [33].

3.2.6 Pitch. The pitch of a screw is a parameter characterized by the distance between consecutive crests on a thread profile. The pitch of a screw though noted as not providing as great an effect on screw performance, has been speculated as having a greater clinical relevance because it is not limited by bone size or cannulation [17]. Screw pitch is able to affect the number of threads per unit length which can alternatively increase the amount of thread engagement with the bone, thus having a significant impact on axial extraction resistance. Studies have shown cancellous bone in particular as being very sensitive to pitch[28] generally shown as providing better resistance with a finer pitch [20]. This has been attributed to the fact that it will provide greater compaction of the bone fragments, as there is lesser space for them to move to, however it must be considered carefully as there still must be adequate space for the fragments.

3.2.7 Thread Shape Factor (TSF). A parameter known as the thread shape factor was mentioned earlier in a predictive equation. This refers to a dimensionless number that can essentially be defined as the ratio of the thread pitch and the thread depth times a constant, plus one half, and it can decrease the predicted shear force depending on the ratio seen in the following expression:

$$TSF = 0.5 + 0.57735\ {}^{d}/_{p}$$

where:

$$d = thread\ depth$$
$$p = thread\ pitch$$

Based on this relationship from the equation, either by increasing the depth, and /or decreasing the pitch, the performance of the screw can be increased. Experimental results have correlated with this relationship though an exact ratio of maximum efficiency has yet to be determined [1].

3.2.8 Thread Shape. The final aspect of screw design to be discussed is the thread shape, which characterizes to the overall profile shape of the screw thread. Among the most common thread shapes seen and discussed for medical screws are the standard V-thread, the trapezoidal, the buttress, and the square thread designs [8, 21, 30, 31], seen below in **Error! Reference source not found.**.

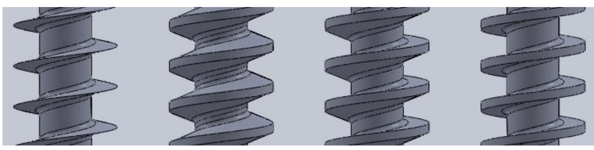

Figure 1. CAD models of common thread shapes. From left to right: V-thread, trapezoidal, buttress, square.

Thread shape has the ability to significantly influence screw performance while other factors may be constrained for geometrical reasons[21]. This is of particular importance when considering the sometimes superior pullout capabilities of polymer screws compared to their metallic counterparts. Metallic screws tend to have sharper threads as they are commonly intended to self-thread, or cut, through bone. In interference screw this is particularly true of surgeries utilizing bone plug grafts. Polymer interferences screws are more often used with soft tissue grafts such as hamstring and quadriceps tendons because of their naturally rounded profiles; metallic screws often must be blunted, or rounded to avoid cutting the soft tissue grafts [3]. Alternatively thread width can play a valuable role as a wider profile will have a reduced chance for laceration. Several different thread shapes have been tested in the literature to see which designs provided the best fixation. V-shaped threads are perhaps the most common overall thread shape as they are used in typical machine screws. In a recent study by Gracco et al 2012, the buttress design was found to exhibit the best extraction resistance. Alternatively a study from Lee et al 2010 found that the square design exhibited the least amount of stress on the bones, possibly indicating a greater resistance. In any case, the primary differences in the particular thread shapes discussed above is the respective flank angles associated with them. The trapezoidal and v-shaped threads designs have proximal and distal half angles of similar dimensions, all a generally equal degrees. However the buttress has one of the half angles, the proximal, at a minimal degree. Square designs generally have both half angles at minimal to zero degrees. A study by Wang et al 2009 observed the effects of the flank angles, specifically the proximal half angle, on extraction resistance. The specific proximal half angles measured in this study were 0, 30, and 60 degrees. It was found from the study that while a 60 degree proximal half angle yielded significantly weaker extraction resistance, a 30 degree angle increased the resistance when compared to that of the zero degreed angle. The reasoning was that a 30 degree angle allowed a more even distribution of force from the screw to the surrounding bone, while a vertical or zero degree proximal half angle focused the force in the peaks of the thread [40]. Krenn et al 2008, however looked at the effects of the flank overlap area, (FOA). The FOA relates to the amount of material the shape will allow in between the screw threads. As discussed a larger amount of compressed material means a stiffer material to support the screw and this can be slightly regulated by thread shape. An increased FOA is often achieved by reducing the minor diameter and/or narrowing the threads; however at a certain point of narrowing threads it becomes detrimental and the bone fails more readily.

