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ADVANCES IN PEDIATRIC RESEARCH

Modeling and analysis of proximal tibial growth plate fractures in adolescents: Theory and potential applications

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Abstract

Background: Overuse injuries in children and adolescents are becoming increasingly common, particularly in those who regularly participate in a single sport. As a result, prevention, early detection and treatment of these injuries is vital. However, existing research in adult populations cannot always be directly applied to analogous cases in younger populations. This study attempts to provide an example of how both mathematical and computer modeling can be utilized to predict alterations in load locations, directions, and magnitudes resulting from maturational changes in a way not possible in vivo.

Methods: A 2D leg extension model was created and used to calculate relevant forces at the proximal knee joint. Individual aspects of the model, such as quadriceps force and leg length, were changed to quantify how increases in a growing adolescent's force generation and limb length may affect the forces at the joint. The derived forces were input into a 3D finite element model incorporating a growing young adult's relatively weaker epiphyseal plate material to calculate the stresses and strains on the tibia of an adolescent.

Results: Findings indicated that a shortened patellar tendon and increased quadriceps muscle strength were potentially greater contributors to increased stress on the proximal tibia, as opposed to aspects such as height and weight changes.

Conclusions: The theoretical and computational methods employed show promise in their ability to predict potential injury risks in populations for whom evidence-based research is lacking. Models incorporating the elbow and shoulder have high impact potential for young baseball pitchers.

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Introduction

Current estimates assert that 30 to 45 million American children participate in at least one organized athletic activity. Every year, about one third of these young people will experience an injury necessitating a visit to a medical professional, resulting in yearly healthcare costs of up to $1.8 billion [1]. Overuse – repetitive micro-trauma to a tendon, bone, or joint without adequate regeneration or healing time – constitutes up to half of these injuries [2].

Today, children are participating in more competitive leagues and specializing in a particular sport earlier than ever before. Many not only participate in one sport year-round, they also play and train with several different clubs at a time. In some cases, athletes as young as 11 or 12 years of age train for their sport for upwards of 27 hours a week [3]. Training loads that are so heavy in duration and frequency have the potential to result in injury, particularly when the athlete in question may not be physiologically, mechanically, and/or neurologically mature enough to perform at such an intense level.

Both mathematical and computer modeling can be employed to estimate how singular aspects of growth (e.g. changes in biomaterial properties, or the change in length of a single physiological parameter) alter stresses, strains, and their locations. Such models can provide additional insight for the prediction and care of injuries, such as physeal fractures, in adolescents. In addition, knowledge of the possible locations of fracture initiation and propagation has important implications for diagnosis and treatment. Two of the major factors contributing to physeal injuries are the mechanical properties of the growth plate itself, and the forces experienced by the physeal cartilage and the surrounding bone. While the elbow and shoulder have been areas of increasing scrutiny, particularly as pertains to young baseball pitchers [4–6], the lower extremity remains less studied.

The goal of this study was to examine potential contributing factors to injuries at the proximal tibia in adolescents by developing a two-dimensional mathematical model to examine changes in force directions and magnitudes during an isometric knee extension. The derived forces were subsequently utilized in a finite element analysis (FEA) of the tibia in order to quantify and visualize differences in location and peak values of stress in the adolescent tibia, particularly in the growth plate region. Proposed factors in overuse injuries in adolescents guided the choice of aspects modeled, and include increased weight, stretched or less flexible ligaments, and increased power generation.

Methods

Leg extension model

A basic two-dimensional, sagittal plane mathematical model of a leg extension was developed to provide a measure of the change in relevant forces at the proximal tibia for use in a finite element model (Fig. 1). Forces included were femorotibial shear and normal forces (FNA1, FNA2), patellar ligament tensile force (FAB), dynamometer force acting at the distal end of the tibia (F_D), and the weight of the foot (FOOT). Male values were chosen for all of the analyses since males typically experience more physeal fractures, and grow more in absolute terms and at a faster rate than girls during puberty [7].

