Theoretical and Numerical Analysis of Anterior Cruciate Ligament Injury and its Prevention

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Abstract

Anterior cruciate ligament (ACL) injury is one of major risks for most athletes. ACL injury can be caused by many risk factors such as anatomic risk factors, biomechanical risk factors and environmental risk factors. In this article, numerical and theoretical analysis is conducted to investigate biomechanical risk factors. An entire three-dimensional finite element knee model is built based on MRI data. Anterior Tibial Translations (ATT) at different knee flexion angles are simulated by finite element models. In the simulations, more attention is given to material properties of different knee components and their effects on ACL injury. Mechanical response of ACL during sport activities is highly determined by its viscoelastic properties. Unfortunately, viscoelastic properties of two bundles of ACL will change dramatically even with several hours? physical aging. As a consequence, ACL will experience mechanical ductile to brittle transition due to daily physical aging.

Index terms— ACL injury; Viscoelasticity; Physical aging; Prevention; Sports.

1 I. INTRODUCTION

here are more than 80,000 anterior cruciate ligament (ACL) tears annually in the United States. 70% of the injuries are the result of sports participation. An investigation of knee injuries in Iranian male professional soccer players shows that anterior cruciate ligament is the most commonly injured ligament of the knee [2]. A review of physical profiling for lacrosse players in the United States shows ACL injury is one of the most common injuries [3]. Global positioning systems (GPS) wearable technology was suggested to monitor athletes' physical profiles in sport fields. For volleyball players, a recent survey [4] shows ankles and T knees are the most likely injury areas. Among all ligaments of knee, ACL is most likely to get injured. ACL injury can be caused by many risk factors. They can be listed as environmental risk factors, anatomic risk factors, and biomechanical risk factors [5]. Environmental risk factors include knee braces effect, shoe and surface interaction and so on. Anatomic risk factors include the differences in femoral notch and ACL size, joint laxity, hip-trunk position and muscle development. Biomechanical risk factors are related to neuromuscular control and proprioception in joint stability. All these risk factors can lead to high stress and strain in ligaments. If strain and stress built on ACL during sport activities are higher than that ACL can bear, ACL will be torn apart.

2 Besides

anatomical effects, mechanical characterization of knee ligaments will be an important factor to understand knee injury. Stress and strain relationships for knee ligaments have been built for more than two decades. Since ligaments and tendons contain collagen fibrils, elastin, proteoglycans, glycolipids, water, and cells, mathematical modelling of mechanical behaviour of ACL is still in its infancy. Because of its complex components, mechanical behaviours of ACL need be modelled as anisotropic nonlinear materials. Recently Marchi et al. [13] built a hybrid constitutive model for medial collateral ligament (MCL) by superposing a slightly compressible, isotropic eight-chain Mackintosh network model with a phenomenological directional component. Since many knee injuries happen when attending sports, time dependent mechanical properties of ligaments cannot be ignored. Ligaments are normally under high speed stretch or twist when a football player jumps and falls on Abstract-Anterior cruciate ligament (ACL) injury is one of major Many experiments and virtual simulations have been performed to understand knee injuries, especially ACL injury [6] [7][8] [9]. Although these research efforts have yielded much information, they have not resulted in a clear understanding of the cause of ACL injuries. Detail review of researches on ACL injury mechanics and mechanical properties of ACL was given by McLean et al [10].In the past several decades, studies of knee injury were mostly from anatomy aspects, for example, muscles, cartilage and tibia and femoral notch [11]. Recently DS Simulia Inc. built a whole knee simulator add-on in Abaqus to consider impacts of all kinds of anatomy components [12].

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Theoretical and Numerical Analysis of Anterior Cruciate Ligament Injury and its Prevention the ground. Time dependent property of ACL is usually modelled as viscoelasticity, a combination of time independent springs and time dependent dashpots. Therefore, viscoelasticity becomes one of the centre topics to understand ACL tearing during sports.

Generally speaking, viscoelasticity is the theory that tell how solid materials flow like liquids. Although it is hard to see by eyes, nearly everything in the world flows. Solid material flows are usually so tiny that no one notices their existence. Mountains flow in millions of years. Metals flow in hundreds of years. Since everything flows, there is no reason to believe human being's tissue is not flowing. The evidence of tissue flow can be seen by comparing kid's smooth face and old man's wrinkled face. Natural aging of human beings from baby age to the aged is related to how bio-tissues degenerate from elastic state to plastic state. Mechanical behaviours of tissues and bones are changed from ductile state to brittle state permanently during person's entire life [14]. This entire life aging from elastic state to plastic state is irreversible. However, physical aging which happens daily on ACL can be reversed due to its short time aging. This inverted process is related to exercises such as stretching and cyclic movements of arms and legs. Therefore, we will see daily exercises have a potential function to delay human being's getting older.

There is a long history to apply viscoelastic theory to understand mechanical behaviours of human tissues. Fung [15] is one of the pioneers who firstly formulated Quasi-Linear Viscoelasticity (QLV), which combines elastic and time dependent components of a tissue's mechanical response using a hereditary integral formulation. By inputting hyper-elastic model [16] into linear viscoelasticity model [17], QLV can model strain stiffing behaviour as well as time effects of many soft tissues [18]. QLV model has also been used to understand mechanical behaviours of ACL [19]. However, for lots of tissues, viscosity is a function of applied strain level which limits the application of QLV [20]. For instance, ACL is a double-bundle anatomic structure which is made of anteromedial (AM) and posterolateral (PL) bundle. Two bundles show very different time-dependent properties in daily stress-strain range [19]. Compared to PL, mechanical response of AM bundle is much stronger since it has a more uniform collagen alignment. Accurately measuring ACL viscoelastic property is challenging because of its two bundle structures and heterogeneous strain-stress distribution during tension tests. Mechanical properties of ACL should be modelled as nonlinear viscoelastic models where relaxation time and relaxation modulus are not constant during its deformation. However, no widely accepted nonlinear viscoelastic model can be used. Today, Quasi-linear viscoelastic model is still widely used although it is still based on linear viscoelastic theory. Recently a new mathematical stress-strain framework was built for amorphous polymers [21]. It can be used to model mechanical behaviours of soft tissues. Nonlinear viscoelasticity and hyper-elasticity models were illustrated by a single mathematical equation. Physical aging of ACL can be added into this model too. Unfortunately, during past several decades, physical aging has been ignored by most if not all researchers when they conducted researches on measuring viscoelastic properties of ACL. Physical aging of ACL means its viscoelastic properties will be shifted with aging time if let ACL rest for several minutes or hours. Accuracy of measured viscoelastic properties of ligaments or tendons without considering physical aging is questionable. Measurement with physical aging considered provides a new approach to study nonlinear viscoelasticity of bio-tissues such as ligaments.

