An Investigation of Humeral Stress Fractures in Racing Thoroughbreds using a 3D Finite
Element Model in Conjunction with a Bone Remodeling Algorithm

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Ryan James Moore

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COMMITTEE MEMBERSHIP

TITLE: An Investigation of Humeral Stress Fractures in Racing Thoroughbreds using a 3D Finite Element Model in Conjunction with a Bone Remodeling Algorithm

AUTHOR: Ryan James Moore

DATE SUBMITTED: February 26, 2010

COMMITTEE CHAIR: Scott Hazelwood, Associate Professor

COMMITTEE MEMBER: Lanny Griffin, Department Chair

COMMITTEE MEMBER: David Clague, Assistant Professor

ABSTRACT

An Investigation of Humeral Stress Fractures in Racing Thoroughbreds using a 3D Finite Element Model in Conjunction with a Bone Remodeling Algorithm

Ryan James Moore

The humerus of a racing horse Thoroughbred is highly susceptible to stress fractures at a characteristic location as a result of cyclic loading. The propensity of a Thoroughbred to exhibit humeral fracture has made equines useful models in the epidemiology of stress fractures. In this study, a racing Thoroughbred humerus was simulated during training using a 3D finite element model in conjunction with a bone remodeling algorithm. Nine muscle forces and two contact forces were applied to the 3-dimensional finite element model, which contains four separate load cases representing fore-stance, mid-stance, aft-stance, and standing. Four different training programs were incorporated into the model, which represent Baseline Layup and Long Layup training programs along with two newly implemented programs for racing, which have an absence of a layup period, last a period of 24 weeks, and a race once every four weeks. Muscle and contact forces were rescaled for all load cases to simulate dirt, turf, and synthetic track surfaces. Bone porosity, damage, and BMU activation frequency were examined at the stress fracture site and compared with a control location called the caudal diaphysis. It was found that race programs exhibited similar remodeling patterns between each other. Damage at the stress fracture site and caudal diaphysis was reduced during all training programs for the turf and synthetic track surfaces with respect to the dirt track surface. Key findings also included changes in bone remodeling at the stress fracture site and caudal diaphysis as a result of turf and synthetic track surfaces. This model can serve as a framework for further studies in human or equine athletes who are susceptible to stress fractures.

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I. Introduction

Thoroughbred Racehorse Injuries

Thoroughbred race horses, which are between two and three years of age, commonly experience humeral stress fractures at a characteristic location making them a useful model in this study. Thoroughbred racehorses can experience multiples injuries during intense periods of training, such as bucked shins or humeral stress fractures, where 70% of young Thoroughbred racehorses develop bucked shins as a result of repetitive loading injury in their third metacarpal bone (MCIII) (Nunamaker, 2002b). When a horse has a musculoskeletal injury, it is important to have a period of layup or rest to allow the damaged area to heal. If the horse becomes injured to the point where they are no longer able to recover, then they are highly susceptible to fracture, which can cost the owner a substantial investment, where the median sale price of a Thoroughbred between the age of 1 and 2 years is \$40,000 (Burr and Milgrom, 2001). The frequency of equine injury coupled with a high-priced investment makes it practical to implement and investigate new training programs, which not only prevent equine musculoskeletal injury, but allows equine athletes to stay physically competitive.

Thoroughbred racehorses between 2 to 3 years of age are more at risk to musculoskeletal injury compared to older Thoroughbreds since younger Thoroughbred's bones are still maturing, are more porous, and have yet to adapt to the racing environment (Boston and Nunamaker, 2000). Mature horses have a large bone radius when compared to younger horses, which allows for lower stress at similar forces due to a greater cross sectional area of bone. Also, it is more likely stress fracture can occur if the horse is initially inactive, but begins training at a high-speed exercise, where the horse exhibits

stiff muscles at a high frequency (Hernandez et al., 2001). Previous studies have shown that incorporating a walking exercise into the horse's training schedule reduces the chance of having bucked shins (Mattila et al., 2007). In the current study, four separate training programs were investigated, which include layup programs Baseline Layup and Long Layup, and two new race programs.

Along with training regimes, additional environmental influences can play a role in the development of equine musculoskeletal injury. Track surface material type, such as dirt, turf, and synthetic, contribute to different effects on ground reaction forces and hoof accelerations during trot and canter (Setterbo et al., 2007). Compared to dirt, turf proved to be associated with lower risk for injury in one study (Mohammed et al., 1991), while another showed the complete opposite (Hernandez et al., 2001). There are frequent contradictions and inconsistencies in previous literature on correlation between risks of equine musculoskeletal injury and track surface type. Although it has been statistically proven that turf, synthetic, and dirt track surfaces contribute to different hoof ground reaction forces and accelerations during gallop (Setterbo et al., 2007), many believe the softer tracks are more advantageous and physically safer for equine races since dirt surfaces have been replaced with synthetic materials at several different race courses over the last 3 years (Setterbo et al., 2007). In the current study, we employ previously calculated hoof ground reaction forces from dirt, turf, and synthetic surfaces (Setterbo et al., 2007) to investigate their effects in a computational model on bone remodeling in a Thoroughbred racing humerus.

The Characteristics, Function, and Biology of Bone

Bone Composition

Bone is composed of 25% water, 32% organic matrix, and 43% apatite mineral, which includes collagen, hydroxyapatite, and small portions of proteoglycans and noncollagenous proteins (Martin et al., 1998). Collagen is a structural protein organized to create strong fibers, which provide high tensile strength and flexibility to bone, where the most common type of collagen found in bone is type I. Hydroxyapatite $[\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2]$ crystals account for the majority of bone's dry weight and are key for providing bone with compressive strength and rigidity. Proteoglycans participate in modulating collagen fibril assembly (Martin et al., 1998). While bone is composed of all these components, its material properties can vary depending on apparent density and porosity (Nunamaker, 2002a).

Bone Function

The primary functions of bone include providing a foundation for other organs to adhere to, protecting organs from exterior trauma, and transmitting forces from one area of the body to a desired location. Bone is designed to bear the weight of loads with minimal deformation and has a high strength to weight ratio. Bone is known to model based on the load it receives by changing its morphology to reduce stresses (Boston and Nunamaker, 2000). Bone changes its geometry to efficiently withstand future loads, which is known as Wolff's Law where bone becomes aligned in principal stress directions. Marrow is contained within the medullary canal of bone and is a tissue composed of blood vessels, nerves, and other types of cells. Its main responsibility is

generating new blood cells (Martin et al., 1998). Bone vasculature is maintained through the medullary canal and periosteal blood supply. Circulation in bone is centripetal, where the afferent blood supply consists of arteries and arterioles and the efferent blood supply is composed of veins and veinules (Nunamaker, 2002a). While bone is perceived by most as a stagnant inert structure, it is dynamic and continuously changing its composition and microscopic architecture due to mechanical stimuli and other environmental factors.

Bone Tissue Types

There are two types of bone tissue called trabecular and cortical bone. Cortical bone or sometimes referred to as compact bone, is dense, has a porosity (void volume/total volume) of 5%-10%, and is mainly found in shafts of long bones and forms a cortex around vertebral bodies and other spongy bones (Martin et al., 1998). Within cortical bone, there are porous spaces composed of Haversian canals, Volkmann's canals, and resorption cavities. Haversian canals are aligned along the length of the bone and contain capillaries, nerves, and are about 50 μm in diameter (Martin et al., 1998). Volkmann's canals also contain blood vessels along with nerves and are transverse canals connecting Haversian canals to each other and to the outside surface of bones (Martin et al., 1998). Resorption cavities are about 200 μm in diameter and are temporary pits created by bone cells which remove bone known as osteoclasts. Figure 1 shows typical features of a long bone.

Figure 1. Sketch of typical long bone features (Martin et al., 1998).

Trabecular or cancellous or spongy bone contains interconnected pores filled with bone marrow and has a porosity of 75%-95%. Trabecular bone is found in the vertebrae, flat bones, and ends of long bones. The bone matrix of trabecular bone normally resides on the endosteum of long bone and forms struts called trabeculae, which are approximately 200 μm thick. Trabeculae can either be randomly organized or form orthogonal arrays. Both cortical and trabecular bone contain two types of bone tissue known as lamellar bone and woven bone (Martin et al., 1998). Lamellar bone is highly organized and slowly formed with an anisotropic matrix, which means its material properties change depending on direction. The anisotropic matrix consists of highly organized parallel layers or lamellae of mineral crystals and collagen fibers. Woven bone is formed quickly and poorly organized, where mineral crystals and collagen fibers are

randomly arranged (Martin et al., 1998). Woven bone is primarily used to temporarily bridge stress fractures to alleviate stress.

Cortical bone tissue contains two categories; primary and secondary bone.

Primary bone lays down new bone on an existing surface during growth, such as the periosteal surface (Martin et al., 1998). Within primary bone, there are two general types called circumferential lamellar bone and plexiform bone. Circumferential lamellar bone contains a matrix of lamellae parallel to the surface of the bone. When blood vessels become incorporated into new periosteal bone, they become surrounded by circular lamellae, which form primary osteons with a primary Haversian canal at the center (Martin et al., 1998). Plexiform bone has a high rate of formation and is a mixture of woven bone (trabeculae) and lamellar bone. Within plexiform bone are rectilinear vascular spaces, which give an appearance of a “brick wall.” Plexiform bone is found in larger animals such as cows and horses. It may also aid in fatigue resistance for racehorses (Stover et al., 1992). Secondary bone results in the removal of existing bone and replacement with healthy, lamellar bone by a process known as remodeling in which teams of bone cells work together to replace old retired bone with new healthy bone. Secondary bone contains cylindrical systems of Haversian canals and secondary osteons (formed through remodeling). For human adults, compact bone is entirely composed of secondary bone and trabecular bone is also mostly secondary bone, but trabecular bone rarely produces osteons since they cannot fit inside individual trabeculae (Martin et al., 1998).

Bone Cells

Bone is a dynamic structure in which remodeling takes place to remove old bone and replace it with new bone, which is initiated by: microdamage, chemical factors, fracture, bone age, and mechanical environment. There are a total of four different types of bone cells known as osteoblasts, osteoclasts, osteocytes, and bone lining cells. The bone cells responsible for removing and replacing bone are known as osteoclasts and osteoblasts, respectively. Osteoclasts tunnel through bone and remove it, while osteoblasts follow, laying down new healthy bone. Collectively, these teams of osteoblasts and osteoclasts are known as basic multicellular units or BMUs. Osteocytes are former osteoblasts which became embedded in the bone matrix by being engulfed by faster, more active osteoblasts. Osteocytes reside in cavities known as lacunae, where they communicate with osteoblasts with processes passing through tunnels called canaliculi (Martin et al., 1998). Bone lining cells are also quiescent osteoblasts, which reside flattened on the surfaces of bone and aid in communication with osteocytes by using gap-junctioned processes (Martin et al., 1998). Bone lining cells contain receptors for parathyroid hormone, estrogen, and other chemical messengers. Both osteocytes and bone lining cells are responsible for the transfer of mineral into and out of bone. The combination of all four bone cells allows the microstructure of bone to repair, communicate with other cells, and transfer necessary nutrients.

Osteoblasts differentiate from mesenchymal stem cells, which derive from the stromal tissue of bone marrow. Complete differentiation of mesenchymal stem cells can take 2-3 days (Martin et al., 1998) and is initiated through mechanical stress to bone tissue. Osteoblasts produce osteoid, the organic portion of the bone matrix, which

contains collagen and noncollagenous proteins, proteoglycans, and water (Martin et al., 1998). Osteoid is laid down by osteoblasts at an apposition rate of approximately 1 $\mu\text{m}/\text{day}$. Unlike osteoblasts which are mononuclear, osteoclasts are multinuclear, produced by the fusion of monocytes originating from the hemopoietic portion of bone marrow (Martin et al., 1998). When osteoclasts break down and resorb bone, it occurs along the ruffled border of the cell, which is sealed to the bone surface. The resorption rate of osteoclasts is approximately 10 $\mu\text{m}/\text{day}$ and is accomplished through the production of acids and enzymes, which demineralize adjacent bone and dissolve collagen, respectively.

Modeling and Remodeling

This study is aimed at the process of remodeling in which teams of osteoblasts and osteoclasts work together to remove and replace bone at a specific location. Modeling is where bone is removed in certain areas (osteoclastic activity) and added to others (osteoblastic activity), so osteoblasts and osteoclasts are working independently. When compared to remodeling, modeling is greatly reduced after skeletal maturity. Modeling also results in the change of bone size and shape, but remodeling normally does not affect size or shape. Both of these features of bone are necessary for the proper development and renovation of the skeleton.

Modeling allows the geometry and size of the skeleton to develop correctly and can be resorptive or formative while a child is developing. It allows bone to customize itself based on loads it receives, which give it optimal material and mechanical properties for future similar mechanical stimuli. Remodeling is a reparative process due to

microdamage in which osteoclasts work to remove old bone and osteoblasts follow replacing it with new bone. It has also been hypothesized that remodeling mechanically “fine tunes” the skeleton to increase mechanical efficiency (Martin et al., 1998). BMUs (Basic Multicellular Units) are a collective team of several hundred osteoblasts and approximately 10 osteoclasts. During a life of a BMU it undergoes three main stages: activation, resorption, and formation. Activation is initiated through a chemical or mechanical signal causing osteoclasts to form. Osteoclasts resorb a volume of bone in the form of a trench for bone surfaces and create a tunnel (200 μm in diameter) in compact bone, moving at approximately 40 $\mu\text{m}/\text{day}$. When osteoclastic activity begins to slow, osteoblasts start differentiating from mesenchymal stem cells and initiate bone tissue formation over the removed tissue (Martin et al., 1998).

