Development of a Set of Force Response Equations to Represent the Musculature in Infants to Study Development Dysplasia of the Hip

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DEVELOPMENT OF A SET OF FORCE RESPONSE EQUATIONS TO REPRESENT THE MUSCULATURE IN INFANTS TO STUDY DEVELOPMENTAL DYSPLASIA OF THE HIP

by

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Abstract

This paper describes how a force response equation was created to model muscles, tendons, and ligaments of the hip joint to improve a biomechanical model of an infant hip to study Developmental Dysplasia of the Hip (DDH). DDH is the most common abnormality in newborn infants and is defined as any amount of instability in the hip including complete dislocation. Researchers at our institution are attempting to increase the success rate of treatment methods by creating computer models of the biomechanics of infant hip instability and dislocation. The computer model used a scaled adult pelvis, femur, tibia, fibula and foot to match the size of an infant for the bone geometry. The current infant muscle model is an undifferentiated model based on the area of a single infant muscle, for all muscles modeled. This muscle model was able to provide some insight into the nature of the biomechanics. To improve the infant muscle model, a set of equations differentiated by muscle area was developed. The new set of equations uses a ratio of infant over adult muscle area of a single muscle to create a ratio that can be used to scale all adult muscle areas to infant areas. This model will be more physiologically accurate because it will be differentiated based on muscle area.

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Chapter 1 Introduction

Developmental Dysplasia of the Hip (DDH) is a physical abnormality that is not very well understood and is the term for infants or children with dysplasia of the hip [1]. It is diagnosed in 67 out of 1000 infants [2]. According to the Centers for Disease Control and Prevention (CDC), in 2010 there were approximately 4 million live births in the United States [3]. Using the average rate of diagnosis, the number of infants born with DDH is approximately 260,000. The majority of cases correct themselves, only about 10% require treatment [4, 5]. The most prominent treatment for DDH in the United States is the use of the Pavlik harness. The Pavlik harness has a proven success rate of 80% over a 14 year period [6]. Although the success rate is good, there are still 20% of cases in which the harness fails, which is unacceptable. This leaves about 5,600 infants a year needing a secondary, sometimes surgical, procedure to correct the condition. It is believed that unless the condition is treated at a young age, the hip joint will have significant damage in adulthood. This could cause severe pain and/or lead to a total hip replacement.

In biomechanics, computer models are used to simulate what is occurring in reality [7-9]. All parts of the model need to be as accurate as possible for the best results. To study DDH, researchers have created a model with correct infant pelvis geometry, scaled femur geometry and a single undifferentiated equation used to represent each muscle for all muscles in the model [9, 10]. The equation used to model the muscles can be greatly improved by differentiating the equations based on cross sectional area of the muscles.

1.1 Anatomy

To fully understand our research and for the sake of completeness, a basic overview of anatomy is given. Starting with a list of basic and relevant anatomical terms, followed by a discussion of the hip joint, and finally the differences between infant and adult hips are described. Refer to an anatomy book for more in-depth information [11].

1.1.1 Anatomical Terms

There are three major movements that the upper and lower limbs can complete; flexion/extension, abduction/adduction, and internal/external rotation. A picture of the movements described is displayed below in Figure 1.1. Flexion is the bending of a joint that results in a decrease in the joint angle. The opposite of this movement is extension, it is the bending of a joint that results in an increase in the joint angle. Abduction refers to the movement of a limb away from the middle of the body. Adduction is the opposite, it is the movement of a limb towards the middle of the body. Internal and external rotation are rotational movements either toward or away from the body, respectively, about the long axis of the limb [11].

Figure 1.1 – Anatomical Movements [12].

1.1.2 Hip Joint

The hip joint is a ball and socket type joint. It is comprised mainly of two bones, the femur/femoral head (ball) and pelvis/acetabulum (socket). The joint also includes many muscles and ligaments that provide movement and stability to the joint. In the region of close proximity to the joint, the bones that make up the joint are enclosed in a thick membrane (joint capsule) called the articular capsule [11].

1.1.2.1 Pelvis

The pelvis is a major bone of the hip joint, Figure 1.2 below. For a mature human, it consists of four separate bones that are fused together in a mature human; left and right hip bones, coccyx, and sacrum. The hip bone consists of three separate bones that are fused together; the ilium, ischium and pubis. The three bones are fused in a Y shape pattern in a deep, hemispherical depression on the outer side of the hip called the acetabulum [11]. The acetabulum is the socket in this ball and socket type joint. The bottom rim of the acetabulum has a section missing, this is called the acetabular notch. In a mature human, a fibrocartilaginous material is attached to the rim of the acetabulum, called the acetabular labrum. The acetabular labrum protects the edges of the acetabulum and adds depth to the socket. Additionally, the acetabulum is lined with the acetabular labrum to provide a smooth surface for the femoral head to rotate on [11].