In the following three sections, we will have an extensive discussion of ACL injury from numerical and theoretical perspective. In the first section we will introduce numerical modelling of ACL by finite element analysis. A full three dimensional knee model is built and run by using a commercial finite element code, Abaqus. Most of the simulation works are to understand material property effect of each knee component on ACL injury under anterior tibial translation. In the second section, limits of finite element analysis of ACL injury will be briefly discussed. In the last section, theoretical analysis of ACL injury will be discussed based on physical aging and viscoelastic theory. It will be demonstrated that initially built-in stress on ACL due to physical aging is a big risk for ACL injury. ACL injury prevention is related to erasure of initial stress on ACL. Finally, a conclusion is given at the end of the article. In order to understand mechanism of ACL injury, a complete virtual knee model was built. In this finite element model, most knee joint components such as lateral meniscus, medial meniscus, tibia, tibial cartilage, femur, femoral cartilage, anterior cruciate ligament, lateral collateral ligament, medial collateral ligament, patella, fibula, fibula cartilage are included. Two dimensional geometries of knee components are got from MRI segmentations. These MRI pictures are imported into a commercial image processing software, Mimics, to get 3D geometry. Singular points on the surface of geometry during 3D model generation are removed by using smoothening process in Mimics. After smooth 3D geometries of all components are created in Mimics, they are imported into Abaqus. Assembling of all ligaments, bones, cartilages and so on is conducted in Abaqus. Assembled three dimensional finite element model is shown in Fig. 1.

Mechanical responses of different components are modelled as different material constitutive models. For bones such as femur, tibia and fibula, linear elastic materials are used. ACL is modelled either as isotropic or anisotropic hyperelastic models **??**22] [23]. PCL is modelled by using Arruda-Boyce model [24]. Anisotropic hyperelastic statistical mechanics model [25] is used for constitutive relationship of femoral and tibial cartilage. A user subroutine VUMAT is written based on this nonlinear anisotropic material model. Linear viscoelastic model is used for material properties of meniscus [26]. It is well known that mechanical responses for different areas of tibial and femoral cartilages are not uniform. In this simulation, if the same material model is assigned to all sections of cartilages, it is called homogeneous model, shown on the left side of One of key reasons that causes ACL injury is ATT. ATT will create a relative movement between femur and tibia. Velocity and acceleration due to this movement will build up a large strain and stress on ACL. In this numerical study, we will focus on modelling knee impact due to ATT. Simulation is performed in two steps. In the first step, femur is rotated by 30 or 45 degree related to tibia as shown in Fig. 3. In the second step, using the results from the first step, a 3g loading is applied to the top of femur which will create ATT. Simulations show ATT will not only create translations but also create certain rotations between femur and tibia. As a consequence, ACL will gain a large stretch force due to this translation. Shear and tensile strain will be built on ACL. Shear strain profile is shown in Fig. 4a for 3g loading when femur and tibia are in 30 or 45 degree flexion angles. The largest shear strain will happen at the end of ACL near to femur. Compared to maximum shear strain 18.4% in 30 degree angle, maximum shear strain for 45 degree angle is 24.1%. Shear strain around middle of ACL is small for both cases. The highest value near the centre of ACL is 7.5%. Since soft tissues such as femoral cartilage and tibial cartilage are generally heterogeneous, it is necessary to understand how heterogeneity affects ACL injury. In Fig. 5a and Fig. **??**b, tensile strain and shear strain profiles are compared for homogeneous and heterogeneous cartilage effects on ACL injury upon 3g loading impact. As shown in Fig. 5a, tensile strain in the center of ACL increases 30% in heterogeneous cartilage case where 7.2% for homogeneous cartilage and 9.7% for heterogeneous cartilage. For both cases, tensile Tensile strains from 30 and 45 degree cases are comparable, as shown in Fig. 4b. They are following the same trends as shear strain profiles. The highest tensile strain happens at the end near to femur side. Maximum strain near the end of femur side is 12%. Comparing to homogeneous case, shear strain in heterogeneous case has a dramatic increase in both middle and end part of ACL, as shown in Fig. **??**b. It increases from 18.4% to 40.8% at the end of ACL near to femur and 7.5% to 10.8% in the middle section of ACL. In general, after replacing homogeneous cartilage with heterogeneous cartilage, shear strain and tensile strain in ACL will increase from 30% to 110%. Finite element analysis has been continuously used to understand ACL injury during sport activities. One reason is that knee injury experiments can only be conducted by using deceased animal knees. But animal knee structures are very different from human beings'. Therefore, finite element analysis becomes an excellent alternative way to understand physics behind ACL injury. In finite element analysis, geometry can be easily changed. Material models can be selected for different knee components based on experimental material characterizations. Impact analysis as well as static analysis can be performed based on study interests.

Another concern about material properties of soft tissues is their heterogeneity. Experimental characterizations of material properties of soft tissues are very limited. Most experimental methods can only get homogenous stress-strain relationship. However, based on high speed camera observation, during tensile tests, stress and strain distributions on ACL are not uniform [27]. This inhomogeneous stress and strain also happen on cartilages. Usage of a single material model for all potions of ACL or cartilages will introduce additional errors to simulation results.

Regarding stability of finite element analysis, in explicit dynamical analysis, time derivative is mostly discretized as central difference method. This numerical method is conditional stabilized. Time increment must be smaller than critic time increment to get physically accepted solutions. In Abaqus simulations, it is hard to calculate this stable time increment due to geometry nonlinearity, material nonlinearity, and contact nonlinearity. Simulation is easily abort due to stability or element distortion. To fix this stability issue is mostly based on investigators' personal experiences. Finite element stability of ACL modelling is also limited by contact between cartilage and meniscus. Contact penetration is one of main reasons leading to simulation failure. Numerical simulation of contact between soft materials such as cartilages is relate to element size adjusting, contact penalty setting, contact type and contact coefficient, and so on. It is time consuming and involves tremendous trial and errors to solve the problems.

In terms of numerical error of dynamical finite element analysis, numerical error will come from both time discretization, space discretization and artificial use quadratic hexahedra element or quadratic tetrahedral element instead of linear tetrahedron element for static, contact and modal analysis [28]. Enhanced quadratic tetrahedral element is promising in terms of contact analysis and computational time [29]. However, it isn't proved that linear tetrahedron element is not good for explicit dynamic analysis. For high performance computation, numerical models are usually decomposed to many computational domains and assigned to different CPUs. Domain decomposition One advantage of finite element analysis is to handle very complex geometry structures. However, in knee injury analysis, since geometry is from MRI or CT scans, it is hard to get perfect geometry shape of ligaments, especially the connection area between ligaments and bones. A little difference in ACL shape can have a large impact on simulation results. The assembly of each knee component is also not perfect. In Abaqus, usually tie constraint is applied to connect different knee components. Two surfaces such as master surface and slave surface need be selected to connect to each other using tie constraint. Selection of nodes on the surfaces is mostly based on researchers' experiences. Different connection will largely affect accuracy of stress and strain in ACL. Another issue related to ACL geometry is that its two bundle structure cannot be modelled well. In reality, AM and PL bundles intertwine with each other. In MRI or CT scans, it is hard to tell which is which. Therefore, in the simulation, AM and PL bundles are modelled as a single piece and material properties are assigned to each of them based on stress-strain curves from experiments. This treatment will introduce additional deviation from real ACL geometry.

damping introduced to the model. It was suggested to Material properties are another concern in modelling ACL injury by using finite element analysis. Material models for soft tissues are mostly taken from other research areas such as polymer mechanics and solid mechanics. Many material models are originally built for certain applications and certain materials. But they have been extensively used to study soft tissues with parameter fitting. It is often questionable whether these material models can be used for soft tissue analysis. To model anisotropic, nonlinear, time and temperature dependent mechanical response of soft tissues is still a challenge problem. Current material models such as hyperelastic model, linear viscoelastic or combination of these two may be the most choices for many soft tissue structure simulations. However, not only do complex mathematical structures of these models limit their usage, but also these models cannot fully capture physical behaviours of soft tissues. The reason why most researchers use these material models is because they are the only choices. Therefore, new material models for soft tissues need be built based on new mathematical structures. Currently the model built by Yang [21] is a good candidate material model which break up traditional material model framework. Anisotropic and hyperelastic property was illustrated by a novel and simple mathematical framework. Nonlinear viscoelastic effect is simply modelled by time dependent Young's modulus. decreases computational time but introduces inconsistent results. That is, geometry shapes of deformed structures are not always consistent when a model is computed with different numbers of CPUs.