Secondary osteons are created by BMU tunnels in cortical bone. BMUs are responsible for replacing approximately 5% of compact bone each year in adult humans. In trabecular bone, BMUs work at the surface by digging and refilling trenches. The average human adult BMU replaces approximately 25% of trabecular bone per year (Martin et al., 1998). Compared to human bone, equine bone has more bone volume, which indicates a higher probability of damage, so the frequency of remodeling is greater in equine than humans.

Figure 2. Illustration of the creation of a secondary osteon through the resorption of osteoclasts followed by osteoblasts, which are forming new bone to establish a newly remodeled haversian system (Nunamaker, 2002a).

There are six separate phases to bone remodeling: activation, resorption, reversal, formation, mineralization, and quiescence. During activation (3 days), osteoclasts are created and recruited to the site of interest. For the resorption period, osteoclasts begin to break down and resorb bone (Martin et al., 1998). During the reversal period, there is a transition from osteoclast to osteoblast activity, which takes several days and a visible line (cement line) is formed where the osteoclasts cease activity. The porosity of bone temporarily increases after the resorption period, thus decreasing the elastic modulus. The resorption and reversal periods combined take about 30 days. After the reversal period, the formation period begins (this phase in adult humans lasts about 3 months) and osteoblasts fill in new bone, starting at the periphery of the tunnel created by the osteoclasts (Martin et al., 1998) (Figure 2). Osteoblasts do not completely refill the tunnel and leave a space for a vascular loop to support the metabolic needs of bone cells,

which create secondary osteons (Figure 3). After formation, the bone site is mineralized and goes into quiescence, where there is a period of inactivity.

Figure 3. Microradiograph of remodeling in horse bone within the cortex. The larger black areas represent resorption cavities, while the grey oval shaped areas represent new-less calcified bone making up secondary osteons (Nunamaker, 2002a).

Material and Mechanical Properties

Bone is a viscoelastic material, meaning it is both viscous (resistant to flow) and elastic (can withstand permanent deformation). Collagen gives bone its elastic properties in tension, while hydroxyapatite supplies a high compressive elastic modulus. The elastic modulus for equine cortical bone is approximately 18-20 GPa (Nunamaker, 2002a), which is similar to human cortical bone with an average of 17.9 GPa (Reilly et al., 1974) in the longitudinal direction. It is also anisotropic, meaning its material properties depend on orientation. Bone is strongest in compression and weakest in tension, but its strength depends on the direction of loading. Longitudinally bone is stronger compared to transverse loading. Adult humans have an average ultimate compressive strength in

the longitudinal direction of 205 MPa and 131 Mpa in the transverse direction, while the ultimate tensile strength is 135 MPa in the longitudinal direction and 53 MPa in the transverse direction (Table 1).

Cortical bone mechanical properties of the cortex are the result of the following osteon activities: replacing highly mineralized bone matrix with less calcified material, increasing cortical porosity, altering collagen fiber orientation, and incorporating cement line interfaces, which have unique mechanical properties. Collagen fiber orientation can affect bone strength, stiffness, density, porosity, and mineralization (Martin et al., 1998). Porosity can affect bone density, stiffness, and be an indicator for a potential stress fracture. Porosity, mineralization, density, histological architecture, collagen fiber organization, fatigue damage, and rate of deformation are all factors affecting the material properties of cortical bone (Martin et al., 1998).

Table 1. Anisotropic data from a bovine and human bone (Reilly et al., 1974; Reilly and Burstein, 1975).

Elastic Modulus, Gpa	Tension		Compression	
	Longitudinal	Transverse	Longitudinal	Transverse
Human	17.9	10.1	18.2	11.7
Haversian				
Bovine	23.1	10.4	22.3	10.1
Haversian				
Primary	26.5	11	N/A	N/A

Ultimate Stress, Mpa	Tension		Compression	
	Longitudinal	Transverse	Longitudinal	Transverse
Human	135	53	205	131
Haversian				
Bovine	150	49	272	146
Haversian				
Primary	167	55		

Stress Fractures

Stress fractures can range from small microcracks to complete fractures through the entire bone and are induced by repetitive mechanical loading of the bone. They are commonly seen in military recruits, long distance runners, and race horses. Stress fractures and their development can be predicted by looking at the damage, BMU activation frequency, porosity, strain, modulus and stress at a specific area on a Thoroughbred's humerus. Cyclic stresses in bone can lead to microdamage, which initiates remodeling, where osteoclasts began to remove damaged bone, which temporarily increases porosity, thus reducing the elastic modulus. This is the primary method in which bone addresses stress fractures, but the strain of the bone increases and becomes susceptible to fracture (O'Sullivan and Lumsden, 2003). For more unstable stress fractures, osteoblasts will bridge the stress fracture gap and lay down woven bone forming a callus. Calluses form to adapt to stress fractures allowing for increased loading and to enlarge bone area, which decreases strain (Johnson et al., 1994). The risk for stress fracture increases when an exercise suddenly becomes more rigorous, such as runners increasing mileage or running frequency. This same theory can be applied to a racing horse, so within this study's exercise simulation; distance should be increased over a longer period of time.

In a previous study, humeral stress fractures were seen more often in 3-year-old Thoroughbreds, when compared to tibial stress fractures, which were predominately seen in 2-year-old Thoroughbreds. Additionally, humeral stress fractures were more commonly seen in extensively raced Thoroughbreds, while lightly raced horses were more likely to experience a tibial fracture (O'Sullivan and Lumsden, 2003). Stress

fractures in young human athletes between 19 and 25 years of age represent approximately 70% of all reported stress fractures in runners (Hazelwood and Castillo, 2007). In the current study, there are two sites of interest on the humerus: a region known to be consistently associated with stress fractures known as the caudoproximal stress fracture site and a control location for comparison called the caudal diaphysis (Figure 4). The stress fracture site and caudal diaphysis were chosen since they provide sufficient mechanical testing data from a previous study (Entwistle et al., 2009). This study explored the differences between materials properties in a healthy bone and callused bone (Entwistle et al., 2009). Stress fractures are ideal to study due to their predictable locations and frequency among athletes.

Figure 4. (a) Lateromedial radiographic projection of one half of the Thoroughbred right humerus cut along the sagittal plane. The red box indicates the location of the stress fracture site at the caudoproximal region. The green box displays the caudal aspect of the diaphysis. (b) The craniocaudal radiographic projection of the complete right humerus (Ferullo, 2007).

The Entwistle et al. (2009) study conducted mechanical testing on equine humeri, where the objective was to compare cortical bone material properties between areas with,

and without, stress fracture. Cortical beams and cores were harvested from the stress fracture site and a site distant from the affected region (caudal diaphysis), respectively. The humeri were classified as having an absence of stress fracture (stage 0) or having a presence of stress fracture with a callus maturity from stages 1-3 (Entwistle et al., 2009). These samples were then tested to failure at an apparent strain rate of 0.01sec^{-1} obtaining material properties from each section. It was found that cortical beam specimens underwent a reduction in material properties, such as a decrease in modulus as the callus staged progressed. At the caudal diaphysis, material properties were initially reduced, but then recovered to control levels as the stage of the callus progressed (Entwistle et al., 2009).

The affected cortical beam specimens produced median longitudinal moduli of 15.5 GPa, 17.3 GPa, and 9.3 GPa for callus stages 1, 2, and 3, respectively, while the median for the control specimens was 17.4 GPa. The median elastic moduli taken from the cores at the caudal diaphysis were 12.8 GPa, 14.0 GPa, and 14.8 GPa for callus stages 1, 2, and 3, respectively, while the median modulus was 14.7 GPa for the unaffected specimens (Entwistle et al., 2009). Two different types of specimens (beams and cores) were used for mechanical testing at the stress fracture site and caudal diaphysis, which explains the difference in modulus between the two. Additionally, 18 Thoroughbred racehorses of ages 2-4 were used (Entwistle et al., 2009), which give a range of moduli since it is dependent on equine age and training stage of the horse. In the current model, at the beginning of training for each program, the modulus for the stress fracture site and caudal diaphysis is 11 GPa and 16 GPa, respectively (Ferullo, 2007).

Finite Element Analysis

Finite element modeling is a useful tool for analyzing stresses and strains in bone at the macro scale. It is a numerical system where stresses, strains, and other mechanical properties are computationally solved within a solid volume, such as a humerus. The volume of the model is divided up into elements (mesh), which have a set of mechanical and materials properties (Martin et al., 1998). Specific material properties such as density, porosity, and elastic modulus can be defined within each element. Boundary conditions, which either supply loads or provide restrictions to the model, can also be applied. The number of elements used will determine the resolution of the mesh and the entire model; with an increasing number of elements making the model more accurate. After boundary conditions, material properties, orientation, and geometry are defined, computational modeling is used to solve a system of equations, which can determine the model's deformation and stress contours along with other results.

Finite element modeling and simulations have become an alternative to traditional laboratory tests with the increasingly availability of powerful computers along with more affordable prices (Ross, 2005). Traditional laboratory tests are time consuming, costly, and require more material. It is also difficult to use experimental devices, such as strain gauges, within a complex geometry of bone, like the temporomandibular joint (Figure 5) in humans (Beek et al., 2000). Finite element modeling has the ability to mesh and observe strain contours of bone containing highly irregular geometries. Finite element modeling is an important tool in orthopedics due to irregular bone shapes and varying material properties. It is used in the current study to examine equine humeri by

investigating stress fractures produced from: racing, impact from various track surfaces, and training regimes.

Figure 5. Three-dimensional finite element lateral display of the right human temporomandibular joint (Beek et al., 1999).

To model the mechanical behavior of a Thoroughbred racing horse humerus, this study used ABAQUS (SIMULIA, Providence, RI), a finite element program where loading and boundary conditions can be applied to a mesh based on the bone's geometry. For the 3D finite element model of the horse humerus, the loading conditions are represented by nine muscle forces and two contact forces, while the distal end of the humerus is fixed as a boundary condition. The muscle force and contact force values placed in FEA model were found from a previous study, which used a mathematical model and a physical testing system consisting of a test fixture, weights, and strain gauges (Pollock et al., 2008a; Pollock et al., 2008b). Also within the finite element model, there are four different loading conditions representing standing, fore-stance, mid-stance, and aft-stance. These four loading cases represent one complete day of training as

four steps. There are 1400 steps in each simulation for the Baseline Layup and Long Layup training programs, which yields 350 days of training per simulation. The race programs have a total of 672 steps, which is 168 days of training and racing. At any of these steps the material behavior of the humerus can be observed by looking at stress contours, strain, and deformation within ABAQUS (SIMULIA, Providence, RI), but in this study the state variables of interest are bone porosity, BMU activation frequency, and damage, which are tracked with a remodeling algorithm at the two humeral sites of interest.

Bone Remodeling Algorithm

The remodeling algorithm developed by Hazelwood et al. (2001) is employed into the finite element model and simulates changes in bone's elastic modulus and porosity resulting from damage and the mechanical environment. This algorithm was initially designed to be incorporated into a finite element model representing a small bone volume or an entire bone (Ferullo, 2007). This simulation uses 8 state variables, which are updated throughout each step of the model and include: elastic modulus, porosity, number of resorbing and refilling BMUs, BMU activation frequency, normalized specific area, damage, strain, and mechanical stimulus. Tables 6 and 7 display state variables and constants with their respective values. At every step during the training regime, bone elastic modulus and porosity are updated as a result of damage formation, damage removal, or disuse. The schematic below (Figure 6) displays the representation of the bone remodeling algorithm by Hazelwood et al. (2001) and features two separate feedback loops by which remodeling rate affects damage. The right loop results in

damage removal, while the left loop increasing the rate of remodeling, which increases porosity, strain, and damage formation (Hazelwood et al., 2001).

Figure 6. Schematic of bone remodeling algorithm. This system contains two feedback loops, where the right side results in damage removal and the left side results in increased porosity, strain, and damage formation (Hazelwood et al., 2001).

The bone remodeling algorithm was incorporated into the finite element model in the current study to investigate stress fractures and bone material property changes in the equine humerus as a result of training regimes, track surface properties, muscle and joint contact loads. The current algorithm was implemented in FORTRAN and designates the activities to which the FEA humerus will undergo by assigning the duration of slow and fast gallops to each step of the gait cycle. The program where the bone remodeling algorithm resides also determines the distance the horse will run depending on the day of the training program. A separate case in the simulation assigns the days a horse will perform a layup, where only the standing case is applied, so the slow and fast gallop variables are set to zero. Every Thoroughbred training program in the current study used the bone remodeling algorithm to simulate the remodeling response.

