A Patient-Specific Model of the Biomechanics of Hip Reduction for Neonatal Developmental Dysplasia of the Hip: investigation of strategies for low to severe grades of DDH.

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21 **Abstract**

22 A physics-based computational model of neonatal Developmental Dysplasia of the Hip
23 (DDH) following treatment with the Pavlik Harness (PV) was developed to obtain muscle force
24 contribution in order to elucidate biomechanical factors influencing the reduction of dislocated
25 hips. Clinical observation suggests that reduction occurs in deep sleep involving passive muscle
26 action. Consequently, a set of five (5) adductor muscles were identified as mediators of
27 reduction using the PV. A Fung/Hill-type model was used to characterize muscle response. Four
28 grades (1-4) of dislocation were considered, with one (1) being a low subluxation and four (4) a
29 severe dislocation. A three-dimensional model of the pelvis-femur lower limb of a
30 representative 10 week-old female was generated based on CT-scans with the aid of
31 anthropomorphic scaling of anatomical landmarks.

32 The model was calibrated to achieve equilibrium at 90° flexion and 80° abduction. The hip
33 was computationally dislocated according to the grade under investigation, the femur was
34 restrained to move in an envelope consistent with PV restraints, and the dynamic response
35 under passive muscle action and the effect of gravity was resolved.

36 Model results with an anteversion angle of 50° show successful reduction Grades 1-3, while
37 Grade 4 failed to reduce with the PV. These results are consistent with a previous study based
38 on a simplified anatomically-consistent synthetic model and clinical reports of very low success
39 of the PV for Grade 4. However our model indicated that it is possible to achieve reduction of
40 Grade 4 dislocation by hyperflexion and the resultant external rotation.

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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 (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 1,000 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., 2001).

There are four grades (1-4) of dislocation with Grade 4 being most severe as specified by the International Hip Dysplasia Institute (IHDI) (Narayanan et al., 9000). 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 (Figure 1a). The Pavlik Harness (PH) (Figure 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 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 Ten-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 of 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 the iliac spines and acetabuli were aligned (Figure 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 orthopedic data found in Hensinger's Standards in Pediatric Orthopedics (Hensinger, 1986). 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 7.9mm, the

acetabular depth-width ratio (ADR) was 45%, and the acetabulum diameter was approximately 17.5mm.

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 (Figure 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° (Figure 2c) .

Once solid models were completed for the hip, femur, tibia, fibula, and foot, Solidworks is used to assemble all the components (Figure 3a). To assemble the femur to the pelvis origins and orientations in both bones were specified as defined by Dostal and Andrews (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 tissue is modeled using a straight-line muscle path representation since lines of action do not intersect in the range of motion of interest, allowing to use classical methods of vector analysis (Dostal and Andrews, 1981). The Adductor Magnus is a large triangular muscle with extensive femoral insertion; consequently, it is commonly represented by three effective components (Dostal and Andrews, 1981): (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 (Figure 3b).

Once the model is assembled, the hip can be computationally dislocated to match physiological dislocations according to the DDH dislocation grade under investigation (Figure 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 constriction of muscles fibers that generate force and bear load, while passive components are due to the stretching of muscle fibers, the basal 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, towels 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 (Fukiage 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 non-linear relationship between the differential Young's modulus and stress 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 Equation 1:

$$\overline{F^{PE}} = \frac{F}{F_o^M} \quad (1)$$

where $\overline{F^{PE}}$ is the normalized passive muscle force , F_o^M 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 Equation 2:

$$F_o^M = C (PCSA) \quad (2)$$

186 where C has a value of 25 N/cm² (Delp, 1990) and was estimated to scale PCSAs that were
 187 measured on young adult cadavers to match moment curves measured on young subjects.
 188 Furthermore, Thelen (Thelen, 2003) reports that $\overline{F^{PE}}$ can be expressed as:

$$\overline{F^{PE}} = \frac{e^{K^{PE}(\lambda-1)/\varepsilon_o^M} - 1}{e^{K^{PE}} - 1} \quad (3)$$

189 where K^{PE} is an exponential shape factor with a value of 4 obtained from OpenSim (Delp et al.,
 190 2007), ε_o^M is the passive muscle strain with a value of 0.60 for young adults, and λ is the optimal
 191 muscle-fiber length (stretch). The stretch can be written as $\lambda=L/L_0$ with L being the deformed
 192 length, and L_0 being the initial muscle length at rest measured at 20° abduction and 120° flexion
 193 for newborns. This flexion and adduction ‘neutral’ configuration accounts for the fact newborns
 194 maintain a posture similar to the fetal positioning for the first few months after birth with the
 195 hips flexed in toward the belly and wide apart. This initial muscle length at rest is also multiplied
 196 by a specified muscle pre-stretch for the lower extremity muscle and is usually between 20-
 197 30%. By combining Equations 1, 2, and 3, an expression can then be derived to obtain passive
 198 muscle force as function of stretch:

$$F = PCSA[a(e^{b(\lambda-1)} - 1)] \quad \left\{ \begin{array}{l} a = \frac{C}{e^{K^{PE}} - 1} = 0.466 \frac{N}{cm^2} \\ b = \frac{K^{PE}}{\varepsilon_o^M} = 6.667 \end{array} \right. \quad (4)$$

199 Using MRI data, a TWOFI PCSA of 0.41cm² for the adductor brevis was estimated by measuring
 200 the volume of the muscle and dividing it by its average length. A shape factor was then defined,
 201 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 Equation 5:

$$\underline{\underline{M}} \ddot{\underline{q}} + \underline{\underline{\beta}} \underline{\underline{\phi}}_q^T - \underline{\underline{A}}^T \underline{F}(\underline{q}, \dot{\underline{q}}) = 0 \quad (5)$$

are numerically solved subject to the natural constraints $\underline{\underline{\phi}}(\underline{q}, t) = 0$. The system mass matrix is $\underline{\underline{M}}$, the vector of generalized coordinates for the displacements is $\underline{q}$, the Lagrange multiplier adjoining the gradient constraints $\underline{\underline{\phi}}_q$ is $\underline{\underline{\beta}}$, the set of configuration and applied motion constraints is denoted by $\underline{\underline{\phi}}$, the matrix $\underline{\underline{A}}^T$ projects the applied forces in the $\underline{q}$ direction, and $\underline{F}$ is the vector of applied forces. The governing equations of dynamics described in Equation 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 Equation 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 1×10^{-4} , 1×10^{-7} , and 1×10^{-2} . 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 articular 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 Equation 4 with their respective PCSAs.