Another important limitation of finite element analysis of knee injury is that it is incapable to simulate physical aging and initially built-in stress and strain in ACL. We will talk this subject in the next section.

5 IV. BRIEF INTRODUCTION TO PHYSICAL AGING OF POLYMERS AND ACL

People noticed physical aging firstly when studying amorphous polymers. In glass, and under isothermal conditions, volume of polymers evolve continuously towards their equilibrium value. During this volume evolution mechanical properties of polymers also change. It is the change in mechanical properties during this volume recovery that has come to be known as physical aging. The fundamental study of physical aging of amorphous polymers and other materials is to understand non-equilibrium glassy state, the molecular scale rearrangements, and how this relates to structural relaxation, which will not be discussed here.

To best of author's knowledge, physical aging of ACL has never been studied. However, physical aging of polymers has been investigated during past several decades. Due to lacking of studies of physical aging of tissues, this section is based on former studies of physical aging of polymers. Analogy between tissues and polymers is based on that both of them have macromolecular chain structures. For bio-tissues, actin is the most dominated protein filament in eukaryotic cells. It forms the cytoskeletal rim. This actin cortex is a polymer gel that provides mechanical supports to cells, and has an important impact in cell motion [32]. Actin forms viscoelastic network. Its basic viscoelastic properties were studied based on Mackintosh chain model. But physical aging of actin was never studied before. Based on this microscopic analogy between amorphous polymers and soft tissues, therefore, it is reasonable to assume physical aging of ACL is similar to that of polymers. In amorphous polymers, physical aging happens between glass transition temperature (? peak) and second relaxation peak (? peak) [33]. Ductility will decrease with physical aging. Physical aging of amorphous polymers is explained by freevolume concept. Transport mobility of particles in a closely packed system depends on the degree of packing on the free volume. After a polymer is cooled to some temperature below glass transition temperature, the mobility will be small, but not zero. This non-zero mobility will cause free volume inside polymers to gradually decrease to an equilibrium value. This nonzero mobility largely dominates viscoelastic property shift. On the other hand, it is widely accepted that ACL is a viscoelastic material. Mathematical representation of physical aging of ACL is given as a shift of relaxation modulus, shown in Fig. 6, where ?? ?? , ?? ?? , ?? ?? are horizontal component of shifts, reference aging time, current aging time respectively. As shown in the figure, physical aging shifts relaxation modulus horizontally to the right where relaxation time will be increased. During this shift, relaxation modulus will become much stiffer than that before physical aging. During sport games, ACL movements of athletics usually happen in less than one second. Short time relaxation modulus will have a big impact on stress changes on ACL. As shown in Fig. 6, stiffening of ACL modulus due to physical aging will make ACL less ductile and have less flexible deformation before torn.

It is shown [30] that the double-logarithmic shift rate, µ is constant over wide ranges of aging time, ?? ?? for most amorphous polymers. µ is defined as

6 ??=-???????? ?? ?? ?? log ?? ??

,where ?? ?? is the shift factor. This shift factor is almost unity up to glass transition temperature and will shapely decrease to 0 when temperature is higher than glass transition temperature. But there is no experimental data reported for ACL.

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Regardless, for ACL, physical aging effects on other mechanical properties such as brittleness, Young's modulus, yielding stress, and modulus of resilience and toughness were rarely studied before. If physical aging happens, modulus of resilience and modulus toughness will become small. Therefore, less elastic strain energy can be absorbed by ACL before fracture starts. Ligaments will become brittle the same as aged polymer. Simultaneously, after several hours physical aging time, stress initially built on ACL will continue to develop.

It was generally accepted that two bundle anatomic structures of ACL, the anteromedial bundle (AMB) and the posterolateral bundle (PLB), are time dependent materials or viscoelastic materials. Their viscoelastic properties such as relaxation time and relaxation modulus can be changed with aging time. It is well known that ACL is always loaded in any anatomical position [10]. There is no anatomical position that no stress is built on ACL. Therefore, initial stress in ACL will relax to its equilibrium state with time based on viscoelastic theory. During this ACL stress relaxation, unevenly distributed stress will build on two bundles of ACL with physical aging time. This stress is not homogeneous and will lead to certain high stress concentration areas which will largely increase ACL injury risks when athletes play football, basketball, baseball, soccer or other sports.

The other point that physical aging interests us for studying ACL injury is because physical aging of ACL can be removed by high level mechanical deformation and large mechanical stress. Particularly, in sports, warming up activities such as cyclically stretching legs and bending of knees before soccer or other games will create these high level mechanical deformations and stresses. As a consequence, these activities will remove physical aging and in turns to remove initial heterogeneous built-in stress and strain on ACL.

Prevention programs such as plyometrics and strengthening are related to creating high level stress and strain in ACL in relatively slow stretch speed. Then initial stress concentration can be removed by these prevention programs. When athletes attend specific designed prevention programs for ACL injury before sports [31], physical aging effects of ACL are erased by high level stretch stresses to some degree. Actually stretching is an effective way to erasing physical aging, for example, calf stretch, groin stretch, and hamstring stretch. Muscle soreness and cramping can be reduced or eliminated by daily exercises. Activities such as dance, martial arts (aikido or karate), tai chi, or yoga are programmed stretches which are designed to reduce physical aging of bodies.

Historically, Struik [30] was the first to remark on what he referred to as erasure of physical aging due to the application of large deformations to a polymer sample. This apparent reversal of physical aging due to large mechanical or other stimulus has also been studied in structural glasses, colloidal systems, ferroelectric relaxors and spin glasses.

How physical aging can be erased by high level stress stretch is illustrated by Fig. 7. Red solid line shows relaxation modulus at high stress without physical aging. Red dash line shows relaxation modulus of the same material at low stress without physical aging. If let both of them age for the same time, red line will shift to blue line and red dash line will shift to blue dash line. It is shown that a material under high stress is less likely to get physical aging. If we look at the curves in a different way, red dash line (low stress without physical aging) is shifted horizontally to the right a big amount of distance and become blue dash line (low stress with physical aging). At this moment, if stress in this material is raised to a higher level, the blue dash line will shift back to the left and become the blue solid line where the material is under high stress with most physical aging effect erased. As shown in the figure, the material becomes less stiff and more ductile after physical aging effect is erased. In this demonstrated example, ACL is virtually modeled by using finite element model, Abaqus 2016. It is to demonstrate how stress will change in ACL when physical aging happens. Real anatomy structure of ACL is very complex in shape and orientation. Anatomically, two bundles of ACL wrap about each other and vary in length and mechanical property. AM bundle averaged 39 mm in length. Its cross section is given as 5.1_0.7 mm in sagittal width, and 4.2_0.8 mm in coronal width. PL bundle, by contrast, is averaged 20.5_2.4 mm in length, 4.4_0.8 mm in sagittal width, and 3.7_0.8 mm in coronal width [11]. Based on this anatomy knowledge, a simplified geometry model of ACL is built for simulation. AM will be modeled to be 39 mm length and PL 20.5 mm length. They are attached to each other.