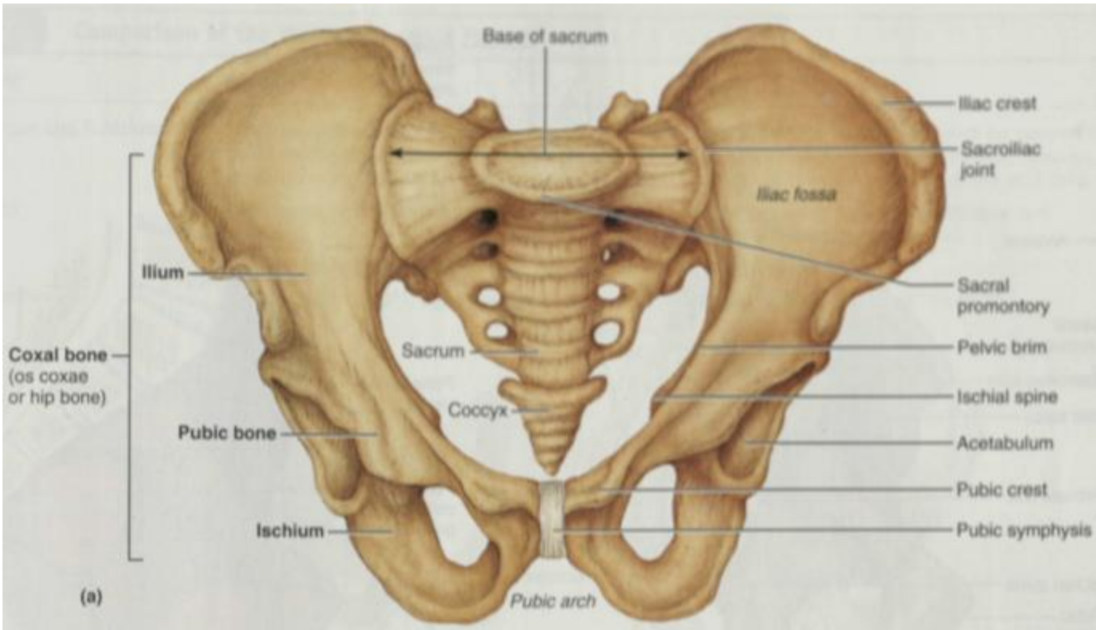

Figure 1.2 – Adult Pelvis with major parts labeled [13].

1.1.2.2 Femur

The femur is the second bone that completes the hip joint. It is the strongest bone in the body. The femur is comprised of 6 major regions; head, neck, greater trochanter, lesser trochanter, shaft/body, and the condyle region, Figure 1.3 below. The head of the femur or femoral head is the ball in this ball and socket type joint. It is not hemispherical like the socket, but more ellipsoidal in shape/volume. Additionally the femoral head is covered in a cartilaginous material which provides cushioning for impacts and reduces surface friction [11]. The shaft of the femur is relatively round past the trochanters and flattens out while approaching the condyle region. Additionally, the shaft of the femur has a ridge on the posterior side that runs along the length of the shaft called the linea aspera, and is the site for many muscle insertions and origins. The condyle region is the point of contact in the knee joint with the tibia [11].

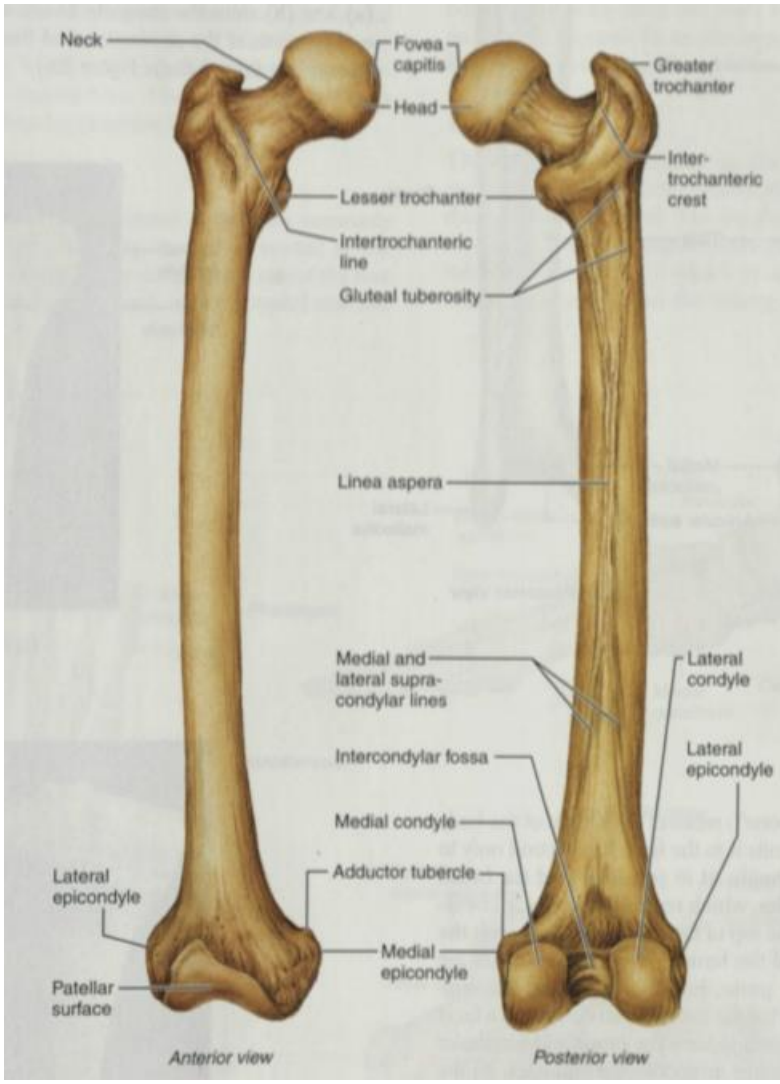

Figure 1.3 – Adult Femur with major parts labeled [13].

