
Victor Huayamave  
*University of Central Florida*, huayamav@erau.edu

Christopher Rose  
*University of Central Florida*

Sheila Serra  
*University of Central Florida*

Brendan Jones  
*University of Central Florida*

Eduardo Divo  
*Emry-Riddle Aeronautical University*

Scholarly Commons Citation  

This Article is brought to you for free and open access by Scholarly Commons. It has been accepted for inclusion in Publications by an authorized administrator of Scholarly Commons. For more information, please contact commons@erau.edu, wolfe309@erau.edu.
Authors
Victor Huayamave, Christopher Rose, Sheila Serra, Brendan Jones, Eduardo Divo, Faissal Moslehy, Alain J. Kassab, and Charles T. Price

Victor Huayamave a, Christopher Rose a, Sheila Serra a, Brendan Jones a, Eduardo Divo b, Faissal Moslehy a, Alain J. Kassab a,*c, Charles T. Price c,d

a Department of Mechanical and Aerospace Engineering, University of Central Florida, Orlando, FL 32816, USA
b Department of Mechanical Engineering, Embry-Riddle Aeronautical University, Daytona Beach, FL 32114, USA
c Pediatric Orthopedic Surgery, Arnold Palmer Hospital and International Hip Dysplasia Institute (IHDI), Orlando, FL 32806, USA
d College of Medicine, University of Central Florida, Orlando, FL 32827, USA

1. Introduction

Developmental Dysplasia of the Hip (DDH) is an abnormal condition where hip joint dislocation, misalignment, or instability is present in infants. According to Bialik et al. (1999), “Congenital dislocation of the hip represents one of the most important and challenging congenital abnormalities of the musculoskeletal system”. It has been reported that in 20 cases per 1000 births, babies have some instability (USPSTF, 2006) and 6 out of every 1000 will require treatment (Bialik et al., 1999). Studies had also shown that DDH was responsible for 29% of primary hip replacements in people up to age 60 years (Furnes et al., 2003).

There are four grades (1–4) of dislocation with Grade 4 being most severe as specified by the International Hip Dysplasia Institute (IHDI) (Narayan et al., 2000). The grade classification of the IHDI has been adopted in this paper to characterize dislocations for DDH because it is applicable to infants and to older children. The IHDI Classification has also been found more reliable than earlier classifications (Fig. 1a). The Pavlik Harness (PH) (Fig. 1b) is a standard non-surgical treatment for DDH (Ramsey et al., 1976; Weinstein et al., 2003) that is designed to maintain the hips in abduction and flexion to bring the femoral head back into the acetabulum (reduction). Clinical reports and previous research indicate very low success of the PH for more severe

---

grades of hip dislocation (Ardila et al., 2013; Atalar et al., 2007; Uras et al., 2014). Also, the incidence of failure doubles when treatment begins after three weeks of age (Harding et al., 1997).

It has been found that PH treatment fails in approximately 15% of cases when initiated shortly after birth (Kitoh et al., 2009; Mubarak et al., 1981; Viere et al., 1990; Weinstein et al., 2003). Statistically it has been shown that for severity Grades 1–3, successful reduction rate is 92% while for Grade 4 the successful reduction rate is approximately 2% (Grill et al., 1988). Therefore, special attention is needed for the fully dislocated hip described by a Grade 4. This paper uses a rigid body dynamics model constructed from MRIs and Ct-scans to quantify the force and moment generation of the adductor muscles when the PH is used. Specifically, the interest is to explore the unsuccessful reduction of Grade 4 and explore alternatives for reduction of this severe dislocation.

2. Methods and materials

In order to study the complex biomechanics of DDH treatment with the PH, a Rigid Body Dynamics (RBD) model has been developed. A 3D model of the pelvis-femur-lower limb assembly of a representative 10-week old female infant (TWOFI) was generated based on a composite of Ct-scans and MRIs of a 6-month and 14-year old female as well as the Visible Human Project with the aid of anthropomorphic scaling of anatomical landmarks. Five adductor muscles were considered as mediators of reduction. A Fung-type muscle model was calibrated to achieve equilibrium at 90° flexion and 80° abduction. The hip was computationally dislocated according to the grade under investigation, the femur was constrained to move in an envelope

![Fig. 1. (a) IHDI dislocation grades (Narayanan et al., 9000) for a representative 10 week old anatomy (ossified regions are enclosed by solid lines), (b) typical Pavlik harness configuration.](image1)