Hazelwood et al.'s (2001) bone remodeling algorithm was used in this study due to its successful contribution to multiple previous studies with different applications in which it is employed with an FEA model to simulate changes in bone material properties. One study observed the effects of marathon training on bone density, remodeling, and microdamage in the human femur. Similar to horse training programs in the current study, different marathon training programs were used for comparison to observe differences in bone damage, density, and activation frequency (Hazelwood and Castillo, 2007). Another study used the same bone remodeling algorithm to simulate and compare stress shielding effects in the femoral head due to resurfacing of the hip and a total hip arthroplasty (Deuel, 2007). This study was able to determine bone loss due to stress shielding from different modes of hip surgery. Hazelwood et al.'s (2001) bone remodeling algorithm will continuously be updated to create additional simulations and will be incorporated into further bone remodeling computational studies.

Layup Training Programs and Race Programs

Ferullo's (2007) study had four different programs lasting 350 days, where each had different periods of layup representing healing periods following injury to the horse. The objective of that study was to compare the mechanism of bone remodeling between the four layup programs. The layup training programs included: Baseline Layup, Long Layup, Short Layup, and Split-Fast Gallop training programs. Layup for the Baseline Layup, Long Layup, and Short Layup training programs lasted 8 weeks, 16 weeks, and 5 weeks respectively. The Split-Fast Gallop program had a layup period exactly the same as the Baseline Layup program, but had the distance of fast gallop split in half during the

week, where the horse ran a fast gallop twice a week at half distance. The Baseline Layup and Long Layup programs were used from Ferullo's (2007) study to validate simulation files in the current study.

It was found from Ferullo's (2007) study that training programs with a period of longer layup displayed lower values for damage when compared to programs with shorter periods of layup. Bone is given time to remodel and remove damage during longer periods of layup. Training programs with shorter periods of layup apply more loading cycles and contribute to increased microdamage at the stress fracture site. However, training programs with longer periods of layup increase the porosity of bone making it more susceptible to fracture at the return of training. Training programs without a period of layup would have a tighter control on the porosity and elastic modulus at the stress fracture site and caudal diaphysis within the humerus.

Two new programs, known as the race programs, were implemented in the current investigation to study bone remodeling during a training regime without a period of layup. These programs did not incorporate a period of layup, which would not allow bone cells to increase porosity. When porosity in the humerus is increased, the bone's susceptibility to stress fracture increases. It is common for stress fractures to occur during the sudden onset of increased microdamage, which is commonly seen during training within 3 weeks after layup (Carrier et al., 1998). Even though microdamage would accumulate for a program without a period of layup, the higher elastic modulus coupled with workout loads might be able to inhibit a catastrophic stress fracture when compared to a very porous humerus experiencing similar workout loads.

Goals and Hypotheses

The three main goals in the current study included: implementing and validating a working Thoroughbred humerus model in ABAQUS (SIMULIA, Providence, RI) using previously created ABAQUS (SIMULIA, Providence, RI) input files and FORTRAN script files by Ferullo (2007), creating two entirely new training programs (race programs) and comparing them between each other, and lastly incorporating turf and synthetic track surface properties into all training programs by scaling down muscle and contact forces. Data was gathered at two sites of interest (Figure 4) on the Thoroughbred racing humerus: the stress fracture and caudal diaphysis sites. The training programs observed underwent different periods of layup, initial training, race programs, and length of exercise.

This study encompassed three hypotheses. The first hypothesis was the FEA model of the current study will produce the same results for the Baseline Layup and Long Layup training programs as those seen in the previously validated model of Ferullo (2007). The second hypothesis was Race Program 1 will accumulate more microdamage when compared to Race Program 2 since Race Program 1 immediately undergoes a race on the first week of race training. Race Program 2 has a bye during the same week with a day of slow gallop instead of a race, which produces less damage allowing the horse to recover throughout training in comparison to Race Program 1. The third hypothesis was turf and synthetic track surfaces will contribute smaller loads when compared to dirt track surfaces causing less bone damage for all training programs at the stress fracture site and caudal diaphysis. Less bone damage implies a lower damage removal rate, lower BMU activation frequency, and decrease in porosity with an increase in elastic modulus.

Additionally, smaller loads can lead to disuse, which increases BMU activation frequency and increases porosity by removing unnecessary bone (bone remodels in response to the load it receives).

II. Methods

Development of Finite Element Model

A CT scan of a healthy humerus from a 6-year-old Thoroughbred was placed into Mimics (Materialise, Ann Arbor, MI) to create a 3-dimensional model of the outer cortical and articular surfaces of an equine humerus. The CT scan contained 370 images of 2mm thick sections with a 1mm overlap (Ferullo, 2007). The surface of the scanned in solid model was automatically meshed using triangular elements in Mimics. Element reduction tools were used to reduce the total number of elements and increase element quality. A surface mesh of the humerus of 5154 total elements was then exported to Patran (MSC Software, Santa Ana, CA) (Ferullo, 2007).

A solid tetrahedral mesh was created using the surface mesh generated in Mimics. The 10-noded tetrahedral elements had 4 integration points (Ferullo, 2007). A tetrahedral mesh was applied to the solid model due to short processing time and low number of degrees of freedom required for convergence (Viceconti et al., 1998). A 1334N point load (Pollock et al., 2008a) was applied at approximately the shoulder joint contact force location of a standing horse on the mesh, and a static analysis was executed to find displacement at a standard location and to find the number of nodes required for results to converge (Ferullo, 2007) (Figure 7). The ABAQUS (SIMULIA, Providence, RI) finite element program was used to determine convergence solutions.

Figure 7. Demonstration of model convergence with maximum displacement (mm) of tetrahedral meshes depending on number of nodes (Ferullo, 2007).

A 13,596 element solid tetrahedral mesh with 20,569 nodes was selected as the final finite element model. Within this model, 96 of the most distal CT slices of the humerus were removed in Patran (MSC Software, Santa Ana, CA) while fixating the distal surface. The distal end of the humerus was fixated since it was determined that forces on the distal portion of the humerus do not affect proximal stresses in the areas of interest (Ferullo, 2007). This model was chosen due to a low run time, averaging 48 seconds/step for Ferullo's (2007) runtime and 10 seconds/step for the current study's runtime since the current study had a faster processor. Additionally, the current model was chosen due to its normal humeral geometry (Figure 8), rather than incorporating a humerus with a callus already present.

Figure 8. From left to right: medial, cranial, and caudal views of the mesh of the humerus displaying the regions of interest including the Stress Fracture Site (a) and the Caudal Diaphysis (b). The Cranial Aspect of the Diaphysis (c) was not an area of interest and was only used to validate the results from the region of the Caudal Diaphysis (b) (Ferullo, 2007).

Contact and Muscle Forces

The current model represents a racing Thoroughbred humerus which takes into account joint reaction forces, muscle forces, and four separate loading conditions representing fore-stance, mid-stance, aft-stance, and standing. Contact and muscle forces magnitudes and direction vary depending on position and speed during the gait cycle.

A right cadaveric forelimb from a euthanized 4-year-old Thoroughbred gelding was used in a previous study to determine muscle force, origin, insertion, and direction (Pollock et al., 2008a). At each muscle origin and insertion, metallic markers were embedded to determine location. Following this, a CT scan was performed while the horse was placed in anatomical standing position to determine three-dimensional

coordinates of each muscle's origin and insertion locations relative to one another (Beaupre et al., 1990a; Beaupre et al., 1990b). The humeral coordinate system origin was placed between the distal condyles on the humerus (Figure 9). EMG data during the stance phase of walking from equine (Korsgaard, 1982) and canine (Tokuriki, 1973) forelimb muscles was used to determine whether muscles were activated while the horse was walking.

Figure 9. CT scan of the lateral side of the scapula, humerus, radius, and ulna. Metallic markers are embedded to represent muscle origins and insertions, contact points, and additional key locations. The top axis (X, Y, Z) represents the global coordinate, while (x, y, z) represents the local humeral system. The x-axis was defined as distal to proximal, the y-axis was caudal to cranial, and the z-axis was lateral to medial (Pollock et al., 2008a).

The values of humeral muscle forces were predicted using optimization techniques and each muscle's maximum force during voluntary contraction (MVC) was found by multiplying physiological cross sectional areas (PCSAs) by an assumed specific

tension of 15 N/cm^2 (Powell et al., 1984). PCSA values were estimated by determining muscle volume and length during gross dissection (Pollock et al., 2008a). The biceps brachii contact force was found using a materials testing machine in vitro (Figure 10). The scapula contact force was determined after the forces of the muscles that crossed the shoulder joint in the model were determined (Pollock et al., 2008a). Tables 2 and 3 show activation levels of selected muscles attached to the humerus.

Figure 10. Example of a typical in vitro materials testing machine containing a lateral view of a left equine forelimb. This set up was used in Pollock et al.'s (2008b) study where a mechanical model (Musculoskeletal Model II) was used to validate the mathematical model (Musculoskeletal Model I) by comparing observed strains in the humerus at various locations (Pollock et al., 2008b).

There was no EMG data found for the subscapularis muscle, so its magnitude was varied until a physiological strain distribution was found (Ferullo, 2007). The brachialis muscle was found to be inactive during fore-stance and mid-stance in dogs, but proved to be a deep muscle after dissection, so EMG measurements may have been inaccurate or missed. Muscles labeled with a "0" or "1" from Tokuriki (1973) are indicated as inactive as well as "Off" from Korsgaard (1982). Like the subscapularis, the brachiocephalicus

and coracobrachialis muscles contained limited EMG data, so a physiologic strain distribution was used to determine force magnitudes. Muscles having small PCSAs and relatively small MVCs were not included in the final model (Ferullo, 2007).

Table 2. EMG walking data from an equine (Korsgaard, 1982) and a canine (Tokuriki, 1973). Canine EMG data ranged from 0 to 5, where 0 represents no activity and 5 corresponds to abundant activity. N/A indicates a lack of EMG data.

Muscle	Activation Level (Equine/Canine)		
	Fore-stance	Mid-stance	Aft-stance
Infraspinatus	On/3	On/3	On/1
Subscapularis	n/a	n/a	n/a
Supraspinatus (Lateral)	On/3	On/3	On/3
Supraspinatus (Medial)	On/3	On/3	On/3
Triceps (Lateral)	Slight/2	Off/2	Off/1
Triceps (Medial)	n/a/3	n/a/3	n/a/1
Deltoid (Long head)	n/a/1	n/a/1	n/a/3
Deltoid (Short head)	3	3	3
Teres Major	1	1	2
Brachialis	0	0	2
Brachiocephalicus	Off/0	Off/0	Off/1
Extensor Carpi Radialis	Off/0	Off/0	Off/1
Ulnaris Lateralis	On	On	Slight
Superficial Digital Flexor	On/2	On/2	Off/3

Table 3. Canine EMG data attached to the proximal humerus during gallop. An activation level of 5 corresponds to the highest EMG activity, while a level of 0 corresponds to no EMG activity (Tokuriki, 1973).

Muscle	Activation Level		
	Fore-stance	Mid-stance	Aft-stance
Infraspinatus	4	4	1
Supraspinatus (Lateral)	5	5	1
Supraspinatus (Medial)	5	5	1
Triceps (Lateral)	5	3	0
Triceps (Medial)	5	3	0
Deltoid (Long head)	2	2	4
Deltoid (Short head)	4	5	4
Teres Major	1	1	4
Brachialis	1	1	3
Brachiocephalicus	0	0	3

A total of nine muscles were incorporated into the model and include: the infraspinatus, subscapularis, brachialis, deltoid, teres major, lateral and medial triceps, coracobrachialis, and brachiocephalicus. Tables 8 and 9 show the force components of each muscle and contact force used in the ABAQUS (SIMULIA, Providence, RI) input file applied during the horse's natural gait cycle and during standing. Figure 11 displays muscle and contact force locations including orientations along the humerus. Each muscle force was applied over several nodes, which correspond to the muscle's attachment area. The scapula contact force was applied at a single node (Table 4), while the biceps brachii contact force was applied over several nodes. Depending on the phase of gait determines the activation level of each muscle. Table 4 displays the number of nodes for each force applied over a specific area and the node number of forces applied at a single node for the scapula force during gait and standing. Table 5 lists node references numbers and coordinates for contact and muscles forces.

Table 4. Number of nodes applied per force applied over a specific area during the gait cycle and standing. Italics for the scapula force represent the node number of forces applied at a single node (Ferullo, 2007).

	Number of Nodes (Node #)			
	Fore-stance	Mid-stance	Aft-stance	Standing
Scapula Force	<i>5053</i>	<i>8244</i>	<i>10030</i>	13
Infraspinatus	21	21	21	21
Infraspinatus (prox)	32	32	-	32
Subscapularis	33	33	33	33
Triceps (L)	16	16	-	-
Triceps (M)	21	21	-	-
Deltoid	-	112	112	112
Teres Major	10	-	10	-
Brachialis	119	119	119	119
Brachiocephalicus	-	-	26	-
Coracobrachialis	-	-	10	-
Biceps Contact	13	13	13	13

Table 5. Locations (mm) of muscle and contact forces for the humeral (Coordinate 1) and global (Coordinate 0) coordinate systems. The representative node is the center of the area for forces applied over more than one node (Ferullo, 2007).