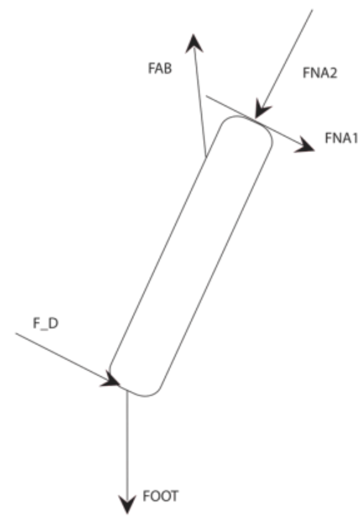

Figure 1. Free body diagram of forces acting on the leg during a knee extension movement

FNA1: femorotibial shear; FNA2: normal forces; FAB: patellar ligament tensile force; F_D: dynamometer force acting at the distal end of the tibia; FOOT: weight of the foot.

Autolev codes were written in order to make comparisons of forces by altering a single anatomic factor hypothesized to increase the risk of tibial fracture. The Autolev™ (OnLine Dynamics, Inc. Sunnyvale, CA, USA) software package was used to generate code for Matlab™ (The MathWorks, Inc. Natick, MA) equations of motion, and create graphs and tables. Codes were written for a 13-year-old male in the 50th percentile of height and weight, and with average dynamometer force. Single parameters of the model were changed as follows (Table 1):

a) Weight in the 90th percentile
b) Height in the 90th percentile
c) With a patellar tendon length three-quarters that of the baseline model
d) With a 75% increase in dynamometer force

Table 1. Values for the factor comparison. Models were defined by the alteration from the baseline model.

Model	Height (m)	Mass (kg)	Force (N)	Patellar ligament length (cm)
Baseline	1.56	46	250	2.94, 2.29, 13.7
Patellar ligament				2.205, 1.72, 10
Force			325	
Weight		60		
Height	1.75			

Percentile height and weight values were obtained from United States census data via the Center for Disease Control [8]. The equations used to calculate center of mass and segment length are derived from equations developed by Jensen et al [9]. The averages of a cross-sectional study of the size and strength of the lower leg muscles during growth were used as inputs for the dynamometer force, body weight [10].

Finite element analysis

The finite element model used in the simulations was based on a tibial part contributed to the open source model database. The basic geometry was imported into the FE software package Abaqus™ (HKS, Pawtucket, RI), and the part was partitioned in order to refine the mesh size, as well as to delineate sections for different material properties. The tibia was 35 cm long, translating to a height of approximately five feet, consistent with a typical pre or early adolescent height. The tibial tray measured 8 cm mediolaterally, and 6 cm anteroposteriorly.

Material properties

Three materials were defined and assigned to sections of the adolescent model: cortical bone, cancellous bone, and growth plate cartilage (Table 2). Values for the Young's Modulus, Poisson's ratio, and shear modulus were determined from a literature survey of mechanical testing, and computational studies attempting to describe the properties of bone and cartilage [11–14]. The main model was created with an approximately 8 mm growth plate partition between two areas of cancellous bone at the proximal end of the tibia. A second model was given upper range strength constants, and was assumed to consist entirely of cortical and cancellous bone.

Table 2. Young's and shear moduli used in Abaqus simulation. Values are in MPa.

Model 1 *(adolescent)*	*Cortical bone (MPa)*	*Cancellous bone (MPa)*
E_1	9000	600
E_2	9000	600
E_3	14000	2000
G_{12}	3500	200
G_{13}	5000	300
G_{23}	5000	300
Model 2	*Cortical bone (MPa)*	*Cancellous bone (MPa)*
E_1	14000	800
E_2	14000	800
E_3	20000	3000
G_{12}	6500	300
G_{13}	8000	400
G_{23}	8000	400

Loading conditions

Abaqus was used to simulate a quasi-static loading of the tibia. Since it is a speed more applicable to athletic activities, an average of 5.24 radians (300°) per second was used, resulting in a total load duration of 300 msec. The load values for the tangential and normal components of the patellar tendon force were derived and input separately to account for the fact that the patellar tendon insertion angle changes as the lower leg flexes and extends [15]. Just under 3000 linear tetrahedral (C3D4) elements were generated on the mesh of the tibia. Field output was requested at 10 evenly spaced intervals, and included all stress and strain components.