4. Existing FEA studies of the ACL screw

4.1 Brief history and overview of FEA

Finite element analysis, (FEA) has increasingly become a popular numerical tool for analyzing problems in stress and displacement analysis, fluid flow, heat transfer, magnetics and many other subjects. Recently it has become an ever more powerful tool in orthopaedic surgery and traumatology due to its ability to model the behavior of sophisticated physiological systems regardless of geometric complexity. FEA can even predict the values of stress in the bone surrounding the implant to predict changes in the stress distribution [22]. The basic principles of FEA originated in 1941 with the advances in the structural analysis of aircrafts during WWII. It’s popularity began grow in the 1960s when the term ‘finite element’ was first coin by R.W. Clough and the first book on FEM was written by Zienkiewicz and Cheung in 1967 [7]. Simply put FEA is a numerical technique where geometry of particular interest is divided into a ‘finite’ number of simplified shapes, or elements. Material properties and additional governing relations are assigned to these elements to produce equations which are solved at nodes, points which connect the elements at their corners. As various applicable loading and boundary conditions are established, these unknowns can be solved in relation to one another at the nodes [7].

4.2 Current Studies

Several studies have successfully utilized FEA in the study of orthopaedic bone screws and thus various techniques have been incorporated. The most commonly used modeling techniques have been two dimensional (2D), axisymmetric and three dimensional (3D) simulations. Two dimensional simulations assume the modeled geometry as an idealized infinitely flat plane, while 3D simulations incorporate a full three dimensional geometry of the screw and/or surrounding anatomy. Axisymmetric simulations are modeled as 2D simulations assumed to be fully revolved around an axis to produce a 3D effect. With screws this inadvertently idealizes the threads as simple rings around a shaft. Three dimensional simulations are the most robust modeling technique but require large numbers of elements and computational resources, leading to large numbers of equations and exceedingly long solution times. It is not uncommon for the simulations to take days and weeks to solve. Several other problems can spawn from this, for example, if the number of equations exceeds the processors memory capabilities then a final converged solution will not be able to present itself. These issues are often referred to as the computational costs of running a FEA simulation. Obviously a method that could consider an infinite number of elements would be desired as it would be the most accurate; however this is an unreasonable request with the current known techniques despite the great advancements in computing power. Still the more elements a geometry can be divided in to, the more accurate the results, therefore, convergence studies are often turned toward to determine the minimum number of elements required to render an accurate, relatively unchanging solution in a reasonable amount of time. To minimize computational costs 2D simulations are often used at the expense of missing stresses not included on the specific plane modeled. Alternatively for this reason axisymmetric models are considered to be the best of both worlds [44]. As stated this modeling technique idealizes the threads as simplified rings which has been regarded by many as having a minimal effects on the results [25, 32].

A common trend in finite element models currently used today for modeling orthopaedic screws is to set the bone to be modeled as a linear elastic isotropic material. Despite bone not naturally behaving as linear elastic and additionally possessing anisotropic materials properties, highly dependent on loading direction, this setting allows for fairly simple quick converging solutions. Additionally despite not describing the precise behavior of natural 'wet' bone this setting has still been shown has a good predictor of natural bone and widely considered acceptable [13-15]. In summary the benefits of FEA include the ability to test various complex geometries and scenarios without the need for expensive equipment, while presening results of stress distribution. Chizari et al has performed various FEA studies analyzing the distribution of stresses in human bone due to the insertion and removal of orthopaedic screws. Studies from Chang et al 2012 and Lee et al 2010 have each performed FEA tests assess the effects of various screw design parameters on pullout resistance. From the studies they were able to see where stresses generally grouped in the bone as well as locations and values of the maximum extraction resistance.

As mentioned the computational costs of FEA can severely limit the robustness of studies. Thus it is not unheard of for a model to inaccurately reflect screw performance. For this reason the importance of verification and validation is stressed. Verification asks the question, "are the equations being solved right", and validation additionally asks "if the right equations are being solved" [19].

4.3 Verification

Verification addresses whether the correct numerical methods and formulations are being used in a particular model. In most commercial and open source FEA software packages there are various methods that can be used to solve a problem such as those mention above. Another example involves the penalty methods for contact problems commonly used to model the screw and biological tissue interaction. Pure penalty is a widely accepted setting for solving problems of frictional contact. It utilizes a penalty term which assumes an imaginary value of penetration of one surface into another and updates this value, during each iteration of the solution. Another FEA contact setting known as the Lagrange method used this same technique the pure penalty method but adds an additional multiplier which increases the accuracy. The Lagrange method generally tends to be more accurate but can also lead to ill converging solutions. The penalty method however tends to lead to inaccurate solutions[38]. In the case of frictional contact it may be more important to obtain a less accurate converged solution than an ill converge one, one must carefully consider what is needed for their particular simulation. In summary model verification is extremely important. There are several ways to verify a model, one example is to run, simplified simulations that can be ran testing the solution of interest to known solutions, or checking for differences in results using different algorithms or methods [19].