Muscle/Joint Contact	Representative Node	Coordinate 1 Location	Coordinate 0 Location
Resultant Contact - Fore Stance	5053	314,-65,-64	85,-138,-36
Resultant Contact - Mid Stance	8244	297,-91,-47	59,-122,-53
Resultant Contact - Aft Stance	10030	298,-93,-28	57,-103,-63
Biceps Contact	1121	332,16,-40	166,-115,-17
Infraspinatus - Proximal	279	315,-35,23	115,-51,-35
Infraspinatus - Distal	7323	287,-12,13	137,-61,-63
Subscapularis	6600	299,-51,-98	98,-171,-51
Lateral Triceps	11458	268,-54,-9	95,-84,-82
Medial Triceps	16248	212,-13,-62	136,-136,-137
Deltoid	13798	201,5,13	154,-61,-148
Teres Major	16837	208,-37,-67	113,-141,-142
Brachialis	14857	245,-69,-48	80,-123,-105
Brachiocephalicus	18984	122,-3,-42	147,-116,-227
Coracobrachialis	19219	133,-22,-66	127,-140,-217

Figure 11. Free body diagrams of a left humerus with muscle and contact forces applied for all load cases from the (a) lateral, (b) medial, and (c) caudal views. Arrows along each axis indicate positive directions (Ferullo, 2007).

During each phase of gait, it was assumed that each joint contact force was proportional to vertical ground reaction forces, which are approximately 0.65 times body weight (Khumsap et al., 2002). While the horse was standing, each forelimb was assumed to support 30% of the horse's total weight with a ground vertical reaction force of 1335N (Dyce et al., 2002). The joint contact force applied to the model to simulate walking was 70% of the vertical ground reaction force (Ferullo, 2007). The total body weight of the horse was assumed to be 1,000 lbs (4,448 Newton's). From dissection of an equine scapula, humerus, radius, and ulna, joint contact force location and directions were estimated (Pollock et al., 2008a). Along with the mathematical model (Pollock et al., 2008a), digital radiographs assisted in finding joint contact force direction and location for each phase of gait (Ferullo, 2007). Joint contact forces from dissection and radiographs were applied to each phase of stance with varying directions and locations. A porosity distribution was produced for each set of loads using the remodeling simulation applied to the humerus model. Joint contact forces and directions were

adjusted until the porosity distribution seen in the current model was similar to radiographs, where a dense band of trabecular bone could be seen from compressive contact forces, which produced the current joint contact and muscle forces for the dirt track surface shown in Figure 11 and Tables 8-9 (Ferullo, 2007).

Remodeling Algorithm

A previously developed remodeling algorithm by Hazelwood et al. (2001) was used in the current model to study the cause and development of stress fractures. This algorithm simulates porosity and changes in elastic modulus caused by internal bone remodeling initiated by disuse, damage, and environmental mechanical stimuli. For each loading condition in the current model, the simulation tracks 8 state variables, which are updated during each step of the analysis and include: elastic modulus, porosity, number of resorbing and refilling BMUs (BMUs/mm^2), BMU activation frequency ($\text{BMUs}/\text{mm}^2/\text{day}$), damage (mm/mm^2), strain ($\mu\epsilon$), and mechanical stimulus (cycles per day). Tables 6 and 7 display values assigned to initial state variables and constants. Figure 6, shown previously, displays the relationship between state variables and whether remodeling results in damage removal or damage formation.

Table 6. Initial values for state variables used in the bone remodeling algorithm (Hazelwood et al., 2001).

Variable (units)	Description	Initial Value
p (void volume/total volume)	Porosity	0.0443213
s ($\mu\epsilon$)	Strain Fore-Stance	0
s ($\mu\epsilon$)	Strain Mid-Stance	0
s ($\mu\epsilon$)	Strain Aft-Stance	0
s ($\mu\epsilon$)	Strain Standing	0
Φ (cpd)	Mechanical Stimulus	0
D (mm/mm ²)	Damage	0.0366294
N_F (BMUs/mm ²)	Number of Refilling BMUs	0.4228
N_R (BMUs/mm ²)	Number of Resorbing BMUs	0.1675
f_a (BMUs/mm ² /day)	BMU Activation Frequency	0.0067

Table 7. Constants used in the remodeling simulation. Loading rates are given in cycles per day (Ferullo, 2007).

Constant (units)	Description	Nominal Value
A (mm ²)	BMU Cross Sectional Area	2.84×10^{-2}
T_R (days)	Resorption Period	24
T_I (days)	Reversal Period	8
T_F (days)	Refilling Period	64
K_D (mm/mm ²)	Damage Rate Coefficient	1.85×10^5
n	Loading Conditions	4
q	Damage Rate Exponent	4
R_{L1} (cpd)	Fore-stance Loading Rate	600
R_{L2} (cpd)	Mid-stance Loading Rate	600
R_{L3} (cpd)	Aft-stance Loading Rate	600
R_{L4} (cpd)	Standing Loading Rate	10
F_S	Damage Removal Specificity Factor	5
D_0 (mm/mm ²)	Initial Damage	0.0366294
f_{a0} (BMUs/mm ² /day)	Initial BMU Activation Frequency	0.00670
Φ_0 (cpd)	Initial Mechanical Stimulus	1.875×10^{-10}
$f_{a(max)}$ (BMUs/mm ² /day)	Maximum BMU Activation Frequency	0.50
k_b (cpd ⁻¹)	Activation Frequency Dose-response Coefficient	6.5×10^{10}
k_c (cpd)	Activation Frequency Dose-response Coefficient	9.4×10^{-11}
k_r	Activation Frequency Dose-response Coefficient	-1.6

The remodeling algorithm is incorporated with the finite element model and bone porosity, damage, and BMU activation frequency were recorded for each element at every step. The load frequency and magnitude or mechanical stimulus determines whether a particular element of bone is in disuse or damage. The relationships between state variables are described below.

In the algorithm, elastic modulus, E , is related to porosity, p , with the equation:

$$E = (8.83 \times 10^5)p^6 - (2.99 \times 10^6)p^5 + (3.99 \times 10^6)p^4 - (2.64 \times 10^6)p^3 + (9.08 \times 10^6)p^2 - (1.68 \times 10^5)p + (2.37 \times 10^4) \quad (1)$$

The rate of change in porosity ($\dot{p}$) is dependent on mean bone refilling (Q_F) and resorbing (Q_R) rates, as well as the density of resorbing (N_R) and refilling (N_F) BMUs/area. In equation (2), $Q_R = A/T_R$ and $Q_F = A/T_F$, where A represents the cross-sectional area of each BMU, while T_R and T_F represent the resorption and refilling periods, respectively.

$$\dot{p} = Q_R N_R - Q_F N_F \quad (2)$$

N_R and N_F were found by integrating over BMU resorption (T_R), reversal (T_I), and refilling (T_F) periods of BMU activation frequency (f_a) history. The resorption period (T_R) begins when teams of osteoclasts are recruited to create a resorbing BMU and ends when osteoclasts no longer resorb. The reversal period (T_I) is the time it takes to transition from osteoclastic to osteoblastic activity. When osteoblasts of the BMU begin forming new bone, the refilling period (T_F) is initiated.

$$N_R = \int_{t-T_R}^t f_a(t') dt' \quad (3)$$

$$N_F = \int_{t-(T_R+T_I+T_F)}^{t-(T_R+T_I)} f_a(t') dt' \quad (4)$$

BMU activation frequency is a result of both disuse and damage, and was scaled by the specific internal surface area (S_A) to make sure the area of BMUs do not exceed available surface area.

$$f_a = (f_{a(disuse)} + f_{a(damage)}) S_A \quad (5)$$

Damage (D) is the total crack length per area of bone and the rate of damage accumulation is the difference between fatigue damage formation ($\dot{D}_F$) and removal rates ($\dot{D}_R$).

$$\dot{D} = \dot{D}_F - \dot{D}_R \quad (6)$$

The rate of damage formation is proportional to the product of the strain range (s , in $\mu\epsilon$) raised to a power and the loading rate (R_L , cycles per time), which is summed over 4 loading conditions representing standing, fore-stance, mid- stance, and aft-stance.

$$\dot{D}_F = k_D \sum_{i=1}^4 s_i^q R_{Li} = k_D \Phi \quad (7)$$

Since damage initiates BMU activation, the efficiency of damage removal is greater than for random remodeling. For this to happen, a damage removal specificity factor, F_S , was placed into the equation, which is a determinant of damage removal rate, along with damage, BMU cross-sectional area, and BMU activation frequency.

$$\dot{D}_R = D f_a A F_S \quad (8)$$

The damage rate coefficient (k_D) is dependent on equilibrium damage (D_0), equilibrium activation frequency (f_{a0}), BMU cross-sectional area (A), damage removal specificity factor (F_S), and equilibrium mechanical stimulus (Φ_0).

$$k_D = D_0 f_{a0} A F_S / \Phi_0 \quad (9)$$

When bone is in disuse, there is an assumed sigmoidal relationship between BMU activation frequency (f_a) and damage (D). For a bone to be in disuse, the current mechanical stimulus Φ must be less than the equilibrium mechanical stimulus Φ_0 .

$$f_{a(\text{disuse})} = \frac{f_{a(\text{max})}}{1 + e^{k_b(\Phi - k_c)}} \quad \text{for } \Phi < \Phi_0 \quad (10)$$

Also, when bone reaches the point where damage starts to occur, there is an assumed sigmoidal relationship between BMU activation frequency (f_a) and damage (D).

$$f_{a(\text{damage})} = \frac{(f_{a0})(f_{a(\text{max})})}{f_{a0} + (f_{a(\text{max})} - f_{a0})e^{\left[\frac{k_r(f_{a(\text{max})})(D - D_0)}{D_0}\right]}} \quad (11)$$

Creation of Equilibrium Model

Prior to the update of the current model, Ferullo (2007) developed an equilibrium model by running the simulation for 1001 days. For every 10 days, remodeling parameters were examined to determine when each parameter reached equilibrium, which was defined as a change in BMU activation frequency, porosity, and damage of less than 5% over a period of 10 days. Remodeling parameters were tracked at two sites of interest: the caudoproximal stress fracture region and a location on the diaphysis on the

caudal aspect of the humerus. These two sites were chosen since mechanical and material data of bone was collected at these two sites (Entwistle et al., 2009).

Remodeling parameter results were averaged over 6 elements at the stress fracture site and averaged over 2 elements at the caudal diaphysis. For the model to reach equilibrium, 143 weeks of walking was applied, which included fore, mid, and aft-stances. Additionally, there was a period of 143 weeks of standing in the model to achieve equilibrium. Once the model reached equilibrium, state variable values were gathered and used as initial conditions for all training programs (Ferullo, 2007).

Thoroughbred race horses normally don't begin training until they are 2 years of age (Boston and Nunamaker, 2000), so the period of equilibrium represents the horse's adaption and physical activities prior to the commencement of training. Joint contact forces, muscle forces, boundary conditions, and material properties were applied to the finite element model to simulate different training programs. Muscle activations and joint contact forces were scaled upward from walking to represent different galloping speeds, which include slow and fast gallops (Ferullo, 2007). Muscle and joint contact forces used for standing and walking are shown in Tables 8 and 9.

Slow gallop and fast gallop joint contact forces were estimated based on peak vertical ground reaction forces, which were based on accelerometer data from a previous study (Witte et al., 2006). In this model, a slow gallop was defined as a velocity of 10m/s at 2.8 times walking force with a peak vertical force of 1.8 times body weight. A fast gallop was defined as a velocity of 17m/s (Swanstrom et al., 2005) at 3.5 times walking force with a peak vertical force of 2.3 times body weight. All contact and muscle forces were assumed to be proportional to walking forces, and were based on muscle MVC and

equine EMG data for walking. The scaling factor for calculating forces was used in this simulation to create an acceptable running time (Ferullo, 2007). Using vertical ground reaction forces as a known value, joint contact and muscle force magnitudes for galloping were scaled from the joint contact force for walking at 50% body weight (Ferullo, 2007).

Table 8. Muscle and contact forces components during the standing load case used in the FEA input file for dirt (Ferullo, 2007), turf, and synthetic tracks. Dirt track values were previously adjusted from a range of literature to fit an accurate physiological porosity distribution (Ferullo, 2007).

	Dirt Standing (N)			Turf Standing (N)			Synthetic Standing (N)		
	(x)	(y)	(z)	(x)	(y)	(z)	(x)	(y)	(z)
Scapula Contact	-122.5	129.8	28.5	-118	125	27.4	-96.8	102.5	22.5
Infraspinatus	0.63	-2.75	0.128	0.61	-2.65	0.123	0.5	-2.17	0.1
Infraspinatus (prox)	0.03	-1.5	0.42	0.03	-1.44	0.4	0.02	-1.19	0.33
Subscapularis	0.16	-2	-0.92	0.15	-1.93	-0.89	0.13	-1.58	-0.73
Deltoid	1.1	-1.6	1.3	1.06	-1.54	1.25	0.87	-1.26	1.03
Brachialis	-3.7	-2.4	-0.26	-3.56	-2.31	-0.25	-2.92	-1.9	-0.21
Biceps Contact	-106.8	-104	-23	-102.8	-100.2	-22.1	-84.4	-82.2	-18.2

Table 9. Dirt track muscle and contact force components during the three separate phases of walking placed in the FEA input file. These forces were increased by a factor of 2.8 and 3.5 within the algorithm to simulate slow and fast galloping speeds, respectively (Ferullo, 2007).