1.1.2.3 Muscles

There are many muscles of the hip joint. This section lists the major muscles in the hip separated by location; anterior femoral muscles, medial femoral muscles, gluteal region, and posterior femoral muscles. For more information on muscles including origin/insertion, muscle actions, and more, please refer to a human anatomy book [11, 13]. The anterior femoral muscles include the sartorius, auadriceps femoris and articularis genus [11, 13]. The medial femoral muscles include gracilis, pectineus, adductor longus, adductor brevis, and adductor magnus [11, 13]. The muscles of the gluteal region include gluteus maximus, gluteus medius, gluteus minimus, gensor fasciae latae, piriformis, obturator internus, gemellus superior, gemellus inferior, quadratus femoris, and obturator externus [11, 13]. The posterior femoral muscles include biceps femoris, semitendinosus, and semimembranosus [11, 13].

1.1.2.4 Ligaments

There are seven major ligamentous structures in the hip joint; articular capsule, illiofemoral ligament, pubofemoral ligament, ischiofemoral ligament, ligament of the head of the femur, acetabular labrum, and transverse acetabular. The three major ligaments, illiofemoral, pubofemoral, and ischiofemoral, work together to keep the leg from hyperextending and from over abducting [11].

1.1.3 Differences between Infants and Adults

There are some key differences between the hip anatomy of infants and adults. The most noticeable is the size of the bones, muscles and ligaments, which are smaller in infants. For example, the femur length of an average 2 month boy is 9.2 cm and the femur length of an average 18 year old boy is 54.3 cm [14]. Another difference is the angle at which the femoral head makes with the shaft of the femur, antiversion angle, changes as the infant ages. The angle decreases from about 50° when the infant is first born to 25° when the patient is 8 years old. The shape of the femur change dramatically from infancy to adulthood, Figure 1.4 below [15]. The neck increases in length and the trochanters are formed as the child ages. Furthermore, the acetabulum deepens as the infant ages because the weight of the body [14].

Figure 1.4 – Model of infant femur from CT scans, from the Ortolani Collection in Padua, Italy [15].

Additionally, there is a difference in the material composition of the bones. In infants the bones are actually more cartilage than bone. As humans age, the cartilage will slowly harden to bone via ossification [11].

Ossification of the hip bones begins from three main centers; one in the ilium, ischium and pubis [11]. By puberty these centers have fused in the acetabulum; however the peripheral of the pelvis is not ossified yet. The peripheral is ossified by five minor ossification centers that are on various regions of the pelvis [11].

Ossification of the femur originates from five centers of ossification. The main center located in the shaft. The minor centers are located at the trochanters, femoral head and middle of the condyles. The last ossification center to appear is one located in the lesser trochanter, which can take 13 to 14 years to appear [11].

1.2 Background of Developmental Dysplasia of the Hip

Developmental Dysplasia of the Hip (DDH) is a weakly known condition where the hip joint of an infant or child shows signs of instability or dysplasia (dislocated). DDH currently is diagnosed in 67 infants out of every 1000 live births. The statistics can vary from study to study depending on the definition of DDH used and varying techniques for screening [16]. The techniques utilized to screen for or diagnose DDH include clinical examinations and imaging techniques. There are two broad approaches to treating DDH, one is by physical manipulation of the joint, closed reduction. Which uses clinical procedures, Ortolani and Barlow examinations, and orthopedic devices, Pavlik Harness, Turbinger brace, Craig splint, von Rosen splint, Hip Abduction Brace, Traction Devices and Hip Spica Cast [1, 4, 17, 18]. The other, open reduction, uses surgical techniques to reduce the hip [16]. The treatment of DDH through closed reduction can cause other problems, including avascular necrosis (AVN) of the femoral head [6, 16, 18-20]. There are two main factors that lead to increased risk of DDH, gender and breech birth [16, 20]. Females are more likely to have DDH and breech birth refers to a pelvis/feet first birth. Additionally, the left hip has higher rates of DDH than the right hip [16].

1.2.1 Diagnosis/Screening

There are two main techniques used when screening for DDH, clinical examinations and imaging techniques [1, 16, 18, 20, 21]. The first clinical examination was developed by Ortolani in 1937 [17]. A second clinical examination was developed decades later by Barlow because the Ortolani test was accurate for infants approaching one year of age but not for newborns [4]. These two tests are essentially opposites, the Ortolani test flexes and abducts the hip while the Barlow test flexes and adducts the hip. Modified versions of both of these clinical examinations are still used today, Figure 1.5 below [16]. In addition to the clinical examinations, physicians use imaging techniques to aid in diagnosing DDH. The two main imaging techniques used are, ultrasonography (ultrasound) and radiography (x-rays) [1, 18, 20, 21].