4.4 Validation

As stated above validation involves determining whether the model accurately reflects the mechanics of the problem. This is an especially important area of concern as it determines whether the finite element model yields any justifiable results. There are several ways to validate a model as well. The perhaps the most popular method is with physical testing. An experimental validation test can be performed to confirm if the output from the FEA simulation directly correlates to the actual real-world results. Another perhaps less expensive and indirect method is comparing the output of the FEA with very similar results found in literature. For instance if the literature states that in experimental testing applying a prescribed load to a screw will cause it to displace x amount then a FEA using similar boundary conditions should yield a similar displacement. However, one must be careful of differences in the geometries as well validity of studies. Among the important considerations for validation are knowing exact outputs that you want to assess, something that both the model and the validation data can each directly assess. In addition it is important to consider which assumptions for the model, boundary conditions, loads, and materials properties, to use. Any and all of the assumptions used in the model must be justified, and any information or data put into the model should come from a justifiable source. Uncertainties should also be noted and assessed, as well as the degree to which the model accurately reflects the system of interest [19].

5. Future Directions

There are many opportunities for future work in the realm of modeling orthopaedic screws. For example the area of medical screws is still advancing as new materials such as magnesium based alloys are being developed. There are a few design parameters can still to be explored as it has been shown from studies such as Chapman et al. 1996 that depth and pitch have some sort of interaction effect within the TSF. Future studies could seek to further understand the effect of pitch and depth as they relate to the TSF further in orthopaedic screw utilizing new developing materials.

It was also mention that the factors such as the proximal half angle have been assessed however there is still room to discern any possible contribution of the distal half angle to screw performance or even any relationship between the two. Buttress designs have commonly been seen performing better yet it is the one design out of the primary thread shapes mentioned which has unequal proximal and distal half angles.

In addition the is still much room in the future of modeling medical screws to focus on much more anatomical modeling incorporating the CT data for actual representation of bones complete with multiple cortical and cancellous bone material layers including fluids and soft tissues. These kinds of models could give more insight to screw performance at various anatomical angles and loading scenarios.

Finally because of the many degradable alloys beginning to emerge models incorporating time dependent degradation while under cyclic loading is a great area of concern regarding how a screw will hold over time as it is being degraded and stressed from daily living activities. These types of models will give insight to the structural integrity of screws over time as they degrade. These models would also give vital insight into how important it is for the bone to heal and the screws to degrade in a synchronized manner. In addition because many of the degradable alloys being developed must be controlled to an extent by surface coatings it would be beneficial to involve layers in future models to represent these surface coatings.

6. Acknowledgents

The support from the NSF Engineering Research Center for Revolutionizing Metallic Biomaterials (ERC-RMB) is gratefully acknowledged.

6. References

1. Asnis, S. E., Ernberg, J. J., Bostrom, M. P., Wright, T. M., Harrington, R. M., Tencer, A., & Peterson, M. (1996). Cancellous bone screw thread design and holding power. *J Orthop Trauma, 10*(7), 462-469.
2. Barber, F. A., Elrod, B. F., McGuire, D. A., & Paulos, L. E. (1995). Preliminary results of an absorbable interference screw. *Arthroscopy: The Journal of Arthroscopic & Related Surgery, 11*(5), 537-548.
3. Beevers, D. J. (2003). Metal vs bioabsorbable interference screws: initial fixation. *Proc Inst Mech Eng H, 217*(1), 59-75.
4. Beynnon, B. D., Johnson, R. J., Abate, J. A., Fleming, B. C., & Nichols, C. E. (2005). Treatment of Anterior Cruciate Ligament Injuries, Part 1. *The American Journal of Sports Medicine, 33*(10), 1579-1602.
5. Brown, C. H., Hecker, A. T., Hipp, J. A., Myers, E. R., & Hayes, W. C. (1993). The biomechanics of interference screw fixation of patellar tendon anterior cruciate ligament grafts. *The American Journal of Sports Medicine, 21*(6), 880-886.
6. Brown, G. A., McCarthy, T., Bourgeault, C. A., & Callahan, D. J. (2000). Mechanical performance of standard and cannulated 4.0-mm cancellous bone screws. [Research Support, Non-U.S. Gov't]. *J Orthop Res, 18*(2), 307-312.