![Fig. 2. (a) Superposition of a solid model of FYOF rendered from a CT-scan (left) with a solid model rendered from a Ct-scan of SMOFI (right), (b) spherical femoral head in neonates: computational model (left) and dissection (right) (Ortolani and Ortolani), (c) illustration of 50° anteversion angle in a 3D solid model (left) and a dissected femur showing anteversion angle (right) (Ortolani and Ortolani).](image2)
consistent with PH restraints, and the dynamic response under passive muscle action and under the effect of gravity was resolved using the ADAMS solver in Solidworks (Dassault Systemes, Concord, MA). Hyperflexion was also investigated for severe dislocations using this model.

2.1. Three-dimensional anatomical solid model of a 10-week old female infant

Anatomical lower limbs models representative of a TWOFI were constructed using a combination of CT-scans from several human subjects and the use of medical segmentation packages: Mimics and 3-matic (Materialise Inc., Plymouth, MI). The lower extremity was composed of the hip, femur, tibia, fibula, and foot. It must be mentioned that symmetry was assumed along the sagittal plane, which divides the body into right and left portions (x–y plane). Thus only the right lower limb was modeled.

Due to the presence of large cartilaginous zones, the MRIs of a neonate offer a challenge in reconstruction of the complete anatomy based on medical segmentation software alone. Moreover, retrospective MRIs and CT-scans of subjects with DDH focus mainly on hip region and therefore do not provide the required geometric information of the rest of the anatomy. As such, a model was constructed from a composite of CT-scans of a 6-month and 14-year old female as well as the Visible Human Project with the aid of anthropomorphic scaling of anatomical landmarks in order to assemble the TWOFI model. It is also known that bone growth is not proportional in all directions, thus, anisotropic parameters will be used for all the lower limb bones.

For hip reconstruction, the starting point was to obtain rough bone geometry contours from CT-scan of a six month old female infant (SMOFI) and then superimpose it on well-defined bone geometry obtained from high quality CT-scan data of a fourteen year old female (FYOF). Both the SMOFI and the FYOF models were superimposed so that iliac spines and acetabuli were aligned (Fig. 2a). This accounts for the hip widening that occurs in females during puberty. Once the geometry was obtained, the SMOFI was scaled to a TWOFI by matching infant

---

Table 1

Lower extremity mass distribution and scaled TWOFI PCSAs using shape adductor brevis factor (f) (Friederich and Brand, 1990).

<table>
<thead>
<tr>
<th>Total body mass (BM)</th>
<th>f = 0.036</th>
<th>Friederich PCSA (cm²)</th>
<th>TWOFI scaled PCSA (cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>7.26 kg</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Limb BM</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thigh 9.45</td>
<td>0.69</td>
<td>Pectineus 9.03</td>
<td>0.321</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Adductor brevis 11.52</td>
<td>0.410</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Adductor longus 22.73</td>
<td>0.809</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Adductor magnus min 25.52</td>
<td>0.908</td>
</tr>
<tr>
<td>Leg 4.2</td>
<td>0.30</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Adductor magnus min</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Adductor magnus middle</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Adductor magnus post</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Gracilis 3.73</td>
<td>0.133</td>
</tr>
<tr>
<td>Foot 1.35</td>
<td>0.10</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Full leg 1.5</td>
<td>1.09</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

---

Please cite this article as: Huayamave, V., et al., A patient-specific model of the biomechanics of hip reduction for neonatal Developmental Dysplasia of the Hip: Investigation of... Journal of Biomechanics (2015), http://dx.doi.org/10.1016/j.jbiomech.2015.03.031
orthopedic data found in Hensinger (1986)’s Standards in Pediatric Orthopedics. The final hip anisotropic scaling factors were determined to be: 0.35 in the x-direction, 0.3 in the y-direction, and 0.35 in the z-direction. The acetabulum depth was approximately 7.9 mm, the acetabular depth-width ratio (ADR) was 45%, and the acetabulum diameter was approximately 17.5 mm.

High-quality CT-scans and MRIs from the visible human project (VHP) (NIH, 2009) were used to reconstruct the femur, tibia, fibula, and foot. The geometry was reconstructed using the superposition approach used for the hip. Scaling parameters for all the other bones of the lower extremity are found to be: 0.23 in the x-direction, 0.22 in the y-direction, and 0.25 in the z-direction. The femoral head had a diameter of 14 mm and was assumed to be perfectly spherical as observed during infant dissections (Fig. 2b).