Material properties of ACL are adapted from literature. Linear viscoelastic model will be used for each of two ACL bundles. Standard linear solid model is used to fit the experimental data by Castile et al [19]. As shown in Fig. 8, stress relaxation happens in a very short time. The short term and long term moduli of AM are much stiffer than those from PL. In this simulation, physical aging will be modeled by shifting the relaxation curve to the right. Approximated shift factor is assumed be one for simplicity.

To proceed, ACL is pulled away by 8 mm for both sides in two seconds and relaxed for 5 seconds. Then ACL is let to age for some time by letting mechanical properties of AM and PL shift to the right. In this calculation, we assumed that relaxation time for both AM and PL increases 10 times due to physical aging where short term and long term moduli remain unchanged. Physical aging simulation starts at the time when original relaxation simulation is completed. Initial condition of physical aging simulation is then imported from original relaxation simulation with relaxation time changed. Stress comparison of two bundles of ACL before and after physical aging is shown in Fig. 9. Compared to original ACL, higher stress is created on both AM and PL by physical aging. Higher stress area is enlarged after physical aging. Stress near to the end of ACL is also increased. The maximum stress along the ligament direction is increased from 0.626 MPa to 0.7889 MPa because of physical aging. Therefore, in this simple demonstration, physical aging increases the risk of ACL injury. In reality, geometry and mechanical properties of two bundles are much more complex. Two bundles of ACL intertwine with each other. It will be expected more stress concentration areas exist due to physical aging. In this article, numerical and theoretical analysis of ACL injury is conducted. The results from finite element analysis show that ACL will have large shear strain in 45 degree flexion angle when it is under 3g loading, comparing with 30 degree flexion angle. Tensile strain is usually smaller than shear strain in both 30 and 45 degree flexion angles. Tensile and shear strain will increase from 30% to 110% if homogenous cartilage is replaced by heterogeneous cartilage. Since heterogeneity is existed in all kinds of soft tissues, a smoothed material property distribution is suggested to use in the future numerical simulations.

Physical aging of ACL is introduced to understand ACL injury in sport activities. It is shown in the simulation that internal stress on ACL will largely increase due to physical aging. Since physical aging of ACL has never been investigated in experimental and theoretical study before, this article opens a new door to understand ACL injury during sports activities. Unlike the simplified geometry of ACL used in current computation, real ACL anatomy and geometry are more complex. AM and PL are intertwined with each other. After physical aging, stress distribution should be much more heterogeneous than the one shown here.

Since mathematical modelling and experimental tests of erasing physical aging of ACL were never attempted before, direct validation of physical aging erasure of soft tissues is not provided. However, it is widely accepted that risks of ACL injuries dramatically decrease if athletes attend specific designed prevention programs for ACL injury such as plyometrics, strengthening and other neuromuscular training exercises. In these programs, ACL will be under slowly loaded high stress state. Physical aging of ACL is believed to be erased and ACL is shifted back from brittle state to ductile state. Since ductile ACL can absorb more energy and brittle ACL is easy to fracture, detail investigations of physical aging caused ductile and brittle transition will be very interesting.

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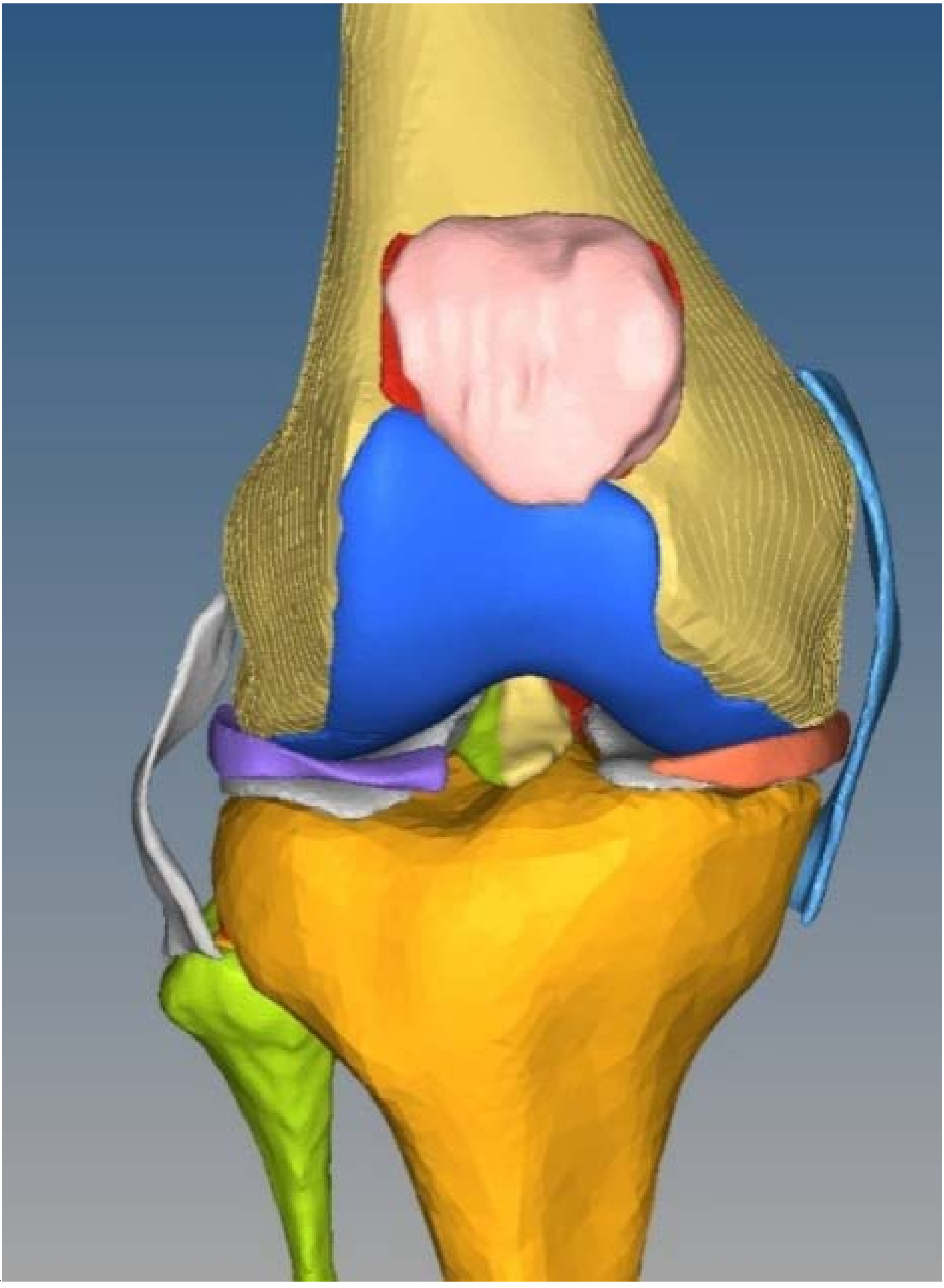
1 