7. Chandrupatla, T. R., & Belegundu, A. D. (2002). *Introduction to finite elements in engineering* (3rd ed.). Upper Saddle River, N.J.: Prentice Hall.
8. Chang, J. Z.-C., Chen, Y.-J., Tung, Y.-Y., Chiang, Y.-Y., Lai, E. H.-H., Chen, W.-P., & Lin, C.-P. (2012). Effects of thread depth, taper shape, and taper length on the mechanical properties of mini-implants. *American Journal of Orthodontics and Dentofacial Orthopedics, 141*(3), 279-288.
9. Chang, J. Z., Chen, Y. J., Tung, Y. Y., Chiang, Y. Y., Lai, E. H., Chen, W. P., & Lin, C. P. (2012). Effects of thread depth, taper shape, and taper length on the mechanical properties of mini-implants. *Am J Orthod Dentofacial Orthop, 141*(3), 279-288.
10. Chapman, J. R., Harrington, R. M., Lee, K. M., Anderson, P. A., Tencer, A. F., & Kowalski, D. (1996). Factors Affecting the Pullout Strength of Cancellous Bone Screws. *Journal of Biomechanical Engineering, 118*(3), 391-398.
11. Chapman, M. W. (2001). *Chapman's orthopaedic surgery* (3rd ed.). Philadelphia: Lippincott Williams & Wilkins.
12. Chapman, M. W. (2001). Principles of Internal and External Fixation *Chapman's Orthopaedic Surgery* (3rd ed.). Philadelphia: Lippincott Williams & Wilkins.
13. Chatzistergos, P. E., Magnissalis, E. A., & Kourkoulis, S. K. (2010). A parametric study of cylindrical pedicle screw design implications on the pullout performance using an experimentally validated finite-element model. *Medical Engineering & Physics, 32*(2), 145-154.
14. Chizari, M., Snow, M., & Wang, B. (2009). Post-operative analysis of ACL tibial fixation. *Knee Surgery, Sports Traumatology, Arthroscopy, 17*(7), 730-736.
15. Chizari, M., Wang, B., & Snow, M. (2007). Experimental and numerical analysis of screw fixation in anterior cruciate ligament reconstruction: International Association of Engineers.
16. Chizari, M., Wang, B., Snow, M., & Barrett, M. (2008). Experimental and Numerical Analysis of Screw Fixation in Anterior Cruciate Ligament Reconstruction. [Article]. *AIP Conference Proceedings, 1045*(1), 61-70.
17. DeCoster, T. A., Heetderks, D. B., Downey, D. J., Ferries, J. S., & Jones, W. (1990). Optimizing bone screw pullout force. *J Orthop Trauma, 4*(2), 169-174.
18. Dworsky, B. D., Jewell, B. F., & Bach Jr, B. R. (1996). Interference screw divergence in endoscopic anterior cruciate ligament reconstruction. *Arthroscopy: The Journal of Arthroscopic & Related Surgery, 12*(1), 45-49.
19. Erdemir, A., Guess, T. M., Halloran, J., Tadepalli, S. C., & Morrison, T. M. (2012). Considerations for reporting finite element analysis studies in biomechanics. *J Biomech, 45*(4), 625-633.
20. Gausepohl, T., Möhring, R., Pennig, D., & Koebke, J. (2001). Fine thread versus coarse thread: A comparison of the maximum holding power. *Injury, 32, Supplement 4*(0), 1-7.
21. Gracco, A., Giagnorio, C., Incerti Parenti, S., Alessandri Bonetti, G., & Siciliani, G. (2012). Effects of thread shape on the pullout strength of miniscrews. *American Journal of Orthodontics and Dentofacial Orthopedics, 142*(2), 186-190.
22. Herrera, A., Ibarz, E., Cegonino, J., Lobo-Escolar, A., Puertolas, S., Lopez, E., . . . Gracia, L. (2012). Applications of finite element simulation in orthopedic and trauma surgery. *World J Orthop, 3*(4), 25-41.
23. Herrera, A., Martinez, F., Iglesias, D., Cegonino, J., Ibarz, E., & Gracia, L. (2010). Fixation strength of biocomposite wedge interference screw in ACL reconstruction: effect of screw length and tunnel/screw ratio. A controlled laboratory study. *BMC Musculoskeletal Disorders, 11*(1), 139.
24. Hughes, A. N., & Jordan, B. A. (1972). The mechanical properties of surgical bone screws and some aspects of insertion practice. *Injury, 4*(1), 25-38.
25. Karunratanakul, K., Schrooten, J., & Van Oosterwyck, H. (2010). Finite element modelling of a unilateral fixator for bone reconstruction: Importance of contact settings. *Med Eng Phys, 32*(5), 461-467.
26. Kohn, D., & Rose, C. (1994). Primary Stability of Interference Screw Fixation. *The American Journal of Sports Medicine, 22*(3), 334-338.
27. Kohn, D., & Rose, C. (1994). Primary stability of interference screw fixation. Influence of screw diameter and insertion torque. [Comparative Study]. *Am J Sports Med, 22*(3), 334-338.