The femoral anteversion angle also plays a critical role on hip reduction since it will affect muscle force and moment contribution while the PH is in use. This angle defines the difference between the axis of the femoral neck and the transchondylar axis of the knee. This angle ranges typically from 30° to 40° at birth, but can reach up to 70°. The anteversion decreases in adulthood to about 15°. It has been shown that anteversion angles between 0° and 40° had relatively small effect on the isometric moment-generating capacity (Delp et al., 1994). Consequently, the present study considers an anteversion angle beyond 40°, specifically, the model under consideration will be based on a representative average (Sankar et al., 2009) anteversion angle of 50° (Fig. 2c).

Once solid models were completed for the hip, femur, tibia, fibula, and foot, Solidworks is used to assemble all the components (Fig. 3a). To assemble the femur to the pelvis origins and orientations in both bones were specified as defined by Dostal and Andrews (1981). The origin of the inertial Cartesian reference frame was located at the center of the acetabulum for the hip and at the center of the femoral head for the femur. Five adductor muscles were identified as mediating muscles during reduction with the PH: (1) Pectineus, (2) Adductor Brevis, (3) Adductor Longus, (4) Adductor Magnus, and (5) Gracilis. For this study, musculoskeletal modeling of the lower extremity muscles included: (a) Adductor Magnus Minimus, (b) Adductor Magnus Middle, and (c) Adductor Magnus Posterior. Origin and insertion points were defined using the anisotropic scaling parameters previously mentioned on coordinate points obtained in experiments by Dostal and Andrews on the lower extremity (Fig. 3b). Once the model is assembled, the hip can be computationally dislocated to match physiological dislocations according to the DDH dislocation grade under investigation (Fig. 3c). Calculations of the total body mass were based on a TWOFI at the 50th length-for-age percentile (CDC, 2009) in order to account for the muscle mass attached to each of the lower limbs (Table 1). Centers of mass were also determined, thus achieving accurate load and moment distribution in the model (Drillis and Contini, 1966).

Gravity acts as the sole external driving load in the dynamics model to be described later.

2.2. Passive tension muscle model

A muscle produces two kinds of forces, active and passive, which sum to compose muscle total force. The active component is due to the contraction of muscles fibers that generate force and bear load, while passive components are due to the stretching of muscle fibers, the basilar lamina, and the muscle fascia. When a relaxed muscle is stretched beyond its resting length, it behaves as a deformable body: it deforms and provides passive resistance to the stretch. The resistance does not require metabolic energy, and hence is called passive (Gajdosik, 2001). Moreover, passive response is characterized as hyperelastic.

According to Iwasaki (1983), Suzuki (1994), reductions of DDH with the PH occur passively with muscle relaxation in deep sleep. In recent clinical studies, sleep was induced to attain muscle relaxation, and, in flexion and abduction, the hips adjusted to the appropriate height were placed under the knees to avoid forcible abduction and it was clinically observed that under mild sedation and in natural abduction and flexion, the spontaneous reduction with the PH is reproduced (Fukiaige et al., 2014). This further justifies the passive muscle model adopted in this study.

Well-known models (Hill, 1970; Fung, 1972) are extensively used for biological soft connective tissue based on the experimental observation that such tissue exhibits a linear relationship between the differential Young’s modulus and stretch leading to an exponential behavior of the force-stretch curve. A Hill-based model (Delp, 1990) had been proposed to estimate passive muscle force by normalizing it using a peak isometric muscle force (PIMF) as shown in Eq. (1):

$$F = \frac{P_{\text{IMF}}}{C_0}$$

where $P_{\text{IMF}}$ is the normalized passive muscle force, $C_0$ is the PIMF, and $F$ is the muscle force. Delp also reports peak isometric force is proportional to its physiologic cross sectional area (PCSA) as seen in Eq. (2):

$$C_{\text{PCSA}} = \frac{C_{\text{PCSA}}}{C_{\text{PCSA}}}$$

where $C_{\text{PCSA}}$ has a value of 25 N/cm² (Delp, 1990) and was estimated to scale PCSSAs that were measured on young adult cadavers to match moment curves measured on young subjects. Furthermore, Thelen (2003) reports that $F_{\text{IMF}}$ can be expressed as