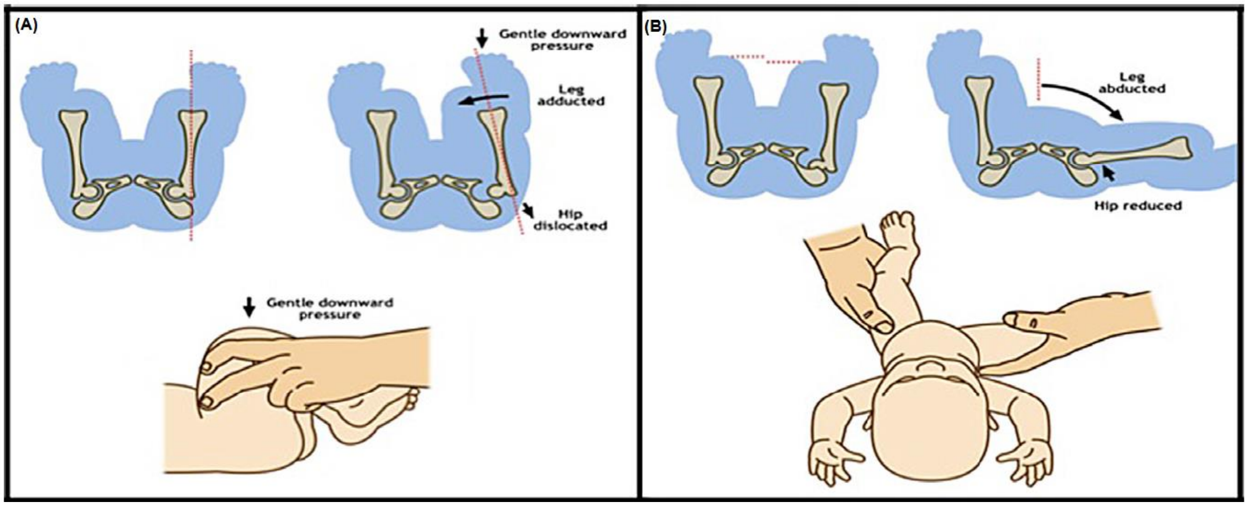

Figure 1.5 – (A) Barlow Examination. (B) Ortolani Examination. [16]

1.2.2 Incidence of DDH

With routine screening and a broad definition of DDH, the known incidence of DDH, number of hips diagnosed over the total number of hips examined, is significant. Incidence of DDH should not be confused with number of infants affected. According to a 1999 retrospective study, reported the number of infants with instability in at least one hip joint was 68 out of 1000 or 6.8%. The same study reported the incidence of DDH was 5.51% for infants or 55 out of 1000 [2]. Incidence is lower because not all infants have instability in both hips. Incidence of DDH varies among ethnic groups and is more pronounced in groups that encourage swaddling [1, 16]. The majority of these cases corrected themselves, only 0.5% hips needed treatment [2]. The percentage of hips corrected without treatment was 90.7% which agrees with Barlow's claim that 88% of hips will be corrected without treatment [2].

1.2.3 Treatment

The goal for treating DDH is to reduce the hip joint without causing damage to the region. Reduction is a medical term for a procedure used to correct a misalignment for dislocation or fracture. There are many options for treating DDH that fall under two categories, closed reduction or open reduction. Closed reduction includes orthopaedic devices and physical manipulation of the femur/femoral head by an experienced orthopaedic surgeon [1, 16, 18]. Open reduction, refers to osteotomies and surgical procedures where the hip joint is opened and tissues that are blocking the femoral head are removed or muscles are released to allow the femoral head to move. These are invasive procedures that are only used when the infant is too old for closed reduction or after closed reduction attempts have failed [1, 16].

There are many orthopaedic devices that are used for closed reduction, including but not limited to: Pavlik harness, Turbinger brace, Craig splint, von Rosen splint, Hip Abduction Brace (Figure 1.6 below), Traction devices, and Hip Spica Cast [1, 18]. The overall concept for these orthopaedic devices is that the devices hold the hip joint in a position, flexed and abducted, that encourages reduction. According to Suzuki, reduction occurs while the infant is sleeping and only the passive portion of the muscle force is acting [22].

Figure 1.6 – Several types of Hip Abduction Braces [1].

The Pavlik harness is one of the most widely used orthopaedic devices used for treating DDH, Figure 1.7 below. One reason for the wide spread use of the Pavlik harness is its proven success rate. The Pavlik harness has a success rate of 80% over a 14 year period [6]. While the success rate is good, there are still 20% of cases where the Pavlik harness fails. The success rate of the harness depends on severity of the dysplasia and the age of treatment. The older the child, the less likely the harness will treat the patient [19]. The more severe the dysplasia, the less likely the harness will treat the patient [18]. A complication when treating or after treating DDH is avascular necrosis of the femoral head, where the femoral head loses its blood circulation [1, 18, 19]. The rate of avascular necrosis of the femoral head is uncommon and occurs in less than 12% of cases treated [6, 16].

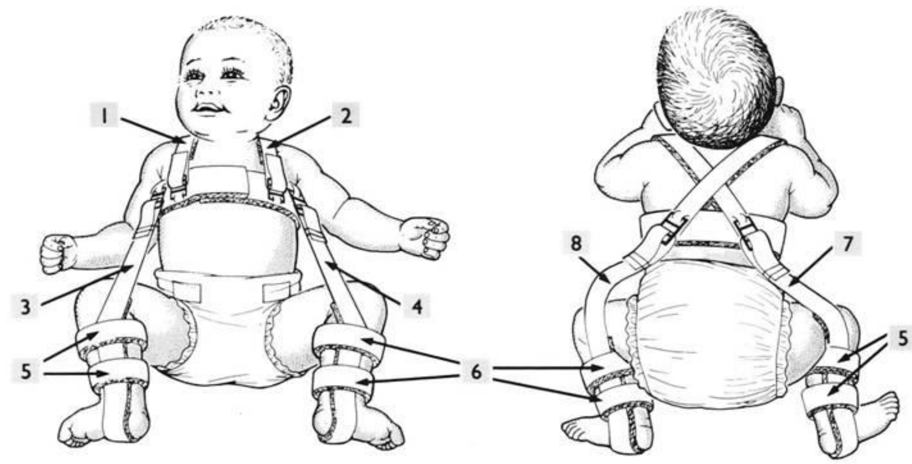

Figure 1.7 – Diagram of Pavlik Harness on a child, with straps labeled [1].