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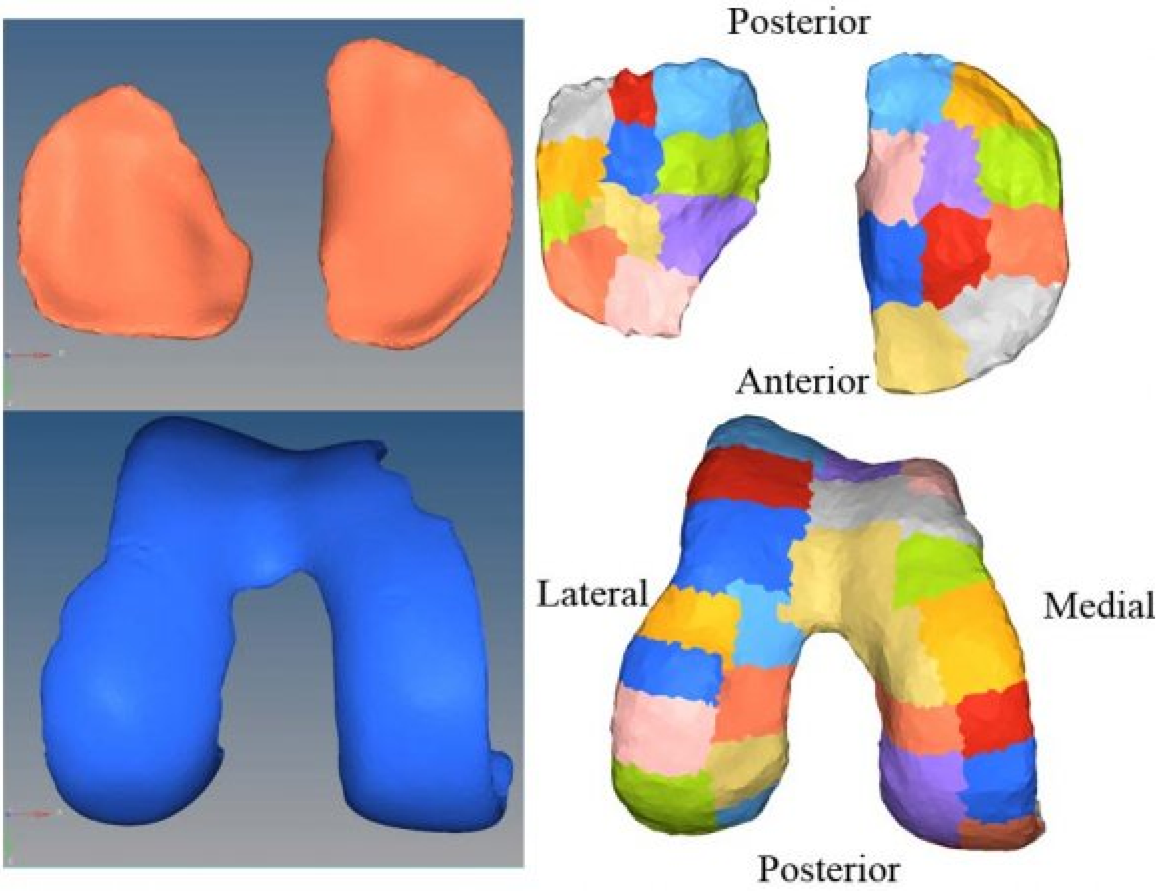

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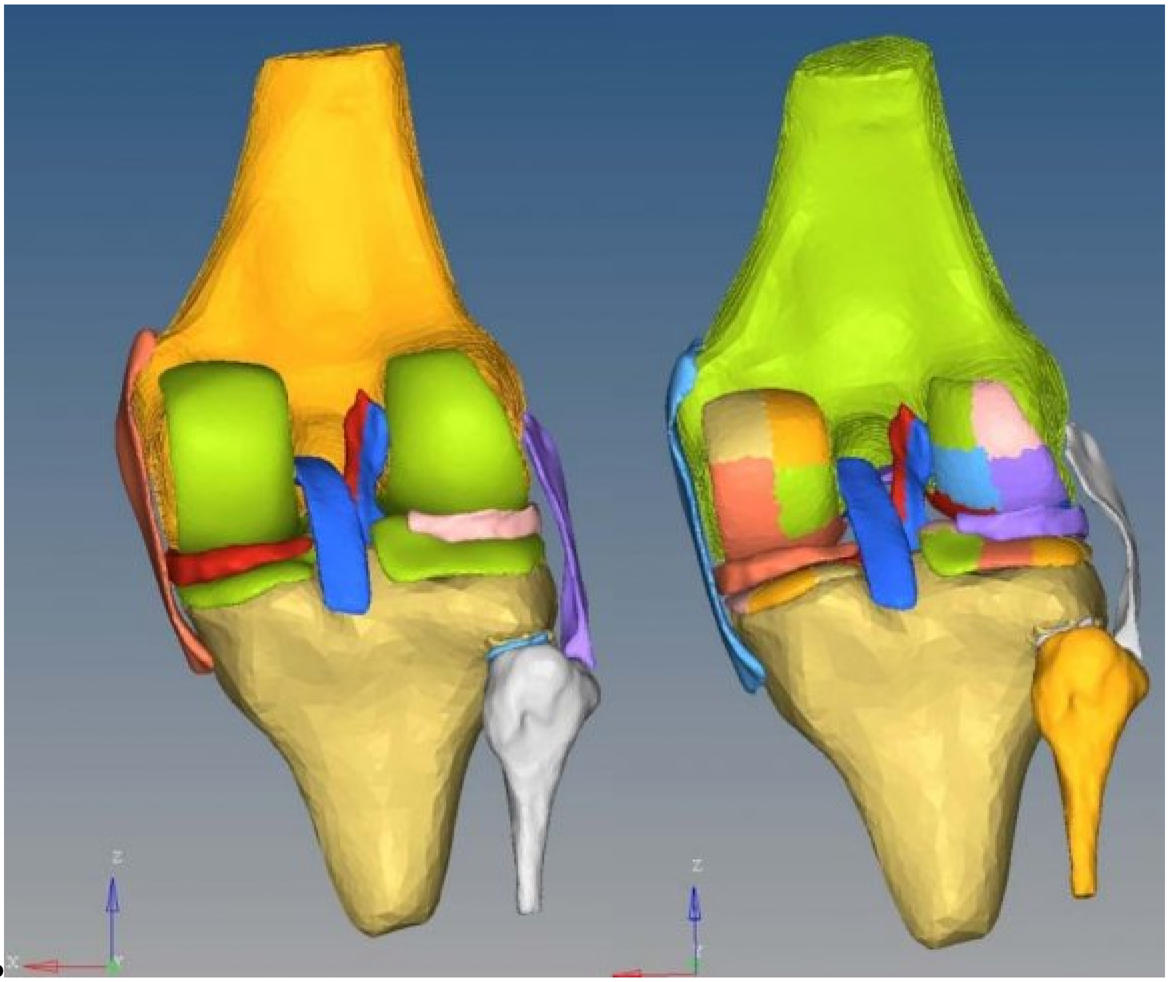
2a2b 