	Dirt Track Walking Muscle and Contact Forces (N)								
	Fore(x)	Fore(y)	Fore(z)	Mid(x)	Mid(y)	Mid(z)	Aft(x)	Aft(y)	Aft(z)
Scapula Contact	-1750	700	550	-1350	1400	-200	-750	1750	-500
Infraspinatus	9.5	-36.3	1.67	3.8	-16.5	0.77	1.26	-5.5	0.26
Infraspinatus (prox)	2.5	-9	0.18	2.5	-9	0.18	0	0	0
Subscapularis	1.7	-5.7	-2.55	1.7	-5.7	-2.55	1.03	-3.4	-1.5
Triceps Brachii (L)	-13	-1.25	0.6	-13	-1.25	0.6	0	0	0
Triceps Brachii (M)	-10.35	-3	-1.23	-10.35	-3	-1.23	0	0	0
Deltoid	0	0	0	0	0	0	2.15	-3	2.6
Teres Major	4.8	-14.5	-4.9	0	0	0	4.8	-14.5	-4.9
Brachialis	-10	-6.5	-0.7	-10	-6.5	-0.7	-15	6	-1
Brachiocephalicus	0	0	0	0	0	0	24	5.6	8.4
Coracobrachialis	0	0	0	0	0	0	7.4	2.9	-0.24
Biceps Contact	-106.8	-104	-23	-106.8	-104	-23	-106.8	-104	-23

Rescaling Contact Forces for Turf and Synthetic Tracks

The previous finite element model contained muscle and contact forces which simulated training on a dirt track. To further investigate how different track surface materials impact a racing Thoroughbred's humerus, muscle and contact forces were updated to simulate synthetic and turf track surfaces for all four load cases, which were based on a previous study (Setterbo et al., 2007). This study used an accelerometer and a dynamometric horseshoe to measure forelimb hoof accelerations and ground reaction forces during trot and canter (Figure 12) (Setterbo et al., 2007). In the current study, two scaling factors were used from ground reaction forces to recalculate muscle and contact forces from dirt surface values to match turf and synthetic forces.

Figure 12. A dynamometric horseshoe fixed to the solar surface of the hoof recorded ground reaction forces (GRF), while a tri-axial piezoelectric accelerometer fixed to the dorsal part of the hoof wall of the left forelimb recorded accelerations (ACC). The coordinate arrows represent the positive Z and X axes of each coordinate system, and were fixed to the solar surface of the hoof (Setterbo et al., 2007).

The previous study by Setterbo et al. (2007) was performed on three three-year-old females, where hoof accelerations, ground reaction forces, and horse speed were measured during trot and canter on a dirt racetrack, synthetic training track, and a turf racetrack at one race course. The turf track evaluated consisted of long grass, which was approximately 4-6 inches in height, while the synthetic track was a proprietary mixture of wax-coated silica sand, polypropylene fibers, and recycled rubber (Setterbo et al., 2007). The average canter speed was 1.8 strides/s and the average trot speed was 1.5 strides/s. It was found that the synthetic track had the lowest Z-force (vertical reaction force), while the turf track had the second lowest Z-force followed by the dirt track, which had the largest Z-force during trot (Figure 13). The peak Z-force from the synthetic and turf track surfaces during trot were 79% and 96.3% of the dirt track surface, respectively.

Two separate types of peak horizontal ground reaction forces (braking and propulsion) exhibited no statistical significant differences between track surface types during canter, but were statistically different during trot (Setterbo et al., 2007). Braking and propulsion peak horizontal forces were not incorporated into the current model since they occur in directly opposing directions along the x-axis (Figure 12). Additionally, the X-force variables (braking and propulsion) were less similar between canter and trot data, while the Z-force canter and trot data contained similar statistical trends between track surfaces for most variables (Setterbo et al., 2007).

All muscle and contact forces used in this model for every load case were scaled down based on the average Z-force peak used from Setterbo et al. (2007) (Figure 13). The Z-force peak maximum (N/kg) represents the maximum vertical reaction force over the total mass of the horse, rider, saddle, and instrumentation equipment. The mean Z-

force peak maximums (N/kg) for dirt, turf, and synthetic track surfaces were 13.5 (N/kg), 13.0 (N/kg), and 10.7 (N/kg), respectively (Setterbo et al., 2007). Trot data was used due to a lower speed compared to canter implying a closer assumption to walking and standing in the current model. Additionally, more trials were conducted during trot, consequently, providing more statistically effective data.

Figure 13. Dirt, turf, and synthetic vertical ground reaction forces (Z-force) during trot. Trot was defined as having a mean velocity of roughly 5 m/s (Setterbo et al., 2007).

The muscle and contact forces while standing and walking were gathered from Pollock et al. (2008a; 2008b) and Ferullo (2007) previously shown in Tables 8 and 9. These forces were based on dirt track surfaces, and were scaled down to calculate the muscle and contact forces for turf and synthetic tracks. The muscle and contact forces were scaled down from the dirt track force by a factor, which depended on track surface (0.79 for synthetic and 0.963 for turf). This was done for all muscle and contact forces during all load cases (Tables 8, 10-11). A scaling factor had to be used to update muscle

and contact forces for turf and synthetic tracks since it was an indeterminate case, meaning more unknowns than equations, so it was assumed that the scaling factors (0.79 for synthetic and 0.963 for turf) for all muscles and contact forces in the model represented different track surfaces, even though the scaling factors were based on Setterbo et al.'s (2007) coordinate system and average maximum Z-force peak during trot. Scaled down muscle and contact forces during walking for turf and synthetic tracks are summarized below in Tables 10-11.

Table 10. Scaled down muscle and contact force components for the turf track during the three phases of walking.

	Turf Track Walking Muscle and Contact Forces (N)								
	Fore(x)	Fore(y)	Fore(z)	Mid(x)	Mid(y)	Mid(z)	Aft(x)	Aft(y)	Aft(z)
Scapula Contact	-1685	674	530	-1300	1348	-193	-722	1685	-482
Infraspinatus	9.15	-34.96	1.61	3.66	-15.89	0.74	1.21	-5.3	0.25
Infraspinatus (prox)	2.41	-8.67	0.17	2.41	-8.67	0.17	0	0	0
Subscapularis	1.64	-5.49	-2.46	1.64	-5.49	-2.46	0.99	-3.27	-1.44
Triceps Brachii (L)	-12.52	-1.2	0.58	-12.52	-1.2	0.58	0	0	0
Triceps Brachii (M)	-9.97	-2.89	-1.18	-9.97	-2.89	-1.18	0	0	0
Deltoid	0	0	0	0	0	0	2.07	-2.89	2.5
Teres Major	4.62	-13.96	-4.72	0	0	0	4.62	-13.96	-4.72
Brachialis	-9.63	-6.26	-0.67	-9.63	-6.26	-0.67	-14.44	5.78	-0.96
Brachiocephalicus	0	0	0	0	0	0	23.11	5.39	8.09
Coracobrachialis	0	0	0	0	0	0	7.13	2.79	-0.23
Biceps Contact	-102.8	-100.2	-22.1	-102.8	-100.2	-22.1	-102.8	-100.2	-22.1

Table 11. Scaled down muscle and contact force components for the synthetic track during the three phases of walking.

	Synthetic Track Walking Muscle and Contact Forces (N)								
	Fore(x)	Fore(y)	Fore(z)	Mid(x)	Mid(y)	Mid(z)	Aft(x)	Aft(y)	Aft(z)
Scapula Contact	-1383	553	435	-1067	1106	-158	-593	1383	-395
Infraspinatus	7.51	-28.68	1.32	3	-13.04	0.61	1	-4.35	0.21
Infraspinatus (prox)	1.98	-7.11	0.14	1.98	-7.11	0.14	0	0	0
Subscapularis	1.34	-4.5	-2.01	1.34	-4.5	-2.01	0.81	-2.69	-1.19
Triceps Brachii (L)	-10.27	-0.99	0.47	-10.27	-0.99	0.47	0	0	0
Triceps Brachii (M)	-8.18	-2.37	-0.97	-8.18	-2.37	-0.97	0	0	0
Deltoid	0	0	0	0	0	0	1.7	-2.37	2.05
Teres Major	3.79	-11.46	-3.87	0	0	0	3.79	-11.46	-3.87
Brachialis	-7.9	-5.14	-0.55	-7.9	-5.14	-0.55	-11.85	4.74	-0.79
Brachiocephalicus	0	0	0	0	0	0	18.96	4.42	6.64
Coracobrachialis	0	0	0	0	0	0	5.85	2.29	-0.19
Biceps Contact	-84.4	-82.2	-18.2	-84.4	-82.2	-18.2	-84.4	-82.2	-18.2

Training Programs

Four different Thoroughbred racing horse training regimes were modeled in the current study on all track surfaces (dirt, turf, and synthetic) by analyzing BMU activation frequency, porosity, and damage. Baseline Layup and Long Layup training programs developed by Ferullo (2007) were compared between each other and to the results from the previously validated model of Ferullo (2007). The race programs were compared between each other, but not between the Baseline Layup and Long Layup training programs since they have different lengths of training (168 days vs. 350 days) and the race programs do not have a period of layup. Every training program modeled started from the same initial equilibrium conditions. Duration for the Baseline Layup and Long Layup training programs lasted a total of 350 days or 50 weeks, while the two race programs only lasted 168 days or 24 weeks. Common horse training exercises involve a slow gallop (10m/s) for 5 days/week, followed by a timed workout or race (17m/s) on the 6th day, and finally a walk or layup on the 7th day; this sequence is known as initial training (Ferullo, 2007).

For the Baseline Layup training program, there were 13 weeks of initial training, 8 weeks of layup, and 29 additional weeks of training (Table 12). For initial training, the day 6 fast gallop distance starts at 0.125 miles and increases by 0.125 every other week until a distance of 0.75 miles is reached. Increasing workout intensity over time is a standard practice seen in the racehorse industry for beginning young Thoroughbreds (Rogers and Firth, 2004). The slow gallop (days 1-5) distance of 0.7 is maintained throughout the entire program except when there is a period of layup in which the horse does not complete a slow gallop workout. Slow and fast gallop speeds of 10m/s and

17m/s were based on stride rates of 2.0 strides/second and 2.2 strides/second, respectively (Witte et al., 2006). After initial training, there is 8 weeks of layup in which the horse only undergoes 0.5 hours of walking and standing each day. Following the period of layup, the horse returns to training and begins day 6 fast gallop at a distance of 0.5 miles for the first week and increases to a distance of 0.75miles the following week, which is maintained throughout the rest of the program. The Long Layup training program has 13 weeks of initial training, 16 weeks of layup, and 21 additional weeks of training (Table 13). The only difference between the Baseline Layup and Long Layup training programs is the duration of the layup and return to training periods where the lengths of layup for the Baseline Layup and Long Layup programs are 8 and 16 weeks, respectively.

Table 13. Long Layup training program with 16 weeks of layup. Distances are given in miles, furlongs, and number of strides during the high speed exercise completed on day 6 (Ferullo, 2007).

Long Layup Training Program																
Initial Training																
Week	1	2	3	4	5	6	7	8	9	10	11	12	13			
Days Slow Gallop	1--5	8--12	15--19	22-16	29-33	36-40	43-47	50-54	57-61	64-68	71-75	78-82	85-89			
Day 6 Fast Gallop	6	13	20	27	34	41	48	55	62	69	76	83	90			
Day 7 Walk (½ hour)	7	14	21	28	35	42	49	56	63	70	77	84	91			
Day 6 Distance(mi)	0.125	0.125	0.25	0.25	0.375	0.375	0.5	0.5	0.625	0.625	0.75	0.75	0.75			
Day 6 Distance (furlong)	1	1	2	2	3	3	4	4	5	5	6	6	6			
Day 6 No. of Strides	27	27	54	54	81	81	108	108	135	135	162	162	162			
Long Layup																
Week	14	15	16	17	18	19	20	21	22	23	24	25	26	27	28	29
Days Layup	92-98	99-105	106-112	113-119	120-126	127-133	134-140	141-147	148-154	155-161	162-168	169-175	176-182	183-189	190-196	197-203
Distance	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
Restart Training																
Week	30	31	32	33	34	35										
Days Slow Gallop	204-208	211-215	218-222	225-229	232-236	239-243										
Day 6 Fast Gallop	209	216	223	230	237	244										
Day 7 Walk (½ hour)	210	217	224	231	238	245										
Day 6 Distance(mi)	0.5	0.75	0.75	0.75	0.75	0.75										
Day 6 Distance (furlong)	4	6	6	6	6	6										
Day 6 No. of Strides	108	162	162	162	162	162										
Restart Training																
Week	36	37	38	39	40	41	42	43	44	45	46	47	48	49	50	
Days Slow Gallop	246-250	253-257	260-264	267-271	274-278	281-285	288-292	295-299	302-306	309-313	316-320	323-327	330-334	337-341	344-348	
Day 6 Fast Gallop	251	258	265	272	279	286	293	300	307	314	321	328	335	342	349	
Day 7 Walk (½ hour)	252	259	266	273	280	287	294	301	308	315	322	329	336	343	350	
Day 6 Distance(mi)	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	0.75	
Day 6 Distance (furlong)	6	6	6	6	6	6	6	6	6	6	6	6	6	6	6	
Day 6 No. of Strides	162	162	162	162	162	162	162	162	162	162	162	162	162	162	162	

Both of the newly implemented race programs do not contain a prolonged period of layup, but only a layup occurring on the 7th day of every week (Tables 14-16). The race programs begin with initial training for the first 8 weeks, where fast gallop distances begin at 0.125 miles, but only reach a distance of 0.5 miles at the end of the 8 week period. Following initial training is a period of race training which occurs for 16 weeks (Tables 15-16). Race Program 1 starts the first week of race training with a race on the

6th day followed by a week with a slow gallop, while Race Program 2 begins the first week with a slow gallop, but has a race on the 6th day on the following week. Races occur once every 4 weeks following the first race for both race programs, where a fast gallop is completed on the other two weeks for this period (Table 14). During the 3 weeks between races, Race Program 1 has a slow gallop on the 6th day (instead of a fast gallop) of the week following race week. After a week of race absence, the horse undergoes a fast gallop on the 6th day for the final two weeks (Table 14). For the 3 weeks between races for Race Program 2, the first two weeks contain fast gallops while the last week's 6th day is replaced with a slow gallop (Tables 14 and 16).