Chapter 2 Background

Researchers in the field of biomechanics use computer models and simulations to study the problem. The software a researcher uses depends on the goal of the research project. This chapter describes the major modeling techniques used in biomechanics. It also describes the behavior of biological materials, equations for muscle modeling, and material constant for some biological materials.

2.1 OpenSim

When the study is focused on skeletal kinematics and moments/torque at the joints, a dynamic simulation software should be utilized. A popular software package to study human movement is OpenSim [7]. This software provides many models of the human skeleton with varying amounts of detail and some focusing on certain areas of the body. OpenSim was developed to be open source to "…accelerate the development and sharing of simulation technology and to better integrate dynamic simulation into the field of movement science." [7]. This allows users to access the source code and add plug-ins or improve the software or make use of the governing equations to use in another software. This includes access to how muscle force is determined and is described later. One limitation of OpenSim is that the joint locations do not move relative to the bones in the joint. Meaning the center of rotation does not change, but the center can move. Therefore this software cannot be used to study reduction of any kind. This software can be used to study how joint dislocations affect motion.

2.2 SolidWorks

A popular dynamic simulation software used to study many types of problems is SolidWorks. This software is robust and comes with contact detection. Any solid model can be imported and used in the simulations. This allows researchers in biomechanics to recreate the human skeleton from medical images, scans of actual specimens, or recreations of the specimen. The specimens include but are not limited to: bones, cartilage, arteries, organs, corrective devices, etc...

Currently, SolidWorks is being utilized to study DDH [9, 10]. The studies use an adult lower limb geometry that is scaled to the size of an infant. The model was altered to account for differences in geometry between infants and adults. The femoral head was made spherical as opposed to elliptical in adults. Additionally, the antiversion angle of the femoral head was increased to match that of an infant [9]. While this model provides quality results. The model can be improved by having actual infant geometry instead of scaled adult geometry. The muscle model used in this study is described later.

2.3 Finite Element Analysis

This type of analysis is performed when the focus of the study is on stresses and deformations. Currently, Finite Element Analysis (FEA) is being used to study adult hip dysplasia [8, 23, 24]. These researchers are interested in contact stress of the femoral head and the acetabulum. This study also provides modeling information for biological materials, including cortical shell thickness, trabecular elastic modulus, acetabulum cartilage elastic modulus and acetabulum cartilage thickness [24]. These parameters can potentially be used to study deformations in the infant hip.

2.4 Muscle Modeling

Muscles, and all biological materials, can be described as hyper-elastic, meaning a larger change in force is needed for the same change of length and the elongation is reversible. For biological tissues, the differential Young's Modulus is a linear function of stress vs strain [25]:

$$\frac{\partial\sigma}{\partial\varepsilon} = a + b\sigma \qquad (1)$$

where σ is stress and ε is strain. This leads to an exponential function for stress as a function of strain. Muscles are extremely deformable, stretching up to 1.6 times their reference length during passive elongation, before locking up and break down of the material.

The equation used to model the passive forces in muscles in infant hip dysplasia studies was based on the equation above. The end result is an exponential equation for passive muscle force, Equation 2 [9, 10]. A single equation for all the muscles, the only variation between the muscles is the relaxed length and the current length, which are unique to each muscle.

$$F = A^{*} \cdot a \cdot \left(e^{b\cdot(\lambda-1)} - 1\right) \qquad (2)$$

Where a and b are material constants, A* is the area of the pectineus muscle for an infant, and λ is the muscle stretch (muscle length)/(reference muscle length). The area of the pectineus was used for all muscles because muscle parameters for infants are not in the literature.

In OpenSim muscles are termed muscle-tendon actuators because the force developed in the muscle depends on the tendon as well. The total force in each muscle-tendon actuator depends on both the active and passive forces in the muscle. The force-length relationship for each muscle-tendon actuator is scaled from a generic Hill model, Figure 2.1. The model is based on peak isometric force, tendon slack length, optimal fiber length and pennation angle [12, 26].

The peak isometric force is the maximum force a muscle can generate and is determined by scaling the Physiological Cross Sectional Area (PCSA) by a constant called the “specific tension” [27]. The “specific tension” of a muscle is constant for all muscles from the same study, i.e. the same human. The purpose of the “specific tension” is to scale the forces of muscles so the combined moment at joints matched experimentally obtained results. Optimal fiber length is defined as the length of the muscle fibers when the muscle is at peak isometric force. Pennation angle is the angle the muscle fibers form with the line of action of the tendon. Optimal fiber length, pennation angle, and PCSA are muscle physiology parameters that are in the literature [28, 29]. Tendon slack length is the length of the tendon at which a force begins to develop in the tendon, similar to spring free length, and must be estimated. Stretching the tendon past this length increases force in the tendon. However, if compressed to a length lower than the tendon slack length, the tendon does not produce a force, it is slack.