$$F_{\text{IMF}} = \frac{C_{\text{PCSA}}}{C_{\text{PCSA}}}$$

where $K_0$ is an exponential shape factor with a value of 4 obtained from OpenSim (Delp et al., 2007), $C_{\text{PCSA}}$ is the passive muscle strain with a value of 0.60 for young adults, and $\lambda$ is the optimal muscle-fiber length (stretch). The stretch can be written as $\lambda = L/L_0$ with $L$ being the deformed length, and $L_0$ being the initial muscle length at rest measured at 20° abduction and 120° flexion for newborns. This flexion and adduction ‘neutral’ configuration accounts for the fact newborns maintain a posture similar to the fetal positioning for the first few months after birth with the hips flexed in toward the belly and wide apart. This initial muscle length at rest is multiplied by a specified muscle pre-stretch for the lower extremity muscle and is usually between 20 and 30%. By combining Eqs. (1)-(3), an expression can then be derived to obtain passive muscle force as function of stretch:

$$F = C_{\text{PCSA}}\left(\frac{K}{E_{\text{PCSA}}}\right)$$

Using MRI data, a TWOFI PCSA of 0.41 cm² for the adductor brevis was estimated by measuring the volume of the muscle and dividing it by its average length. A shape factor was then defined, by using a PCSA of 11.52 cm² of an adult’s adductor brevis (Friederich and Brand, 1990), as the ratio between an adult PCSA and the TWOFI PCSA. The adductor brevis shape factor of 0.036 was then used to scale all of the other muscles measured by Friederich (Table 1).

2.3. Rigid body dynamics (RBD) model

Numerical RBD was carried out using the ADAMS dynamics motion analysis module in Solidworks. The governing differential equations of motion provided in

![Fig. 4](http://dx.doi.org/10.1016/j.jbiomech.2015.03.031)
are numerically solved subject to the natural constraints $\phi(q, t) = 0$. The system mass matrix is $M$, the vector of generalized coordinates for the displacements is $q$, the Lagrange multiplier adjoining the gradient constraints $\phi_t$ is $\beta$, the set of configuration and applied motion constraints is denoted by $\gamma$, the matrix $A^T$ projects the applied forces in the $g$ direction, and $F$ is the vector of applied forces. The governing equations of dynamics described in Eq. (5) are Newton's laws of motion augmented via a Lagrange multiplier by constraints. The Force vector is in general a function of displacement, linear in the case of a Hookean solid or non-linear as in our case as in Eq. (5), and a function of time-derivative of the displacements in case of the presence of damping. In this study, the Gear Stiff (GSTIFF) integrator was used, the Jacobian was updated at each time-step, 25 sub-level iterations were taken per time-step, and the initial, minimum, and maximum integrator step sizes were set to $1 \times 10^{-2}$, $1 \times 10^{-3}$, and $1 \times 10^{-4}$. The hip joint is a multi-axial ball and socket joint and physiologically considered a synovial joint on which the bone ends are covered by a slippery tissue called cartilaginous cartilage to create a frictionless bearing point. Therefore, the contact between surfaces was assumed to be frictionless using a restitution coefficient of zero. Moreover, the constraints serve to restrain the motion within the envelope realized with the PH during Grade 1-4 studies and to constrain the knees in a hyperflexed configuration for Grade 4 studies. Custom functions were written in Solidworks for each of the muscles studied using the model described in Eq. (4) with their respective PCAs.

### 2.5. Anatomical dislocation configuration

The TWOFI model was dislocated as previously explained to investigate dislocations Grades 1-4. Subtle dislocations are modeled for Grades 1 and 2 as described by the HDI (Fig. 6a and b). For Grade 3, ossification center is at the level of the superolateral margin of the acetabulum. The femoral head was placed at the rim of the acetabulum (Fig. 6c) and the center of the femoral head was placed at $x = -3.44$ mm, $y = 1.14$ mm, $z = -6.59$ mm relative to the origin located at the center of the acetabulum. For Grade 4, ossification center is above the level of the superolateral margin of the acetabulum. The femoral head was fully dislocated and positioned on the posterior wall of the acetabulum. Specifically, the center of the femoral head was placed at $x = -10.05$ mm, $y = -12.69$ mm, and $z = 1.87$ mm relative to the origin located at the center of the acetabulum (Fig. 7a).