Table 14. Comparison of the first four weeks of race training for the race programs.

Race Program 1				
Race Training				
Week	9	10	11	12
Days Slow Gallop	57-61	64-69	71-75	78-82
Day 6 Fast Gallop	62(Race)	N/A	76	83
Day 7 Walk (½ hour)	63	70	77	84
Day 6 Distance(mi)	0.75	0	0.625	0.625
Day 6 Distance (furlong)	6	0	5	5
Day 6 No. of Strides	162	0	135	135
Race Program 2				
Race Training				
Week	9	10	11	12
Days Slow Gallop	57-62	64-68	71-75	78-82
Day 6 Fast Gallop	N/A	69(Race)	76	83
Day 7 Walk (½ hour)	63	70	77	84
Day 6 Distance(mi)	0	0.75	0.625	0.625
Day 6 Distance (furlong)	0	6	5	5
Day 6 No. of Strides	0	162	135	135

Table 15. Race Program 1 with no extended period of layup. High speed gallop and race distances are given in miles, furlongs, and number of strides. Races occur once every four weeks during the race training period of the program.

Race Program 1																
Initial Training																
Week	1	2	3	4	5	6	7	8								
Days Slow Gallop	1-5	8-12	15-19	22-26	29-33	36-40	43-47	50-54								
Day 6 Fast Gallop	6	13	20	27	34	41	48	55								
Day 7 Walk (½ hour)	7	14	21	28	35	42	49	56								
Day 6 Distance(mi)	0.125	0.125	0.25	0.25	0.375	0.375	0.5	0.5								
Day 6 Distance (furlong)	1	1	2	2	3	3	4	4								
Day 6 No. of Strides	27	27	54	54	81	81	108	108								
Race Training																
Week	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24
Days Slow Gallop	57-61	64-69	71-75	78-82	85-89	92-97	99-103	106-110	113-117	120-125	127-131	134-138	141-145	148-153	155-159	162-166
Day 6 Fast Gallop	62(Race)	N/A	76	83	90(Race)	N/A	104	111	118(Race)	N/A	132	139	146(Race)	N/A	160	167
Day 7 Walk (½ hour)	63	70	77	84	91	98	105	112	119	126	133	140	147	154	161	168
Day 6 Distance(mi)	0.75	0	0.625	0.625	1	0	0.75	0.75	1.25	0	0.75	0.75	1.5	0	0.75	0.75
Day 6 Distance (furlong)	6	0	5	5	8	0	6	6	10	0	6	6	12	0	6	6
Day 6 No. of Strides	162	0	135	135	216	0	162	162	270	0	162	162	324	0	162	162

Table 16. Race Program 2 with no extended period of layup. High speed gallop and race distances are given in miles, furlongs, and number of strides. Races occur once every four weeks during the race training period of the program. This program is a week behind the Race Program 1 in scheduled races.

Race Program 2																
Initial Training																
Week	1	2	3	4	5	6	7	8								
Days Slow Gallop	1-5	8-12	15-19	22-16	29-33	36-40	43-47	50-54								
Day 6 Fast Gallop	6	13	20	27	34	41	48	55								
Day 7 Walk (½ hour)	7	14	21	28	35	42	49	56								
Day 6 Distance(mi)	0.125	0.125	0.25	0.25	0.375	0.375	0.5	0.5								
Day 6 Distance (furlong)	1	1	2	2	3	3	4	4								
Day 6 No. of Strides	27	27	54	54	81	81	108	108								
Race Training																
Week	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24
Days Slow Gallop	57-62	64-68	71-75	78-82	85-90	92-96	99-103	106-110	113-118	120-124	127-131	134-138	141-146	148-152	155-159	162-166
Day 6 Fast Gallop	N/A	69(Race)	76	83	N/A	97(Race)	104	111	N/A	125(Race)	132	139	N/A	153(Race)	160	167
Day 7 Walk (½ hour)	63	70	77	84	91	98	105	112	119	126	133	140	147	154	161	168
Day 6 Distance(mi)	0	0.75	0.625	0.625	0	1	0.75	0.75	0	1.25	0.75	0.75	0	1.5	0.75	0.75
Day 6 Distance (furlong)	0	6	5	5	0	8	6	6	0	10	6	6	0	12	6	6
Day 6 No. of Strides	0	162	135	135	0	216	162	162	0	270	162	162	0	324	162	162

III. Results

Equilibrium Model and Remodeling Parameters

It took approximately 180 days for porosity to reach a state of equilibrium (differential of less than 5%) at the stress fracture site (Figure 14). Equilibrium for damage was reached at the 220th day (Figure 15), while BMU activation frequency did not reach equilibrium until the 900th day (Figure 16) (Ferullo, 2007). After 1001 days of remodeling, equilibrium values for state variables porosity, damage, and BMU activation frequency at the stress fracture site were 16%, 0.0299 mm/mm², and 0.0109 BMUs/mm²/day, respectively. Equilibrium values for porosity, damage, and BMU activation frequency at the caudal diaphysis were 6.5%, 0.1003 mm/mm², and 0.0114 BMUs/mm²/day, respectively.

Figure 14. Percent change in porosity recorded every 10 days. The blue lines indicate the criteria for equilibrium, which is a less than 5% change in remodeling parameters (Ferullo, 2007).

Figure 15. Percent change in damage recorded every 10 days. The blue lines indicate the criteria for equilibrium, which is a less than 5% change in remodeling parameters (Ferullo, 2007).

Figure 16. Percent change in BMU activation frequency recorded every 10 days. The blue lines indicate the criteria for equilibrium, which is a less than 5% change in remodeling parameters (Ferullo, 2007).

Baseline Layup Training Program

Stress Fracture Site

During initial training (weeks 1-13), BMU activation frequency and porosity remained constant at the stress fracture site, while damage increased for both the Baseline Layup and Long Layup training programs. For the beginning of layup (week 14), both the Baseline Layup and Long Layup programs showed a sharp increase in BMU activation frequency (Figures 17-18); which increased from 0.005801 BMUs/mm²/day to 0.4694 BMUs/mm²/day over a 10 day period starting on the 90th day. This corresponded to an increase in porosity from 16.7% on day 90 to 28.2% on day 130 (Figures 19-20). There was also a decrease in damage from 0.05318 mm/mm² on day 90 to 0.03745 mm/mm² on day 130 (Figures 21-22). When training is restarted for the Baseline Layup program (week 22), BMU activation frequency remains steady as it did during initial training, while damage increases significantly through the end of training to 0.09266 mm/mm². Porosity gradually decreased following layup to 17.2% but then increased to 18.3% at the end of training.

Caudal Diaphysis

For initial training during the Baseline Layup and Long Layup programs, BMU activation frequency increased from 0.01430 BMUs/mm²/day to 0.05682 BMUs/mm²/day at the caudal diaphysis (Figures 17-18), while porosity increased from 6.5% to 9.2 % (Figures 19-20). Percent increase in damage seen at the caudal diaphysis was similar to the trend seen at the stress fracture site during initial training (Figures 21-22). During the period of layup, BMU activation frequency declined from 0.05682 BMUs/mm²/day on day 90 (End of Initial Training) to 0.03329 BMUs/mm²/day on day

150 (beginning of Restart Training) and did not experience a large peak value, which occurred at the stress fracture site. Porosity continued to steadily increase during the beginning of layup to 10.8% on day 130, but then decreased following layup to 9.9% on day 170. Damage decreased from 0.1670 mm/mm^2 to 0.1374 mm/mm^2 during layup (Figures 21-22), which is not as drastic as the trend seen at the stress fracture site. Upon return to training for the Baseline Layup program, BMU activation frequency increased through the end of the program to $0.1770 \text{ BMUs/mm}^2/\text{day}$. This corresponded to a 121% increase in porosity during the same period. Damage increased from 0.1374 mm/mm^2 to 0.2110 mm/mm^2 . Damage values are greater at the caudal diaphysis, but the stress fracture site has a greater percent increase in damage over the equilibrium value (Figures 21-22).

Long Layup Training Program

Stress Fracture Site

The first 147 days of the Long Layup training program are exactly the same as the Baseline Layup training program. During the weeks of long layup (Weeks 14-29), there are two points where there is a substantial increase in BMU activation frequency, the second being the lesser of the two (Figures 17-18). The second peak in BMU activation frequency occurs during the latter half of long layup and increases from $0.004615 \text{ BMUs/mm}^2/\text{day}$ on the 140th day to $0.1801 \text{ BMUs/mm}^2/\text{day}$ on the 180th day. During this period of layup, the stress fracture site returns to a state of disuse and attempts to reach equilibrium values causing oscillations in BMU activation frequency (Ferullo, 2007). This corresponded to a 22.5% decrease in porosity and a 4% decrease in damage at the same two points in time (Figures 19-22). When training is reestablished for the

Long Layup training program (Weeks 30-49), BMU activation frequency increased from 0.004483 BMUs/mm²/day to 0.01776 BMUs/mm²/day. Additionally, damage increased from 0.04047 mm/mm² to 0.09415 mm/mm² (Figures 21-22) during the restart training period, while porosity decreased from 23.5% to 19.3%.

Caudal Diaphysis

During long layup, BMU activation frequency, porosity, and damage experienced a decrease of 69%, 12%, and 31% from the 90th day to the 200th day of layup, respectively. Compared to the stress fracture site's BMU activation frequency increase of 192%, the caudal diaphysis experienced a greater increase in BMU activation frequency of 340% (Figures 17-18) from day 220 to the end of training. Porosity had an increase of 151% during the restart of the training period (Figures 19-20). Damage experienced an increase of 36% for the same time period (day 220 to end of training) (Figures 21-22). Upon return to training, damage increased initially then leveled out during the end of training.

Model Validation (Hypothesis 1)

The first objective and hypothesis of the current study was to validate the current model, which was based on a preexisting, previously validated model created by Ferullo (2007). Validation of the current model was successfully accomplished by comparing Ferullo's (2007) results with the Baseline Layup and Long Layup training program's shown in Figures 17-22. Model validation was successful for BMU activation frequency, porosity, and damage at both sites on the humerus. It is important to validate computational models since the environment of a computational workstation can vary, such as employing different operating systems or versions of a finite element program,

which can have a large impact on results, duration of simulation, and reliability for any given identical model with the same ABAQUS (SIMULIA, Providence, RI) input and FORTRAN script files.

Figures 17-18. BMU activation frequency (BMUs/mm²/day) recorded every 10 days for 350 days at the stress fracture site and caudal diaphysis during the Baseline Layup and Long Layup training programs (Ferullo, 2007). Model comparison between Ferullo's (2007) results (top) and current model results (bottom) validate the current model.

Figures 19-20. Porosity (%) recorded every 10 days for 350 days at the stress fracture site and caudal diaphysis during the Baseline Layup and Long Layup training programs (Ferullo, 2007). Model comparison between Ferullo's (2007) results (top) and current model results (bottom) validate the current model.

Figures 21-22. Percent increase in damage over the equilibrium value (stress fracture site = 0.0299mm/mm^2 , caudal diaphysis = 0.1003 mm/mm^2) recorded every 10 days for 350 days at the stress fracture site and caudal diaphysis during the Baseline Layup and Long Layup training programs (Ferullo, 2007). Model comparison between Ferullo's (2007) results (top) and current model results (bottom) validate the current model.

Race Programs

Race Program 1

During initial training, BMU activation frequency remained constant, which corresponded to an almost constant porosity at the stress fracture site (increasing by only 2.6% throughout the entire program). BMU activation frequency at the caudal diaphysis displayed a linear increase throughout the entire program increasing from 0.05690 BMUs/mm²/day to 0.08641 BMUs/mm²/day (Figure 25). Porosity at the caudal diaphysis remained fairly constant during initial training, but increased linearly during the race training period from 9.2% on day 90 to 13.6% on day 150 (Figure 26). Damage increased throughout the entire training program and exhibited a highly linear relationship with time at the stress fracture site (Figure 23). For the initial training period, the rate of increase in damage is greater at the caudal diaphysis, but this rate decreased once the race training period was initiated and becomes similar to the rate of damage formation seen at the stress fracture site (Figure 23).