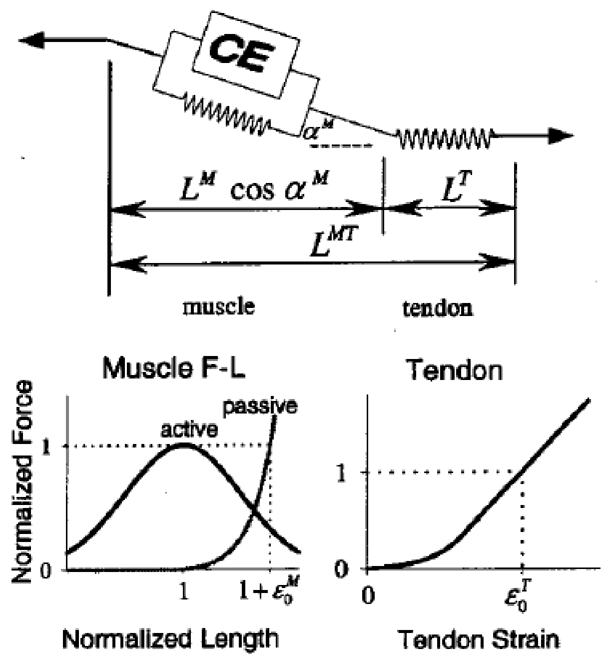

Figure 2.1 - Hill Muscle Model where L^{MT} is muscle-tendon length, L^M is muscle length, L^T is tendon length, α^M is pennation angle. Muscle and Tendon Force are normalized with respect to peak isometric force. Muscle length is normalized with respect to optimal fiber length. Tendon strain is defined as (tendon length – tendon slack length)/(tendon slack length). [12]

The passive force equation for muscle-tendon actuator is based off the normalized passive force-length relationship [7, 12].

$$F = A \cdot C \cdot \bar{F}^{PE} \tag{3}$$

Where A is the PCSA, C is the specific tension, and $\bar{F}^{PE}$ is the normalized passive force-length relationship. An equation that describes this relationship was developed by Thelen, Equation 4 [30].

$$\bar{F}^{PE} = \frac{e^{\frac{k^{PE}(\bar{L}^M - 1)}{\varepsilon_0^M}} - 1}{e^{k^{PE}} - 1} \tag{4}$$

Where $\bar{F}^{PE}$ is normalized passive muscle force, normalized by peak isometric force, k^{PE} is the passive shape factor, $\bar{L}^M$ is normalized muscle length, normalized by optimal muscle length, and ε_0^M is the passive muscle strain at peak isometric force. Combining the Equations 3 and 4:

$$F = A \cdot C \cdot \left(\frac{1}{e^{k^{PE}} - 1}\right) \cdot \left(e^{\frac{k^{PE}(\bar{L}^M - 1)}{\varepsilon_0^M}} - 1\right) \tag{5}$$

Comparing the passive force equations used to model muscles from the infant dysplasia studies and OpenSim, they have the same form and the variables can be symbolically equated as followed.

$$A^* = A \qquad \lambda = \bar{L}^M$$

$$a = C \cdot \left(\frac{1}{e^{k^{PE}} - 1}\right) \qquad b = \frac{k^{PE}}{\varepsilon_o^M}$$

Chapter 3 Problem Definition

The model used to study DDH is good, improvements can be made to the model. The size of the geometry is correct, although the shape of the femoral head and greater trochanter area is different between adults and infants as described before. The equation used to model muscles however, can be improved by implementing a similar strategy as Delp in OpenSim [7, 12]. This paper will attempt to determine a single force response equation that is equivalent to the muscle and tendon equation's used in OpenSim.

3.1 Problem

This research focuses on the physical abnormalities of infants, more specifically DDH. DDH, is a condition in infants where the hip joint is unstable or displaced. For a more extensive explanation, review the earlier sections in this work. To study DDH, this research employs the principles of Statics, Solid Mechanics, and Numerical Methods. This research seeks to find a numerical passive muscle model that is more physiologically accurate than the current muscle model utilized to study DDH.

The equation used to model muscles to study DDH does not account for the size of the muscle, only the length compared to its reference length. This undifferentiated model does not account for the different PCSA muscles have. Meaning the pectineus and gracilis produce the same passive force at the same stretch, λ, which is not physiologically correct. Additionally, the current muscle model does not account for the force of the tendon. To improve the model, a single force response equation that is equivalent to the muscle and tendon equations in OpenSim should be developed for each muscle in the infant hip model. The equivalent force response equation should be based on the muscle's PCSA and λ.

3.2 Hypothesis

An adequate equivalent force response equation, that models both muscle and tendon, will not be achieved by substituting the equivalent values for ***a*** and ***b*** from the OpenSim muscle model equations into the model used to study DDH and using infant PCSA corresponding to the muscle. The force response equation would not be achieved because not all of the muscles are modeled, like in OpenSim, and the tendons are not modeled. To overcome this, the variables ***a*** and ***b*** must be tuned.

3.3 Contributions

Contributions from this research to the scientific community include:

- Improved infant hip model
- Insight into the scaling effects of the muscle models
- New passive muscle model for infants
- Undergraduate, Honors in the Major, Thesis on the research

3.4 Novelty and Significance

There is little research on biomechanics for infants. Therefore, almost any research in this area can be useful and/or significant. This research is significant and useful because the procedure used can be replicated to include more muscles, or used on a different joint in the body. Based on the extensive literature review, an infant muscle model has not been described in the literature.

Chapter 4 Approach

To improve the muscle model in the model of infant hip used to study DDH, the infant muscle equations should be differentiated based on muscle PCSA. The equations must be tuned to match the passive equilibrium position of the hip. This section will explain the approach in further detail.

4.1 Infant PCSA

The infant PCSA of each muscle must be estimated because a database infant PCSA is not described in the literature. The PCSA of the infant adductor brevis was reported as 0.41 cm^2 by Ardilla et al. [10]. To estimate the infant PCSA of the other muscles the PCSA of the infant adductor brevis was divided by the PCSA of the adult adductor brevis, to give a ratio. This ratio was multiplied by the PCSA of an adult muscle to estimate the corresponding PCSA of the infant muscle for each muscle modeled.