### 3. Results

The TWOFI lower extremity model was used to quantify force and moment generation of the adductor muscles. Results provide insight of their contribution to reduction of DDH on severe Grade 3 and 4 dislocations whenever the PH is used and gravity is acting. It is known from clinical observation that reduction is successful whenever the PH is used on Grades 1 and 2 (Uras et al., 2014), thus these two grades were analyzed but values are not reported. Force and moment contribution of the adductor muscles while the PH is being used are reported for Grade 3 and Grade 4 (Table 3).

### 4. Discussion

As expected, Grades 1 and 2 were successfully reduced, which is consistent with our previous study based on a synthetic model (Ardila et al., 2013) as well as from clinical observation (Uras et al., 2014). Grade 3 was successfully reduced while Grade 4 dislocation was not. However, Grade 4 dislocation reduction was achieved by an external rotation of the lower extremity followed by a semi-circular path taken

---

Table 2

<table>
<thead>
<tr>
<th>Muscle</th>
<th>$\lambda$ (Stretch)</th>
<th>Total force (N)</th>
<th>X-moment (N m)</th>
<th>Y-moment (N m)</th>
<th>Z-moment (N m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pectineus</td>
<td>1.710</td>
<td>45.84</td>
<td>0.2834</td>
<td>0.3328</td>
<td>0.0738</td>
</tr>
<tr>
<td>Adductor brevis</td>
<td>1.523</td>
<td>15.01</td>
<td>0.1353</td>
<td>0.1342</td>
<td>-0.0165</td>
</tr>
<tr>
<td>Adductor longus</td>
<td>1.422</td>
<td>1.35</td>
<td>0.0138</td>
<td>0.0028</td>
<td>-0.0006</td>
</tr>
<tr>
<td>Adductor magnus</td>
<td>1.392</td>
<td>1.83</td>
<td>0.0189</td>
<td>0.0013</td>
<td>-0.0014</td>
</tr>
<tr>
<td>Adductor magnus</td>
<td>1.316</td>
<td>0.84</td>
<td>0.0085</td>
<td>-0.0013</td>
<td>-0.0012</td>
</tr>
<tr>
<td>Adductor magnus</td>
<td>1.229</td>
<td>0.33</td>
<td>0.0031</td>
<td>-0.0030</td>
<td>-0.0007</td>
</tr>
<tr>
<td>Goalis</td>
<td>1.357</td>
<td>0.33</td>
<td>0.0040</td>
<td>0.0016</td>
<td>-0.0010</td>
</tr>
</tbody>
</table>

---

Fig. 5. (a) Contact forces (single headed arrow) and muscle lines of action (double headed arrow) on x-z-plane. (b) Contact forces (single headed arrow) and muscle lines of action (double headed arrow) on x-y-plane. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)
by the femoral head along the posterior wall of the acetabulum along the ischial plane, and the subsequent entrance into the acetabulum via the acetabular notch (Fig. 7b). This path of reduction is generally consistent with clinical studies performed to avoid surgical release of adductor muscles by Papadimitriou et al. (2007) where the reduction phase is divided in three sub-phases: (1) the femoral head travels from the grade 4 dislocated position arriving at the ischial tuberosity, (2) the femoral head rides over the ischial tuberosity to obturator...
foramen, and (3) the femoral head departs from the obturator foramen and enters the acetabulum via the acetabular notch, achieving reduction. Papadimitriou studies showed successful reduction of severe dislocations by this alternative hyperflexion procedure using the Hoffmann–Daimler harness keeping the hip flexed at 120°.

In summary, a patient-derived geometry was successfully reconstructed to model the right lower extremity of a TOFWI. The model developed allows for the detailed study of the biomechanics of DDH using a hyperflexion approach. Foramen and enters the acetabulum via the acetabular notch, achieving reduction with the PH or in combination with other orthosis. A

### Conflict of interest statement

None.

### Acknowledgments

This study was supported in part by the National Science Foundation under Grant number CBET-1160179, Orlando Health, and the International Hip Dysplasia Institute. The authors gratefully acknowledge the photos of Ortolani Dissection provided by George H. Thompson, M.D. and Scott J. Mubarak, M.D. We thank Freeman Miller, MD and Julie Zielinski, MD for their valuable discussions and the insight they provided in the course of this project.

### References


Please cite this article as: Huayamave, V., et al., A patient-specific model of the biomechanics of hip reduction for neonatal Developmental Dysplasia of the Hip: Investigation of... Journal of Biomechanics (2015), http://dx.doi.org/10.1016/j.jbiomech.2015.03.031