28. Kong, L., Zhao, Y., Hu, K., Li, D., Zhou, H., Wu, Z., & Liu, B. (2009). Selection of the implant thread pitch for optimal biomechanical properties: A three-dimensional finite element analysis. *Advances in Engineering Software, 40*(7), 474-478.
29. Kotani, A., & Ishii, Y. (2001). Reconstruction of the anterior cruciate ligament using poly-l-lactide interference screws or titanium screws: a comparative study. *The Knee, 8*(4), 311-315.
30. Krenn, M. H., Piotrowski, W. P., Penzkofer, R., & Augat, P. (2008). Influence of thread design on pedicle screw fixation. *Journal of Neurosurgery: Spine, 9*(1), 90-95.
31. Lee, C.-C., Lin, S.-C., Kang, M.-J., Wu, S.-W., & Fu, P.-Y. (2010). Effects of implant threads on the contact area and stress distribution of marginal bone. *Journal of Dental Sciences, 5*(3), 156-165.
32. MacLeod, A. R., Pankaj, P., & Simpson, A. H. (2012). Does screw-bone interface modelling matter in finite element analyses? [Research Support, Non-U.S. Gov't]. *J Biomech, 45*(9), 1712-1716.
33. Mann, C. J., Costi, J. J., Stanley, R. M., & Dobson, P. J. (2005). The effect of screw taper on interference fit during load to failure at the soft tissue/bone interface. *Knee, 12*(5), 370-376.
34. Namkoong, S., Heywood, C. S., Bravman, J. T., Ieyasa, K., Kummer, F. J., & Meislin, R. J. (2006). The effect of interference screw diameter on soft tissue graft fixation. *Bull Hosp Jt Dis, 63*(3-4), 153-155.
35. Pena, F., Grøntvedt, T., Brown, G. A., Aune, A. K., & Engebretsen, L. (1996). Comparison of Failure Strength Between Metallic and Absorbable Interference Screws. *The American Journal of Sports Medicine, 24*(3), 329-334.
36. Selby, J. B., Johnson, D. L., Hester, P., & Caborn, D. N. (2001). Effect of screw length on bioabsorbable interference screw fixation in a tibial bone tunnel. *Am J Sports Med, 29*(5), 614-619.
37. Shimokochi, Y., & Shultz, S. J. (2008). Mechanisms of Noncontact Anterior Cruciate Ligament Injury. *Journal of Athletic Training, 43*(4), 396-408.
38. Ştefancu, A. I., Melenciuc, S. C., & Budescu, M. (2011). Penalty Based Algorithms for Frictional Contact Problems.
39. Tsai, W. C., Chen, P. Q., Lu, T. W., Wu, S. S., Shih, K. S., & Lin, S. C. (2009). Comparison and prediction of pullout strength of conical and cylindrical pedicle screws within synthetic bone. [Comparative Study]. *BMC Musculoskelet Disord, 10*, 44.
40. Wang, Y., Mori, R., Ozoe, N., Nakai, T., & Uchio, Y. (2009). Proximal half angle of the screw thread is a critical design variable affecting the pull-out strength of cancellous bone screws. *Clin Biomech (Bristol, Avon), 24*(9), 781-785.
41. Weiler, A., Hoffmann, R. F., Siepe, C. J., Kolbeck, S. F., & Sudkamp, N. P. (2000). The influence of screw geometry on hamstring tendon interference fit fixation. [Research Support, Non-U.S. Gov't]. *Am J Sports Med, 28*(3), 356-359.
42. Woo, S. L.-Y., Hollis, J. M., Adams, D. J., Lyon, R. M., & Takai, S. (1991). Tensile properties of the human femur-anterior cruciate ligament-tibia complex. *The American Journal of Sports Medicine, 19*(3), 217-225.
43. Woo, S. L.-Y., & Wayne, J. S. (1990). Mechanics of the anterior cruciate ligament and its contribution to knee kinematics. *Applied Mechanics Reviews, 43*(5), S142-S149.
44. Zanetti, E. M., Salaorno, M., Grasso, G., & Audenino, A. L. (2009). Parametric analysis of orthopedic screws in relation to bone density. *Open Med Inform J, 3*, 19-26.