4.2 Calibration

The infant muscle model cannot be validated using the same procedure as OpenSim because the moment curves at the hip joint are not in the literature. To calibrate the infant model, the relaxed position of the legs was chosen as the calibration point. This was done because the relaxed position of the leg is the position in which the passive muscle contribution equals the weight of the leg. The calibration position is 90° flexion and 70° abduction. This is also the method the group chose to calibrate the muscle model currently in use. SolidWorks will be used to determine if the set of force response equations can hold the leg in the calibration position.

4.3 Tuning Variables

To tune the force response equations to allow the model to reach equilibrium at the calibration position of 90° flexion and 70° abduction, the variable's ***a*** and ***b*** can be altered. The first approach will be attempting to tune just the ***a*** variable, while keeping the ***b*** variable the same as OpenSim equivalent. This is done because OpenSim is a recognized software in gait analysis and the muscle model in this software has been validated thoroughly. If equilibrium cannot be achieved by the first method, the ***b*** variable will be changed slightly, then attempt to find an ***a*** that will allow the model to reach equilibrium.

Chapter 5 Results and Discussion

5.1 Infant Muscles PCSA

The PCSA for infant muscles was estimated by scaling the adult areas given in OpenSim by a scaling factor. This scaling factor was the area of the infant adductor brevis area, 0.41 cm^2 divided by the area of the adult adductor brevis, 11.52 cm^2. The resulting value of the scaling factor was 0.03559. Multiplying an adult muscle PCSA by the scaling factor will result in an estimated value for the infant PCSA, for any muscle. For the muscles currently modeled in the infant hip model, the results from the scaling are in Table 5.1.

Table 5.1 – Adult PCSA and Estimated Infant PCSA for muscles in infant hip model

Muscles	Adult PCSA (cm^2)	Infant PCSA (cm^2)
Adductor Brevis	11.52	0.41
Pectineus	9.03	0.32
Adductor Longus	22.73	0.81
Adductor Magnus (minimus)	25.52	0.91
Adductor Magnus (medius)	18.35	0.65
Adductor Magnus (posterior)	16.95	0.60
Gracilis	3.73	0.13

5.2 Equivalent *a* and *b* from OpenSim

The first attempt to create a set of force response equations was to use the OpenSim equivalents of ***a*** and ***b***. The ***a*** and ***b*** equivalents from OpenSim were used in Equation 6, where ***a*** = 0.466 N/cm^2, ***b*** = 6.67 and ***A*** corresponds to the PCSA of the muscle.

$$F = A \cdot a\left(e^{b(\lambda-1)} - 1\right) \tag{6}$$

The force response equations are plotted in Figure 5.1, below. This set of force response equations was unable to achieve equilibrium at the calibration position using

SolidWorks®. The leg fell well past the equilibrium position and the simulation was stopped. The moment generated by the weight of the leg was greater than the moment generated by the set of force response equations. This set of force response equations was unable to hold the leg in the equilibrium position because it is incomplete musculature and the tendons are not accounted for like in OpenSim. Additionally, the tendons are not accounted for, and are responsible for stopping the joints at the extreme ranges of motion.

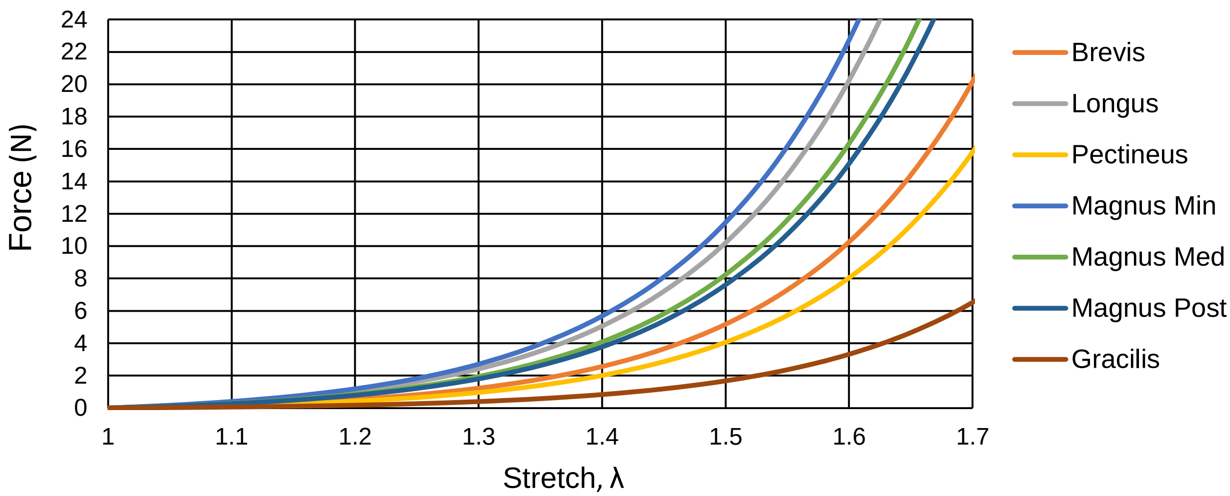

Figure 5.1 – Force Response Equations Plotted using *a* and *b* equivalents from OpenSim.