Results

Factor comparison

The individual factor comparison results provide some clarification as to which aspects of growth may play a leading role in the overloading of the physeal region. First, none of the aspects diverged greatly when shear force was examined (Fig. 2).

However, the increased dynamometer force and shortened patellar tendon models produced similar magnitudes and resulted in the most significant increases for both femorotibial contact and patellar tendon forces. The largest changes in loading occurred at the patellar tendon insertion, increasing by 33% and 78% for slow and medium speeds, respectively (Fig. 3).

This supports the idea that the force at the proximal tibia is increased when the patellar tendon acts from a stretched initial length, and therefore at an increased preload. This decreased range before the tendon reaches the limit of its slack means the tibia experiences a greater tensile force for a greater period of time. If this is indeed a root cause, it suggests that issues such as Osgood-Schlatter's disease could be alleviated through stretching programs, particularly before bouts of physical activity. It is also important to realize that though differences may be moderate, these differences can accumulate when performing a movement with frequency. For example, an athlete may execute a soccer kick dozens of times a practice session, four or five times a week. These types of force estimations would be useful in risk analysis, using a load/cycles-to-failure model.

Changes in weight contribute the least to the forces at the proximal tibia. The idea of weight as a minor factor is consistent with clinical experience of the knee [16], and divergent from the findings of finite element analysis of the spine [17]. This emphasizes that etiology and mechanism of failure in different bones, and even at proximal versus distal ends of the same bone, may be different, and as such need to be studied individually.

Finite element analysis

In both the adult and adolescent models of the tibia with dynamometer loading, the largest stress values occurred in the posterior portion (Fig. 4), with the stress values slightly higher in the stronger material model. However, because the ultimate strength of adolescent bone is less than that of adults, and less resilient than that of children, it will reach the damage threshold more quickly, and therefore will be able to absorb less energy before fracturing [18–20]. The locations of these stresses are of particular note, as fractures initiated in, or traversing through the posterior aspect of tibia, have the potential to damage the popliteal artery, which provides necessary nutrients for growth in the physeal plate and the surrounding bone.

The strain profiles for the two models are less similar, both in value and location. Maximum strains were 0.040 and 0.035 for the weaker/adolescent and stronger/adult models, respectively. In addition, the strain is localized in the growth plate region in the weaker material model, whereas there is a more gradual and spread out strain in the stronger material model. The principal stress is at a maximum on the anteromedial portion of the tibia. This location suggests a tendency towards Salter classification type II and IV fractures.

Discussion

Two of the major determining factors for physeal injuries are the mechanical properties of the growth plate itself, and the forces experienced by the physeal cartilage and the surrounding bone.

The present study involved movement in a single plane, and looked at the action of just one muscle–tendon unit acting across a joint. In addition, the finite element model included the tibia only, and was therefore unable to assimilate any possible attenuation of forces by the surrounding bones and structures. Future studies can build on the initial survey of factors presented here by adding degrees of complexity and specificity, particularly by expanding from 2D to 3D movements, such as cutting and jumping.