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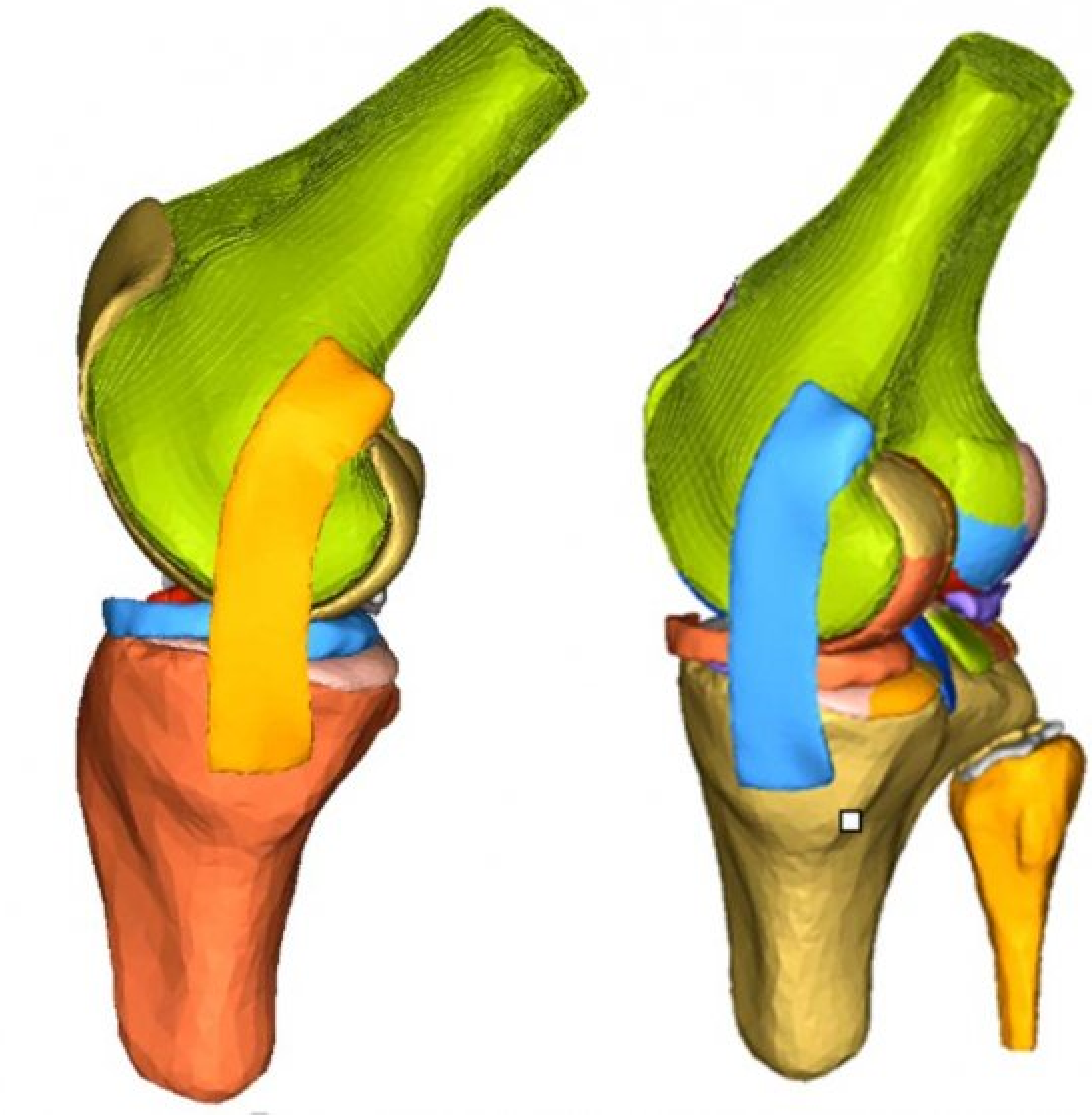

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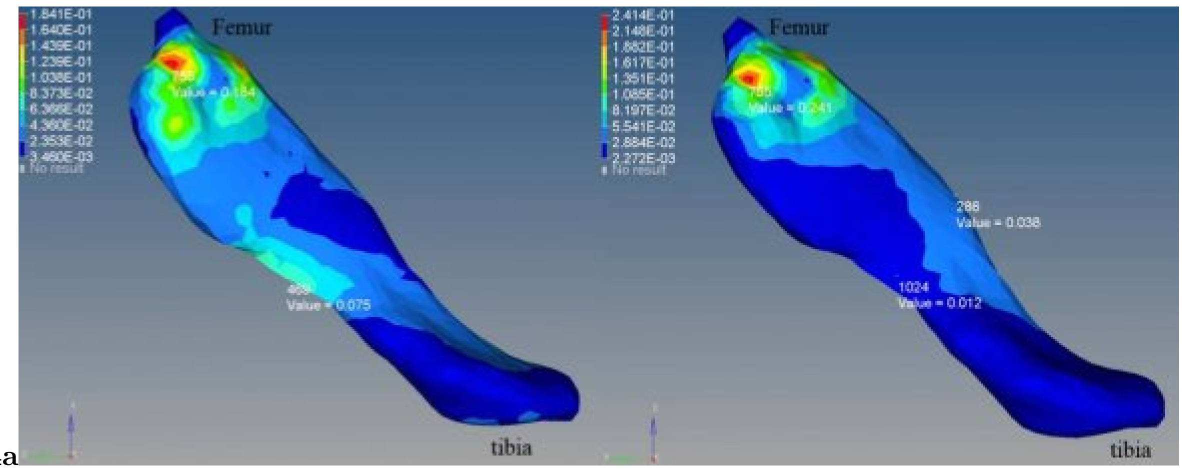

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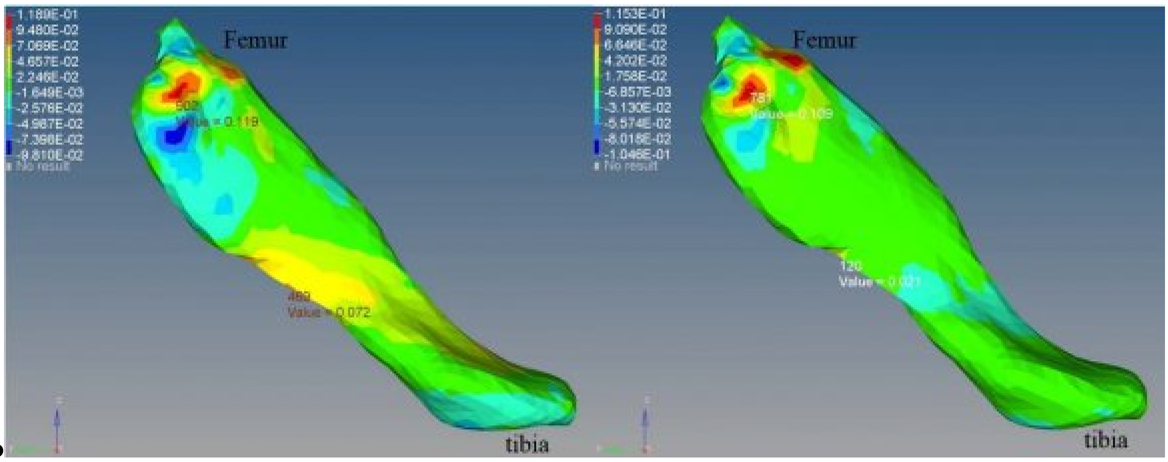
4b 