5.3 Equivalent *b* from OpenSim and Tuned *a*

In an attempt to achieve equilibrium at the calibration point, the ***a*** value was tuned using an Excel® spreadsheet developed by Christopher Rose and the equivalent ***b*** value from OpenSim®. The spreadsheet developed by a member of the research group, Christopher Rose, uses the origin and insertion points from the SolidWorks® model to compute the muscle lengths and forces produced when the leg is in various positions. The spreadsheet is faster than a SolidWorks® simulation and allows researchers to edit the variables in the force response equations and see the results much faster. The tuned ***a*** and equivalent ***b*** from OpenSim® were able to achieve equilibrium in the spreadsheet. However, when the equations for force response were substituted into the SolidWorks® model and equilibrium was simulated, the force response equations were unable to hold the leg in the equilibrium position. This is because the moment in the plane of flexion was not strong enough to overcome the moment due to the weight of the leg. From this, it was determined that the ***b*** variable must be tuned.

5.4 Tuned *a* and *b*

The next attempt to achieve equilibrium was with tuned ***a*** and ***b*** values. The ***b*** value was changed in the spreadsheet mentioned above, and the resulting ***a*** value to achieve equilibrium was automatically calculated. In order to achieve equilibrium, the value of ***b*** must be greater than 10. To simulate the effect of ligaments, which limit the range of motion of joints, the value of ***b*** was found to be 13.95 and the value of ***a*** was 0.0337 N/cm^2. This value of ***b*** was chosen because it limits the maximum angle of abduction to about 80°. While there is no unique set of ***a*** and ***b*** values, because of the single equilibrium point, it is important to note that physiological considerations, mentioned above, were taken into account when selecting the value for ***b***.

After achieving equilibrium in the spreadsheet with the proper moments to ensure equilibrium, the force response equations were substituted into SolidWorks® and a simulation was run. The simulation was successfully able to achieve equilibrium at the calibration position. The value of ***b*** is significantly higher than the OpenSim® equivalent because it attempts to account for muscles, tendons, and ligaments that are not included in the model. The force response equations for the tuned ***a*** and ***b*** are shown below in Figure 5.2.

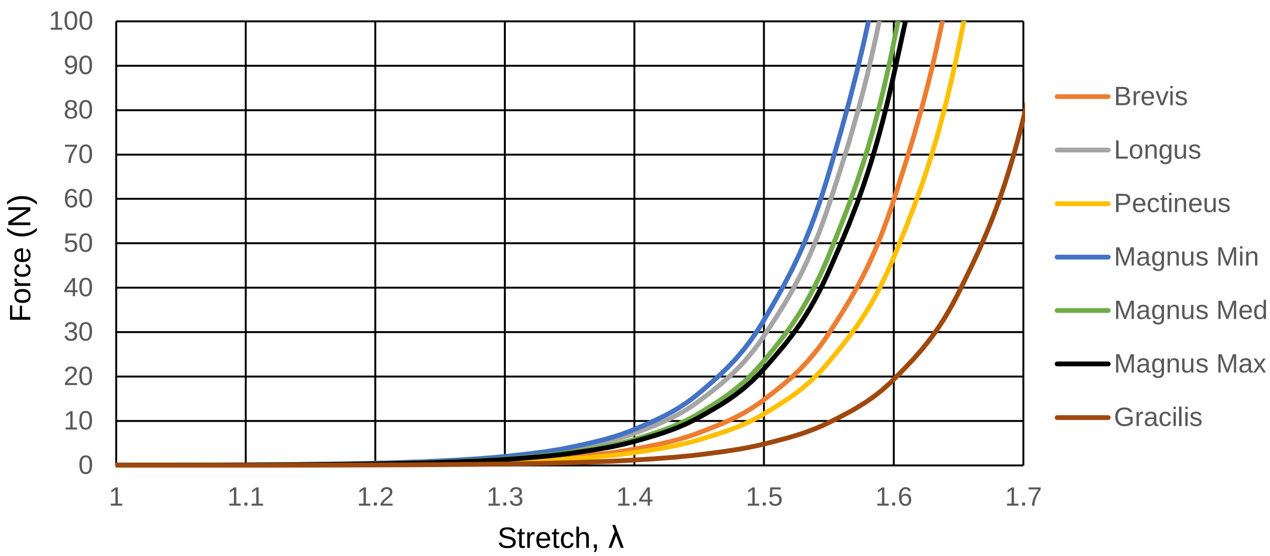

Figure 5.2 – Force Response Equations Plotted using tuned *a* and *b*.

Chapter 6 Conclusion

In conclusion, this paper offered an approach to determine an improved set of force response equations that model the muscles, tendons and ligaments in an infant hip model used to study DDH. The new set of equations is an improvement over the old model that only accounted for muscles and was not differentiated based on PCSA of the muscles. The final set of force response equations, with tuned ***a*** and ***b*** values, accounted for muscles, tendons and ligaments, and differentiated based on PCSA of muscles modeled. The set of equations was able to hold the leg in the infant hip model in the equilibrium position in a simulation using SolidWorks®.

The set of force response equations or improved muscle model, was used in additional research by other members of the research group [31, 32]. A major conclusion drawn from these studies is the effect of the pectineus on reduction. In the most severe cases of dysplasia, the pectineus is the largest force producer based on its stretch in the equilibrium position and when displaced.

Further work can be done to improve the infant hip model and muscle model. First, adding more muscles into the infant hip model will help provide a more physiologically accurate representation. Second, modeling the tendons using the tendon equation in OpenSim. Third, modeling the ligaments using the tendon equation from OpenSim.

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