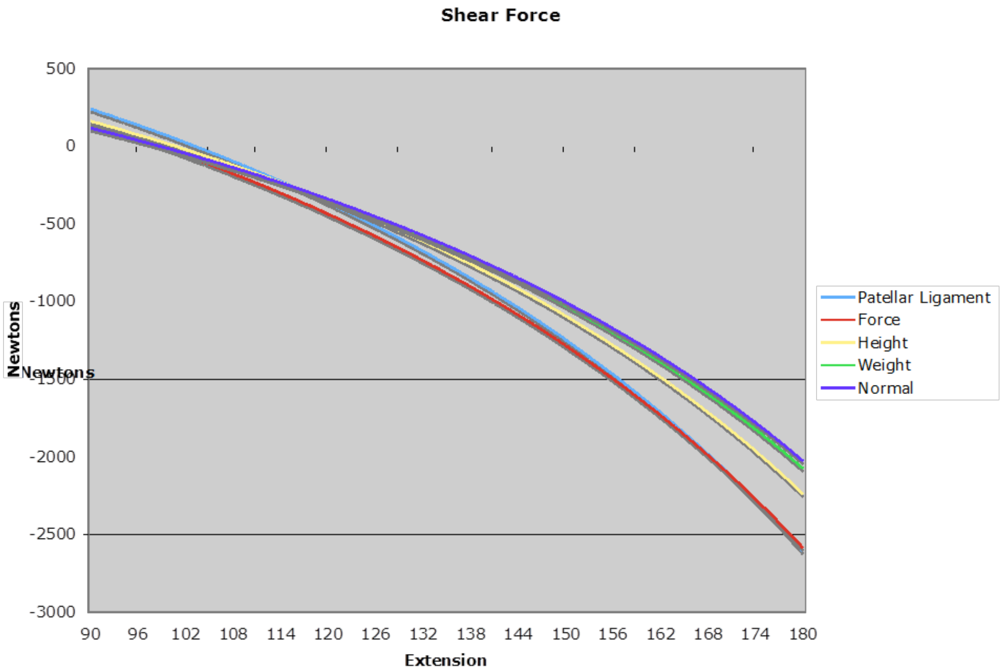

Figure 2. Plot of tibiofemoral shear force (N) throughout an isokinetic (180°/sec) leg extension for normal, increased weight, increased height, increased force, and shortened patellar ligament models.

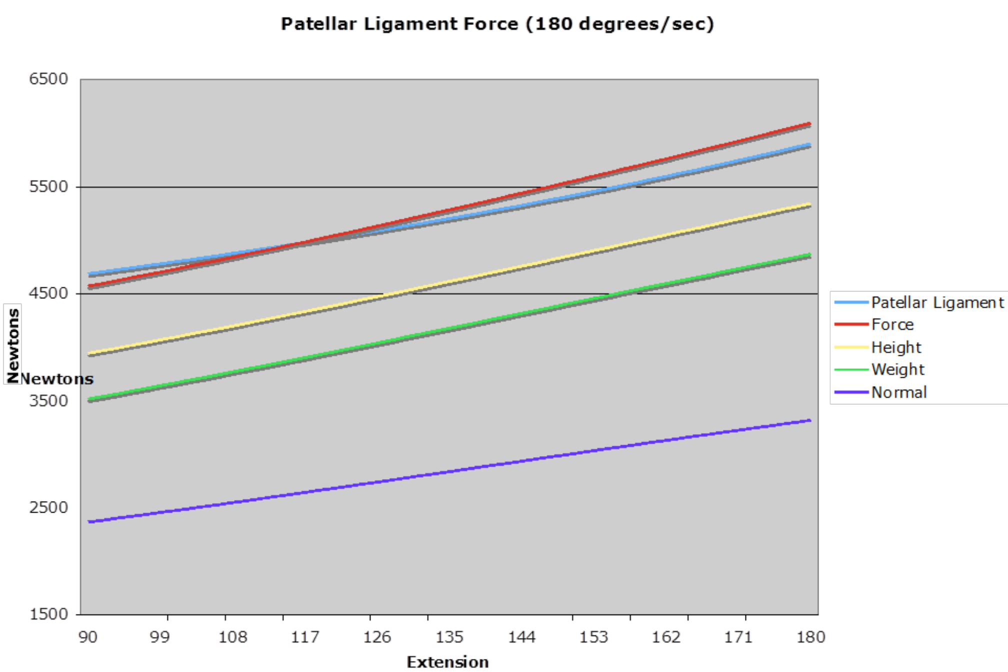

Figure 3. Plot of patellar tendon force (N) throughout an isokinetic (180°/sec) leg extension for normal, increased weight, increased height, increased force, and shortened patellar ligament models.