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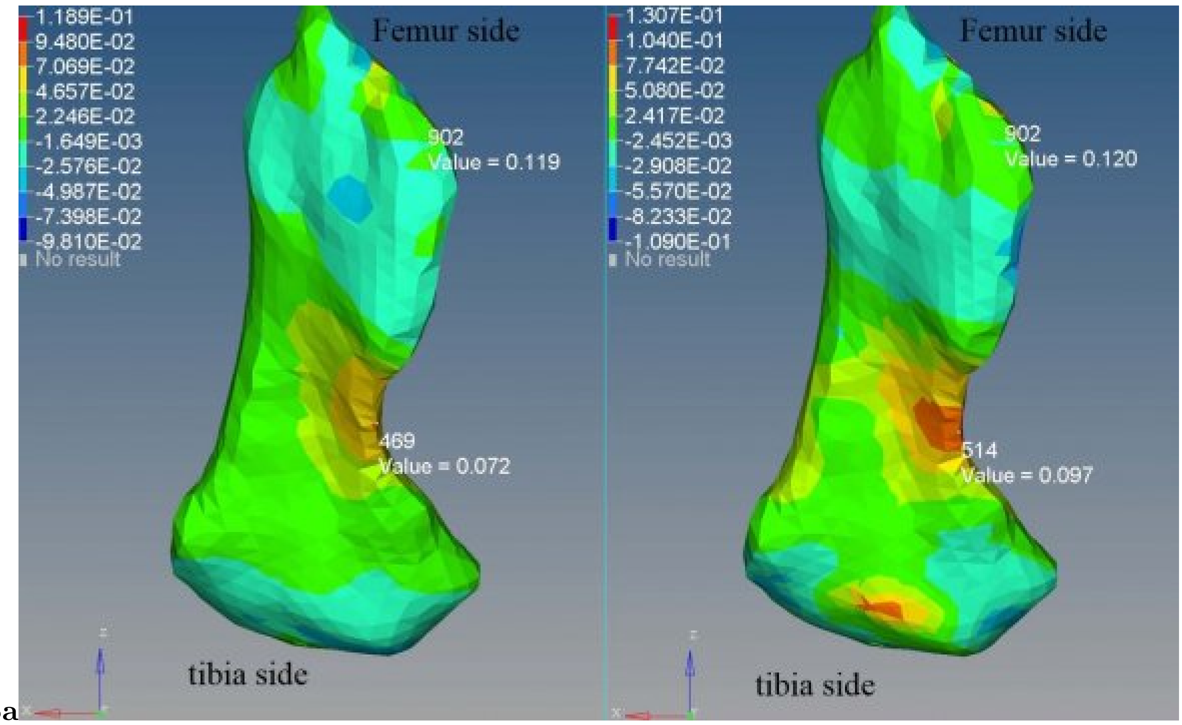
5a 

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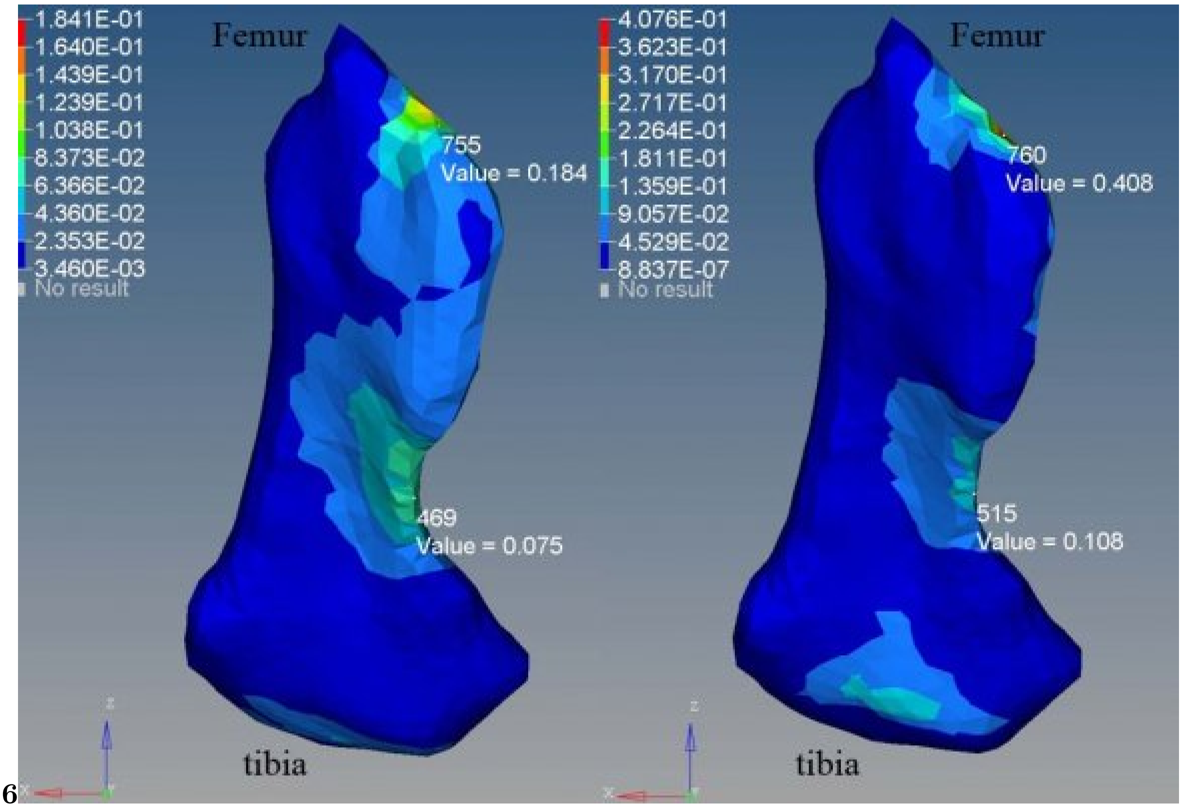