Figure 23. Damage (mm/mm²) recorded every 10 days for 160 days of training at the stress fracture site and caudal diaphysis during Race Program 1.

Race Program 2

Along with Race Program 1, Race Program 2 experienced a constant increase in damage at both the caudal diaphysis and stress fracture site (Figure 24). Both race programs have almost identical results for porosity at both the caudal diaphysis and the stress fracture site (Figure 26). BMU activation frequency at the stress fracture site for Race Program 2 exhibited an identical pattern to Race Program 1 (Figure 25). Bone damage experienced differences at the caudal diaphysis between the race programs on the 90th, 120th, and 150th days, where on average, Race Program 2 experienced lower damage compared to Race Program 1 (Figure 28). An opposing trend was seen at the stress fracture site, where minimal differentials in damage occurred on the 90th, 120th, and 150th days (Figure 27). On average, BMU activation frequency was higher at the caudal diaphysis (Figure 25) for Race Program 1 when compared to Race Program 2 with the

highest differential of 0.006567 BMUs/mm²/day. At the end of both race programs, Race Program 2 produced the lowest damage at both sites, but not by a large margin.

Figure 24. Damage (mm/mm²) recorded every 10 days for 160 days of training at the stress fracture site and caudal diaphysis during Race Program 2.

Figure 25. BMU activation frequency (BMUs/mm²/day) recorded on the 90th, 120th, and 150th days of training at the stress fracture site and caudal diaphysis during Race Program 1 and 2.

Figure 26. Porosity (%) recorded on the 90th, 120th, and 150th days of training at the stress fracture site and caudal diaphysis during Race Program 1 and 2.

Figure 27. Damage (mm/mm²) recorded on the 90th, 120th, and 150th days of training at the stress fracture site during Race Program 1 and 2.

Figure 28. Damage (mm/mm²) recorded on the 90th, 120th, and 150th days of training at the caudal diaphysis during Race Program 1 and 2.

Effects of Turf and Synthetic Tracks on Bone Remodeling

BMU activation frequency, porosity, and damage at both locations on the humerus were all impacted from the differences in track surface type. The synthetic track surface had the largest effect on the stress fracture site and caudal diaphysis for all training programs when compared to the dirt track surface. The turf track displayed noticeable changes in bone remodeling at both sites for every program except for porosity at the stress fracture site for the race programs. Both the turf and synthetic track surfaces proved to reduce bone damage at both sites for all programs when compared to dirt, but the synthetic track surface accomplished the most reduction in damage (Figures 29-34). Damage at the stress fracture site during layup for the Baseline Layup and Long Layup training programs went below equilibrium damage values, which explains the negative percent change in damage for Figures 29 and 31.

Figure 29. Percent increase in damage over the equilibrium value (stress fracture site = 0.0299 mm/mm^2) recorded every 10 days for 350 days at the stress fracture site during the Baseline Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Figure 30. Percent increase in damage over the equilibrium value (caudal diaphysis = 0.1003 mm/mm^2) recorded every 10 days for 350 days at the caudal diaphysis during the Baseline Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Figure 31. Percent increase in damage over the equilibrium value (stress fracture site = 0.0299 mm/mm^2) recorded every 10 days for 350 days at the stress fracture site during the Long Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Figure 32. Percent increase in damage over the equilibrium value (caudal diaphysis = 0.1003 mm/mm^2) recorded every 10 days for 350 days at the caudal diaphysis during the Long Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Figure 33. Damage (mm/mm^2) recorded every 10 days for 160 days of training at the stress fracture site and caudal diaphysis during Race Program 1 for dirt, turf, and synthetic track surfaces.

Figure 34. Damage (mm/mm^2) recorded every 10 days for 160 days of training at the stress fracture site and caudal diaphysis during Race Program 2 for dirt, turf, and synthetic track surfaces.

Baseline Layup Training Program

BMU activation frequency for the turf track surface at the stress fracture site was slightly higher at days 70, 100, 210, and 280 when compared to the dirt track surface (Figure 35). The synthetic track contributed to a much greater BMU activation frequency at the stress fracture site at the same four points in time, which represent four distinct peaks (Figure 35). These four peaks are reflective of the layup period during the Long Layup training program from Ferullo (2007), where disuse contributed to oscillations in BMU activation frequency. The same result occurred in Hazelwood et al.'s (2001) study of disuse, where smaller loads contributed to longer durations in BMU activation frequency oscillations. There was an opposite trend at the caudal diaphysis where the turf track contributed to a lower BMU activation frequency throughout the entire program. Again, the synthetic track contributed to the most extreme value when compared to the dirt track by causing the caudal diaphysis to have the lowest BMU activation frequency among track surfaces (Figure 36).

Porosity for the turf track surface was distinctly greater from the 120th day to the 170th day of training at the stress fracture site when compared to the dirt track surface (Figure 37). Throughout the entire training program, the synthetic track contributed to an increased porosity compared to the both the turf and dirt track surfaces at the stress fracture site, where a porosity of 38% was attained at day 130 with the synthetic track surface (Figure 37). Porosities for the dirt and turf track surfaces were 28.2% and 30%, respectively. An opposing trend was seen at the caudal diaphysis where the synthetic track surface displayed the lowest porosity and the dirt track surface had the greatest

porosity (Figure 38). The turf track surface only had a slight reduction in porosity when compared to the dirt track surface (Figure 38).

Figure 35. BMU activation frequency (BMUs/mm²/day) recorded every 10 days for 350 days at the stress fracture site during the Baseline Layup training program for dirt (Ferullo, 2007), turf and synthetic track surfaces.

Figure 36. BMU activation frequency (BMUs/mm²/day) recorded every 10 days for 350 days at the caudal diaphysis during the Baseline Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Figure 37. Porosity (%) recorded every 10 days for 350 days at the stress fracture site during the Baseline Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Figure 38. Porosity (%) recorded every 10 days for 350 days at the caudal diaphysis during the Baseline Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Long Layup Training Program

BMU activation frequency for the turf track surface at the stress fracture site had noticeable increases on days 70, 100, and 180 of training when compared to the dirt track surface (Figure 39). BMU activation frequency experienced a larger increase for the synthetic track when compared to the other track surfaces for the same days of training. Most notable differentials in BMU activation frequency between the dirt and synthetic track surfaces occurred on the 70th, 100th, and 170th days of training, which are represented by three peaks (Figure 39) and reflective of results seen in Ferullo (2007) and Hazelwood et al. (2001), where smaller loads contribute to oscillations in BMU activation frequency. Unlike the stress fracture site, the caudal diaphysis experienced lower values of BMU activation frequency for both the turf and synthetic track when compared to the dirt track, where the synthetic track surface had the lowest BMU activation frequency (Figure 40).

Porosity at the stress fracture site experienced an increase for the turf track surface compared to the dirt track surface from the 120th day of long layup to the end of training (Figure 41). The synthetic track surface contributed to an even greater increase in porosity at the stress fracture site from day 20 to the end of the training program (Figure 41). Opposite to the trend seen at the stress fracture site, the turf and synthetic tracks had reductions in porosity at the caudal diaphysis when compared to the dirt track surface (Figure 42). The synthetic track surface had the lowest porosity at the caudal diaphysis (Figure 42).

Figure 39. BMU activation frequency (BMUs/mm²/day) recorded every 10 days for 350 days at the stress fracture site during the Long Layup training program for dirt (Ferullo, 2007), turf and synthetic track surfaces.

Figure 40. BMU activation frequency (BMUs/mm²/day) recorded every 10 days for 350 days at the caudal diaphysis during the Long Layup training program for dirt (Ferullo, 2007), turf and synthetic track surfaces.

Figure 41. Porosity (%) recorded every 10 days for 350 days at the stress fracture site during the Long Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Figure 42. Porosity (%) recorded every 10 days for 350 days at the caudal diaphysis during the Long Layup training program for dirt (Ferullo, 2007), turf, and synthetic track surfaces.

Race Programs

BMU activation frequency at the stress fracture site for the turf track surface increased most noticeably on the 70th and 140th days of training when compared to the dirt track surface (Figures 43-44). The synthetic track surface had the most drastic increase on the same two days, which represent two large peaks (Figures 43-44) and are the result of smaller loads from the synthetic track surface contributing to oscillations in BMU activation frequency. The values for BMU activation frequency for dirt, turf, and synthetic track surfaces during Race Program 1 at the stress fracture site on 140th day were 0.0142 BMUs/mm²/day, 0.0282, BMUs/mm²/day and 0.2 BMUs/mm²/day, respectively. BMU activation frequency at the caudal diaphysis was reduced when compared to the dirt track surface for the turf and synthetic track surfaces, where the synthetic track had the lowest BMU activation frequency (Figures 43-44).

Porosity at the stress fracture site for the turf track surface experienced minimal differences in comparison to the dirt track surface porosity values (Figure 45-46). The synthetic track surface produced an increase in porosity at the stress fracture site compared to the other track surfaces (Figures 45-46). Porosity at the caudal diaphysis for the turf and synthetic track surfaces was reduced in comparison to the dirt track surface, where the synthetic track surface had the largest reduction in porosity (Figures 45-46).

Figure 43. BMU Activation Frequency (BMUs/mm²/day) recorded every 10 days for 160 days at the stress fracture site and caudal diaphysis during Race Program 1 for dirt, turf, and synthetic track surfaces.

Figure 44. BMU Activation Frequency (BMUs/mm²/day) recorded every 10 days for 160 days at the stress fracture site and caudal diaphysis during Race Program 2 for dirt, turf, and synthetic track surfaces.

Figure 45. Porosity (%) recorded every 10 days for 160 days at the stress fracture site and caudal diaphysis during Race Program 1 for dirt, turf, and synthetic track surfaces.

Figure 46. Porosity (%) recorded every 10 days for 160 days at the stress fracture site and caudal diaphysis during Race Program 2 for dirt, turf, and synthetic track surfaces.

IV. Discussion

Bone Response and Remodeling

Equilibrium

Prior to the commencement of training, the mechanical stimulus (Φ) at the caudal diaphysis of 1.0E-9 cycles per day was greater than the stress fracture site's mechanical stimulus of 2.1E-10 cycles per day (Ferullo, 2007). Disuse is defined as $\Phi < \Phi_0$ and Φ_0 is 1.875E-10 cycles per day (Hazelwood et al., 2001), which is close to the stress fracture sites mechanical stimulus. If an area of bone is in a state of disuse, bone has a faster removal rate than formation rate, so bone density is decreased. Disuse activates remodeling, which adds haversian canals in cortical bone making it difficult to increase bone density (Hazelwood et al., 2001). During equilibrium, the stress fracture site becomes susceptible to a complete fracture due to low bone density and reduced stiffness. If the horse were to immediately commence a rigorous work out, such as race training, after equilibrium, there would be an increased likeliness of stress fracture.

Initial Training

During the commencement of initial training, damage increased for all training programs and at both sites due to an increase in mechanical stimulus, Φ . Throughout every training program, damage is substantially greater at the caudal diaphysis compared to the stress fracture site, but the stress fracture site has a higher rate of percent increase in damage. BMU activation frequency at the commencement of initial training is considerably different between both regions, where on average; the caudal diaphysis has a higher BMU activation frequency. Extensive damage causes bone cells to increase activity (BMU activation frequency), which is caused by a larger mechanical stimulus.

There is insignificant apposition or deposition of osteoid at the stress fracture site since porosity remains constant. However, there is a slight increase in porosity at the caudal diaphysis due to an increasing resorption rate. This increase in porosity is accomplished through the removal of bone by osteoclasts, which have contributed to the rate of resorption exceeding the deposition of bone. The apparent difference between mechanical stimuli is a key factor in dissimilarities between state variables at both regions.

Layup

Training programs with a period of layup due to injury displayed an immense spike in BMU activation frequency during the onset of layup at the stress fracture site. This immense initiation of remodeling is the result of the region undergoing disuse since mechanical stimulus drops to $6.9E-11$ cycles per day (Ferullo, 2007), which is below the initial mechanical stimulus. A state of disuse at the stress fracture site means bone density is being removed, so porosity during this period is drastically increased. During this period of bone removal at the stress fracture site, damaged and unhealthy bone is resorbed, thus decreasing damage. BMU activation frequency at the caudal diaphysis decreases due to a lower mechanical stimulus, which causes less damage. Porosity increases slightly due to damage removal, but not as evident as the stress fracture site.

During long layup, both regions try to return to equilibrium conditions since during the later portion of layup there is a lower spike of porosity, BMU activation frequency, and damage at the stress fracture site. This would formulate the assumption that if layup were to continue longer, consecutive spikes would gradually decrease in

magnitude and oscillate around a common value as state variables did in the equilibrium case.