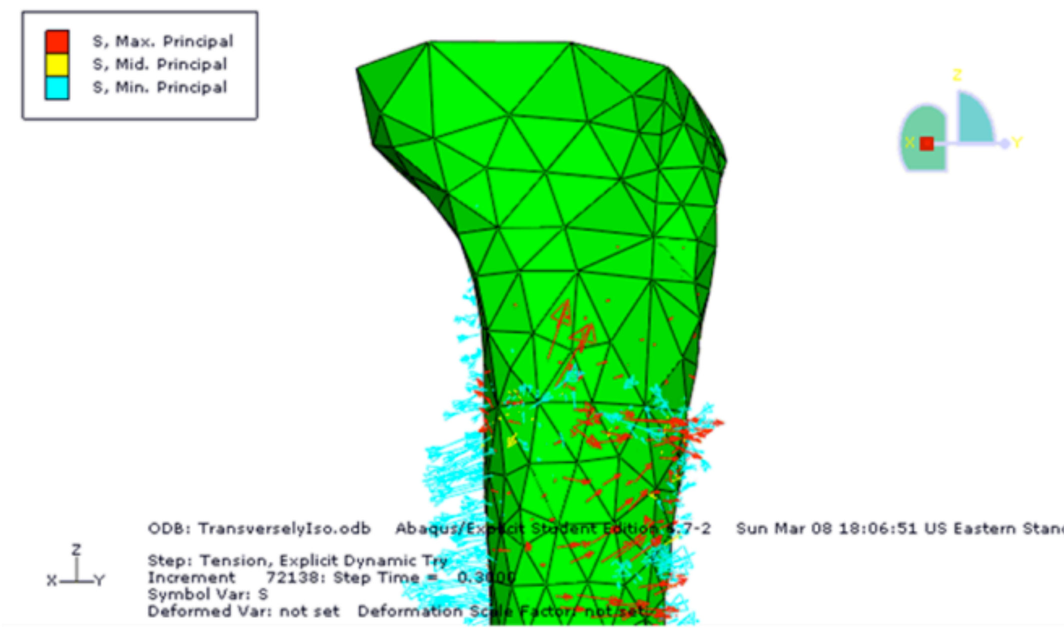

Figure 4. Vector representation of average maximum (red) and minimum (blue) principal stresses acting on the proximal tibia of an adolescent during an isokinetic (180°/sec) leg extension as calculated via finite element analysis.

These models would also need to include the actions of several muscles. In this simulation, loading values did not include the action of the hamstrings, which may well increase or decrease the stresses and strains. Inclusion of other soft tissues, the patella, and femur would also give a more holistic result of the stresses and interactions of the model. This could be accelerated through the use of CT and MRI scanning, with software capable of generating 3D images that are importable into finite element packages such as Abaqus. Such a model could then be scaled to correlate to individual adolescents, reducing the necessity of exposure to x-rays. The finite element model could be used as a supplement to the mathematical model, and use force and moment values obtained from that analysis to suggest which activities, when overdone, are more likely to lead to pain or injury, and which activities can provide a 'break' to the at-risk area.

For this simulation of a knee extension movement, the occurrence of physeal fractures at the proximal tibia was determined to be affected by: 1) increased force being transmitted through the patellar ligament, whether by increased muscle force, or because of a lag in the growth of the patellar ligament with respect to the tibia; 2) presence of a weaker growth plate that is both less stiff and less able to absorb energy. As a result, the growth plate and the surrounding area are more susceptible to microfracture, and less able to remodel in time to accommodate increased load. The major differences in stress and strain locations and variations in magnitude underline the need to develop more specific models for various sports, as well as better descriptions of the material properties of bone and growth plate, particularly in younger populations. This analysis was able to simulate dynamic, short loading rates, and may still be considered representative of the overall differences in loads and stresses in adults, children, and adolescents. Knowledge of the possible locations of fracture initiation and propagation has important implications for diagnosis and treatment. Both mathematical and computer modeling can be employed to quantify how loads, stresses, and strains change from childhood to adulthood, and in doing so, provide additional information for the prediction and care of physeal fractures in adolescents.

As athletes continue to specialize in sports and do so earlier in their development, the majority of the injuries in this population are likely to be the result of overuse [21]. Therefore, it will become increasingly important to improve our ability to effectively prevent and treat their unique injuries. The methods employed in this study allow the estimation of forces, stresses, and strains during a kicking motion at the proximal tibia, and can be expanded to include other joints and activities for both injury avoidance and rehabilitation. For example, there has been speculation that overuse and/or poor mechanics during youth are underlying the increasing number of ulnar collateral ligament reconstructions (Tommy John surgeries) necessary among major league baseball players. Models quantifying the stresses and strains produced at the elbow can aid in further refining rules limiting pitch counts in younger players, or help refine their mechanics, keeping them from incurring joint damage that will affect them later in their careers.