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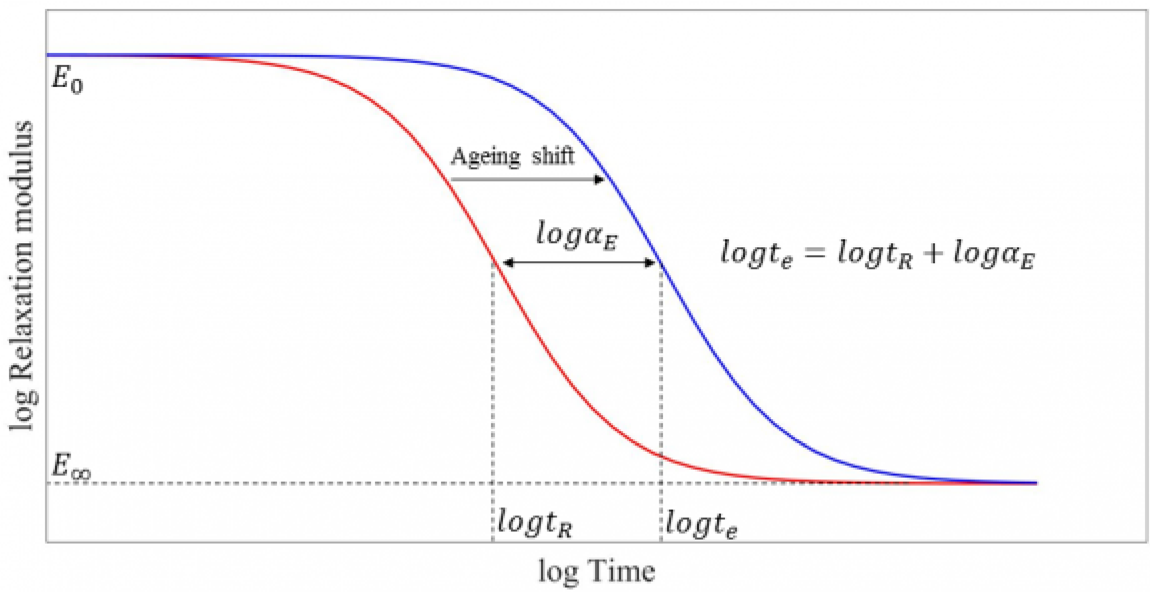

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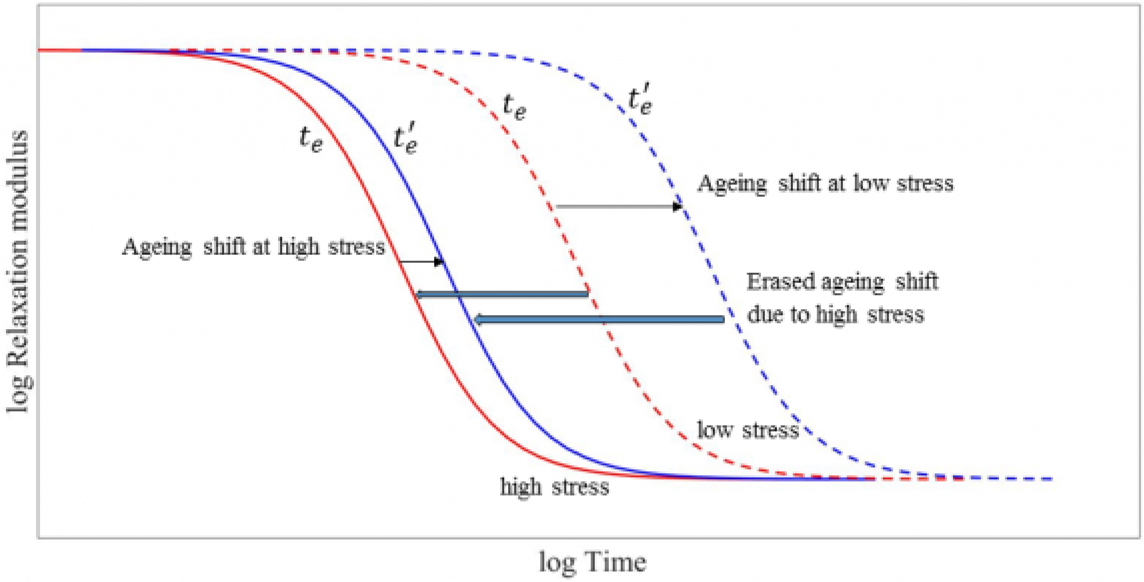

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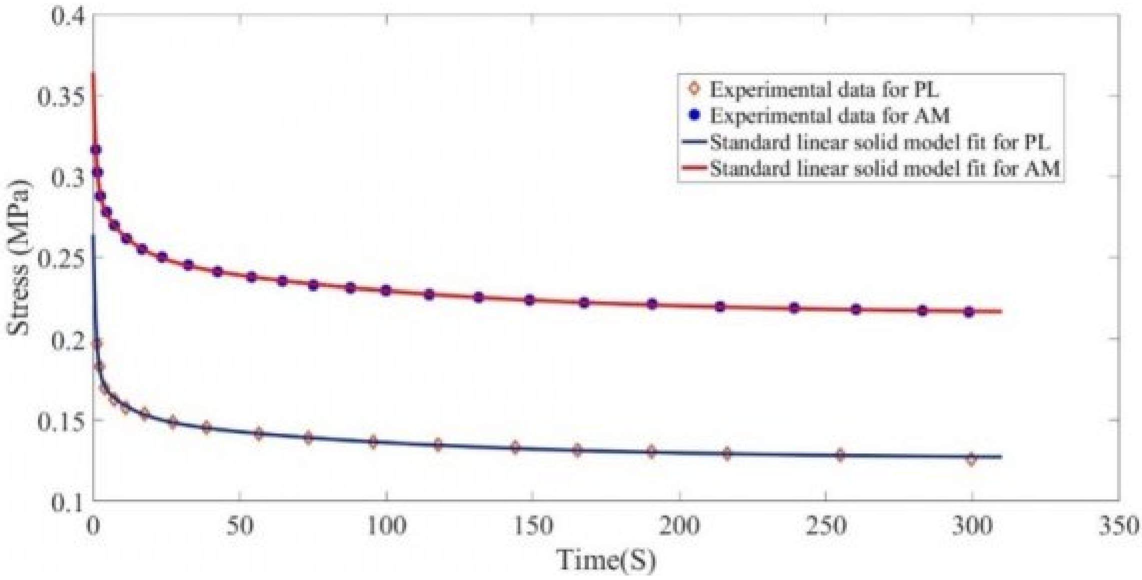

Figure 11: Figure 8 :

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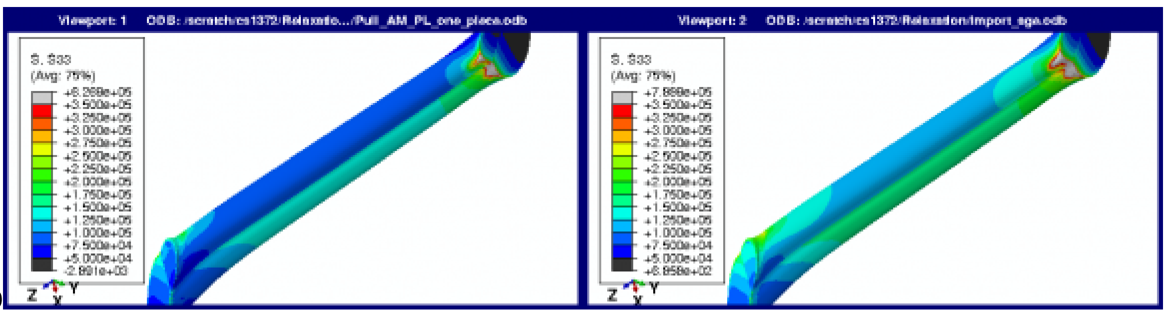

Figure 12: Figure 9 :

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[Although finite element analysis is a very promising way to study ACL injury, its usage and validation are limited by nonlinear a
Although finite element analysis is a very promising way to study ACL injury, its usage and validation are limited by nonlinear and time dependent material models, contact analysis, accuracy of geometry, (and numerical error accumulations)

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