Post Layup

Immediately following layup, there is a dangerous rapid increase in damage at both regions in the humerus. BMU activation frequency is constant at the stress fracture site, since disuse is no longer present in this area, reducing remodeling activity. Porosity decreased due to less osteoclastic activity and flattens out to become constant. At the stress fracture site, a lower BMU activation frequency with a constant porosity indicates an area of profound damage accumulation with a growth of further microcracks and unhealthy bone. While there is more damage seen at the caudal diaphysis, there is an immediate cellular response to remove poor bone tissue due to a greater mechanical stimulus, which results in a gradual increase in porosity (unhealthy bone being removed) and a reduction in rate of damage accumulation. The caudal diaphysis accomplishes removal of damaged bone tissue, while the stress fracture site only accumulates more damage.

Race Programs

Race programs achieved almost identical trends for BMU activation frequency, damage, and porosity at the stress fracture site and the caudal diaphysis. However, key differences occurred between the stress fracture site and caudal diaphysis. The stress fracture site remains at approximately a constant porosity throughout the entire program, which is an indicator that insignificant cellular remodeling is taking place compared to the caudal diaphysis region. Without the proper removal of damaged bone and minimal cellular activity or mechanical stimulus, damage continues to rise throughout the rest of

training (Figures 23-24). As seen in previous training programs, the caudal diaphysis has a high enough mechanical stimulus or load to initiate remodeling, where microdamage is removed and replaced with new osteoid.

Although there is minimal BMU activation frequency at the stress fracture site compared to the caudal diaphysis throughout both race programs (Figure 25), there are noticeable changes in BMU activation frequency on days 70 and 140, especially under turf and synthetic track conditions (Figures 43-44). Towards the completion of initial training, BMU activation frequency at the caudal diaphysis increases at a greater rate. If training were to continue for a longer period of time, microdamage would continue to develop and eventually be removed resulting in a higher porosity at the caudal diaphysis.

The period of race training generates a greater mechanical stimulus, which eventually leads to the removal of bone creating a higher porosity. This is more pronounced at the caudal diaphysis since it responds more immediately to race training than the stress fracture site. Even though damage stays at a constant increase throughout the entire program at the stress fracture site, it is possible this rate would begin to decrease as BMU activation frequency and porosity increase. To further investigate cellular effects at both regions, the frequency output of state variables damage, BMU activation frequency, and porosity could be increased from every 10 days to every 5 days enabling a more detailed and further analysis. The length of each training program could also be increased for more in depth research.

Race Programs (Hypothesis 2)

Race programs only have one day of layup per week, so there is no extended duration of layup. An absence of layup generates a growth in microdamage at the both regions of interest. The race programs were not compared to the Baseline Layup and Long Layup training programs since they contain a period of layup, which develop completely different results. On average, Race Program 1 experienced more damage than Race Program 2, which supports the second hypothesis. This was most apparent on days 90, 120, and 150 at the caudal diaphysis (Figure 28). During the transition from initial training to race training, Race Program 2 has a week off from fast gallop and is replaced with 6 days of slow gallop and a day of layup, which gives the humerus time to remove bone damage at the caudal diaphysis. Race Program 1 for the same time period has 5 days of slow gallop, a race day, and a day of layup, which reduces the amount of damage removed from the humerus during this time. The transition from 8 weeks of initial training to race training is the key to lower damage at the caudal diaphysis for Race Program 2 since this is the time period with the largest absence of a race (9 weeks). Race Program 1 only has 8 weeks of race absence before the first race (Tables 15-16). Due to the similar workout schedule for the race programs, it was difficult to note differences in bone remodeling at both sites of interest.

Even though there were insignificant differences in damage at the stress fracture site, there was a highly positive correlation (Figures 47 and 48) between day of training and percent increase in damage at the stress fracture site. This was possible to predict through previous literature (Ferullo, 2007) since it was observed during a period of training for Baseline Layup and Long Layup programs that damage had a linear increase

at the stress fracture site. Additionally, the race programs do not give the stress fracture site a sufficient amount of time to remove bone microdamage.

Figures 47-48. A fitted line plot observing the correlation between step (4 steps in a day) and increase in damage at the stress fracture site for Race Program 1 (top) and Race Program 2 (bottom) with an R-Squared value (Coefficient of Determination) of 100%. Meaning 100% of the variation in percent increase in damage can be explained by day of training.

It is difficult to predict if a significant amount of microdamage will be removed at the stress fracture site if the race programs are extended. Increases in microdamage are correlated with stress fractures (Burr and Milgrom, 2001) and incorporating a period of layup into these two programs would lower the possibility of stress fracture or injury. There is an enigma incorporating a period of layup into a training program at the stress fracture site since during layup, bone density is removed and the bone resorbing rate exceeds the bone refilling rate causing the bone to become more porous and susceptible to fracture. Although microdamage is removed, there must be a compromise between layup and training intensity so the stress fracture site doesn't enter a state of disuse or endure an excessive amount of damage leading to stress fracture. Further investigation and testing is still needed to determine if layup, training, or a combination has the greatest influence on remodeling at the stress fracture site.

Turf and Synthetic Track Surfaces (Hypothesis 3)

Reduced ground reaction forces caused by turf and synthetic track surfaces contributed to a decrease in muscle and contact forces. Dirt track surfaces resulted in the highest ground reaction forces, while synthetic track surfaces contributed to the lowest ground reaction forces. It was hypothesized that lower muscle and contact forces on the humerus would be responsible for a decrease in microdamage at both regions on the humerus including all training programs. This hypothesis was confirmed, but the decrease in damage was the result of two different pathways between the caudal diaphysis and stress fracture site. Damage decreased at both sites and for all training programs as ground reaction forces were decreased. The synthetic track surface

contributed to the highest porosity, BMU activation frequency, and lowest damage at the stress fracture site for all training programs, but carried out the lowest porosity, BMU activation frequency, and damage at the caudal diaphysis.

Decreasing loads reduce the amount of damage done to the bone, which can lead to two different bone remodeling events. Both porosity and BMU activation frequency can be increased, in the case of disuse, or they can both be decreased, in a non-disuse state. When the muscle and contact loads were reduced from dirt track to turf and synthetic track loads, it was apparent that the stress fracture site was behaving as an area in disuse, while the caudal diaphysis was a bone not in disuse, but undergoing less damage accumulation. The large spikes in BMU activation frequency at the stress fracture site seen in Figures 35, 39, 43, and 44 are the result of a drastic drop in mechanical stimulus. When mechanical stimulus drops significantly, BMU activation frequency increases exponentially. Multiple peaks in Figures 35, 39, 43, and 44 are reflective of disuse in Hazelwood et al.'s (2001) results, where a decreasing load (small enough to initiate disuse) resulted in a series of BMU activation frequency spikes or oscillations. When the humerus experiences a severe disuse situation, such as the synthetic track case with respect to the other track surfaces, BMU activation frequency at the stress fracture site oscillates for a longer period of time before returning to normal conditions (equilibrium) (Hazelwood et al., 2001). Decrease in damage at the caudal diaphysis can be explained by lowering the damage formation rate while maintaining the damage removal rate. It is possible for bone disuse and overload to occur simultaneously at different regions (Hazelwood et al., 2001) and has been demonstrated in this study at the two sites with the scaling down of muscle and contact forces due to track surfaces.

Although the synthetic and turf track surfaces successfully reduced the amount of damage seen at the stress fracture site and caudal diaphysis, porosity increased for all training programs at the stress fracture site except under the conditions of a turf track during the race programs. If a trainer wanted to maintain bone density within the horse, but also wanted to reduce damage, then the turf track surface would be the ideal choice. A horse experiencing a lot of bone damage, but still healthy enough to train could be placed on the synthetic track surface since it contributed to the lowest ground reaction forces and consequently, reducing damage to the greatest extent compared to the other track surfaces. While an increase in porosity implies the removal of damaged bone, it can be dangerous to a horse by reducing the elastic modulus of the bone, inducing susceptibility to deformation. Training a horse initially on a synthetic or turf track and then switching them to race on a dirt track might be a hazardous transition since the horse's bones would be more porous and have a lower elastic modulus initially, which could lead to extensive damage from higher loads. It would be interesting to model bone activity for a training program which alternated between track surface types since race horses frequently train and race on different surfaces (Setterbo et al., 2007). Having a Thoroughbred train on a dirt track surface and race on a synthetic surface could be one possible scenario for a simulation in future updates to the current model.

Past literature has shown contradictory results relating the risk of injury with track surface type (Setterbo et al., 2007). It was found that there was no significant effect between all-weather and turf training surfaces for risk of fracture (Verheyen, 2005). Some argue turf and synthetic tracks are less hazardous for race horses compared to dirt tracks, while others claim there is no statistical difference. Past research has

demonstrated the effects of track surface material properties on hoof impact accelerations and ground reaction forces (Setterbo et al., 2007). It is possible that upper forelimbs may be less affected by track surface properties compared to lower forelimbs since the lower limbs are closer to impact from ground reaction forces and the distance is shorter for energy dissipation. However, bone is a stiff material and efficient in the transfer of energy with minimal deformation. The current study demonstrated the benefit of using the softest track surface (synthetic) by reducing the amount of damage done to bone at both regions on the humerus for all training programs.

Assumptions had to be incorporated to recalculate muscle and contact forces, which may have contributed to minor inaccuracies. It was assumed that trot vertical ground reaction forces from Setterbo et al. (2007) for all track surfaces were directly proportional to standing and walking forces used in the current study. Although, the standing load case would exhibit the same vertical ground reaction forces, regardless of surface type, it was updated since it simulates a quarter of a day of training. Ground reaction forces change during a dynamic impact case due to energy dissipation and deformation differences between materials. The percent decrease in ground reaction forces used from the Setterbo et al. (2007) study contained variations between individual race horses. Also, horses did not run at racing speeds to minimize fatigue. Increasing the number of horses used in the study, testing at different race courses, and using various instruments for recording would generate more statistically significant results, which could be used to determine more accurate muscle and contact forces on the humerus.

Limitations and Future Updates

The model used in the current study is limited through many factors mainly relating to assumptions and a lack of preexisting data. Equine EMG data only indicated if the muscle was active or not, while canine EMG data specified the amount of activity, which determined the amount of muscle force. Further research into equine EMG data would assist in developing more accurate muscle force activation patterns. It would enable the confirmation of inactive and active muscle forces during each phase of galloping. Muscle and contact forces were scaled up from a standing load case to simulate slow and fast galloping speeds. The scaling factor was based on a ratio of ground reaction forces during walking and galloping activities. It is difficult to measure race ground reaction forces at racing speeds, so these estimates of ground reaction forces could result in inaccuracies at contact and muscle forces, and alter remodeling activities (Ferullo, 2007).

Within the remodeling algorithm used in this study, the principle strain with the highest magnitude determined the mechanical stimulus, which governed the response of bone cells throughout the entire humerus model. Realistically, there are a variety of factors, other than strain, that contribute to remodeling such as chemical or age factors. Like humans, remodeling activity diminishes as the horse ages. Also within the remodeling algorithm, human femur remodeling initial conditions and constants were used. Replacing human remodeling conditions with a horse's, such as plexiform bone geometry, would improve the accuracy of the model as well. A more conclusive bone response would be observed for state variables damage, BMU activation frequency, and

porosity if equine remodeling conditions and other environmental factors were included in the model.

Even though the current model has its limitations and drawbacks, it provides a framework for investigative evidence of stress fractures occurring in racing Thoroughbreds due to a variety of influences. All material used in this study is sufficient enough to generate a model through which there is a general understanding of the influence of muscle and contact forces, ground reaction forces, track surfaces, and training regimes on bone remodeling in the humerus. There are many areas in the model where factors affecting remodeling can be improved on or further investigated. For example, all muscle and contact forces used in the current model are estimates based on in vitro testing, mathematical models, and even other animals. Other forces were even further adjusted to match a physiologic porosity distribution seen in radiographs (Ferullo, 2007). Testing all muscle and contact forces on the humerus in vivo would be ideal, yet highly complicated.

Analyzing additional locations on the humerus can establish new comparisons between the stress fracture site and caudal diaphysis. Also, an algorithm incorporating pharmaceutical simulations, such as effects from biphosphonates, could be implemented to investigate the bone remodeling response within the humerus. Biphosphonates promote the formation of new bone by osteoblasts, while blocking the destruction of bone by osteoclasts preventing the loss of bone mass. Biphosphonates could be a factor in assisting fracture prevention in race horses. It is certain new training programs and track surfaces will continue to be incorporated into the current model to further investigate the effects of equine exercise on stress fractures.

Project Summary

The purpose of this project was to simulate and investigate bone remodeling response in a 2-3 year old Thoroughbred racing horse humerus using a finite element model in conjunction with a remodeling algorithm. Modeling was not accounted for in this model, even though horses who are 2 years of age are still undergoing modeling and are equivalent to the human adolescent stage (Ferullo, 2007). We updated an existing finite element model by incorporating muscle and contact forces for a turf and synthetic track surface for all load cases. Two entirely new training programs (race programs) were also implemented for further investigation into additional training regimes. An in depth analysis of each training regime will aid in the selection of a suitable program for stress fracture prevention. The equine humerus is an excellent model for researching stress fractures due to the characteristic location frequently resulting in damage combined with the abundant supply of injured humeri. Not only is this study beneficial to Thoroughbred race horses, but this model can be used as a tool for further studies in human or animal athletes who are susceptible to stress fractures.

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