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References

1. Adirim T, Cheng T. Overview of injuries in the young athlete. Sports Med. 2003;33(1):75–81.
2. Dalton SE. Overuse injuries in adolescent athletes. Sports Med Auckl NZ. 1992;13(1):58–70.
3. Caine D, DiFiori J, Maffulli N. Physeal injuries in children's and youth sports: reasons for concern? Br J Sports Med. 2006;40(9):749–60.
4. Fleisig GS, Andrews JR, Cutter GR, Weber A, Loftice J, McMichael C, et al. Risk of serious injury for young baseball pitchers: a 10-year prospective study. Am J Sports Med. 2011;39(2):253–7.
5. Olsen S, Fleisig G, Dun S, Loftice J, Andrews J. Risk factors for shoulder and elbow injuries in adolescent baseball pitchers. Am J Sports Med. 2006;34(6):905–12.
6. Sugalski MT. Avulsion fracture of the lesser tuberosity in an adolescent baseball pitcher: a case report. Am J Sports Med. 2004;32(3):793–6.
7. Alexander CJ. Effect of growth rate on the strength of the growth plate-shaft junction. Skeletal Radiol. 1976;1(2):67–76.
8. Centers for Disease Control and Prevention [Internet]. Growth charts – clinical growth charts [cited 2013 Oct 3]. Available from: http://www.cdc.gov/growthcharts/clinical_charts.htm.
9. Jensen RK. Changes in segment inertia proportions between 4 and 20 years. J Biomech. 1989;22(6–7):529–36.
10. Kanehisa H, Yata H, Ikegawa S, Fukunaga T. A cross-sectional study of the size and strength of the lower leg muscles during growth. Eur J Appl Physiol. 1995;72(1–2):150–6.
11. Ford CM, Keaveny TM. The dependence of shear failure properties of trabecular bone on apparent density and trabecular orientation. J Biomech. 1996;29(10):1309–17.
12. Hoffmeister B, Smith S, Handley S, Rho J. Anisotropy of Young's modulus of human tibial cortical bone. Med Biol Eng Comput. 2000;38(3):333–8.
13. Sairyo K, Goel VK, Masuda A, Vishnubhotla S, Faizan A, Biyani A, et al. Three dimensional finite element analysis of the pediatric lumbar spine. Part II: biomechanical change as the initiating factor for pediatric isthmic spondylolisthesis at the growth plate. Eur Spine J. 2006;15(6):930–5.
14. Yoon Y, Yang G, Cowin S. Estimation of the effective transversely isotropic elastic constants of a material from known values of the material's orthotropic elastic constants. Biomech Model Mechanobiol. 2002;1(1):83–93.
15. Blackburn JT, Padua DA. Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. Clin Biomech (Bristol, Avon). 2008;23(3):313–9.
16. Niemeyer P, Weinberg A, Schmitt H, Kreuz PC, Ewerbeck V, Kasten P. Stress fractures in adolescent competitive athletes with open physis. Knee Surg Sports Traumatol Arthrosc. 2006;14(8):771–7.
17. Gómez-Benito MJ, Moreo P, Pérez MA, Paseta O, García-Aznar JM, Barrios C, et al. A damage model for the growth plate: application to the prediction of slipped capital epiphysis. J Biomech. 2007;40(15):3305–13.
18. Chung, S.M. Shear strength of the human femoral capital epiphyseal plate. J Bone Jt Surg. 1976;58(1):94–103.
19. Bright RW, Burstein AH, Elmore SM. Epiphyseal-plate cartilage. A biomechanical and histological analysis of failure modes. J Bone Joint Surg Am. 1974;56(4):688–703.
20. Williams JL, Do PD, Eick JD, Schmidt TL. Tensile properties of the physis vary with anatomic location, thickness, strain rate and age. J Orthop Res. 2001;19(6):1043–8.
21. Jayanthi NA, LaBella CR, Fischer D, Pasulka J, Dugas LR. Sports-specialized intensive training and the risk of injury in young athletes: a clinical case-control study. Am J Sports Med. 2015;43(4):794–801.