2.4 *Equilibrium Reference for Model Calibration*

Since the passive muscle model developed previously was for a young adult, the constants ***a*** and ***b*** in Equation 4 needed to be recalibrated for a TWOFI and also to stiffen the exponential function to account for ligaments that start acting at the limit of ranges of motion, namely, the iliofemoral, pubofemoral and ischiofemoral ligaments which help to stabilize the hip joint. The muscle model was calibrated by calculating the factors that allowed to precisely replicate the clinical observation that equilibrium is attained when infants are supine and the hips are placed in abduction at 80° and 90° flexion. That is, in this position, the passive action of the muscles holds the legs in place. Constants ***a*** and ***b*** were tuned in this model until the equilibrium configuration was achieved by only using muscle passive tensions. Constant ***a*** was found to be 0.0337 N/cm² and ***b*** was found to be 13.95. It is important to mention that there is no unique set of values for ***a*** and ***b***. However there is a minimum value requirement for ***b***, which is needed to stiffen the curve in order to achieve physically realistic equilibrium at 80° abduction. The constants ***a*** and ***b*** are inversely related; while ***b*** needs a minimum value to allow the model to achieve realistic muscle behavior, ***a***, which is used to find final equilibrium, is found through iteration once ***b*** has been established. The passive muscle force behavior for our calibrated TWOFI model is presented in Figure 4a where it is observed that the exponential behavior becomes very pronounced after a lambda (λ) value of about 1.35. Our equilibrium model

stiffens significantly after this value due to the previously mentioned hip joint ligaments that prevent any further abduction of the legs past an angle of 80° . Figure 4b also shows a comparison of the force-stretch behavior between our TWOFI calibrated model and the young adult of the adductor brevis where it is clearly seen our calibrated model stiffens faster once it gets close to the limits of ranges of motion as expected. After a close comparison between all the adductor muscles, it is found that all muscles display the same behavior as the brevis.

Figure 5a and Figure 5b provide the x-y and x-z views at equilibrium with reaction force(s) imparted at the contact(s) of the femoral head and the acetabulum shown as single headed (yellow) arrows whose magnitude is proportional to the size of the arrow. Muscle force and moment contribution at equilibrium clearly show that the Pectineus and the Adductor Brevis are the main muscle contributors (Table 2).

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 IHDl (Figure 6a and Figure 6b). 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 (Figure 6c) and the center of the femoral head was placed at $x = -3.44\text{mm}$, $y = 1.14\text{mm}$, $z = 6.59\text{mm}$ 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 $x = -10.05\text{ mm}$, $y = 12.69\text{ mm}$, and $z = 1.87\text{ mm}$ relative to the origin located at the center of the acetabulum (Figure 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 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 (Figure 7b). This path of reduction is generally consistent with clinical studies performed to avoid surgical release of adductor muscles by Papadimitriou (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 TWOFI. The model developed allows for the detailed study of the biomechanics of DDH treatment with the PH. Results suggest muscle force contributions on hip reduction are functions of severity of hip dislocation. Our findings also strongly indicate muscle tensions contribute to reduction with the harness for subluxated and intermediate dislocations (Grades 1-3). It is also notable that the Pectineus and Adductor Brevis muscle force contribution favors reduction significantly in Grades 1-3 while it disfavors reduction in Grade 4. The latter dislocations, which have been considered irreducible by most physicians, could be reduced by using a hyperflexion method. Further optimization of PH configuration may help improve DDH treatment effectiveness. This multi-physics computational approach is consistent with clinical observations reported in the literature and may potentially provide physicians means to determine modifications to evaluate treatment that could lead to new insights for non-surgical management of DDH by passive reduction with the PH or in combination with other orthosis. A limitation of this study is the absence of abductor muscles as well as a fixed anteversion angle of 50°, which may influence total force and moment contribution. Plans to include additional lower limb muscles, the effect of different anteversion angles, and the effect of cartilage deformation, are contemplated for future studies.

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Figure 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.

Figure 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).

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433 Figure 3. a) Reconstructed lower limb from infant and VHP CT-scans and MRIs, b) Adductor

434 muscles origin and insertion points, c) Healthy Hip (Left), Grade 3 (Center), and Grade 4 (Right).

Figure 4. a) TWOFI passive force versus stretch for calibrated model, b) Adductor brevis force-stretch comparison between TWOFI and young adult models.

Figure 5. a) Contact Forces (single headed arrow) and muscle lines of action (double headed arrow) on xz-plane, b) Contact Forces (single headed arrow) and muscle lines of action (double headed arrow) on xy-plane.

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443 Figure 6. a) Grade 1 initial dislocation achieved reduction in TWOFI, b) Grade 2 initial dislocation

444 achieved reduction in TWOFI, c) Grade 3 initial dislocation achieved reduction in TWOFI.

Figure 7. a) Grade 4 dislocation on TWOFI (xz plane), b) Grade 4 dislocation on TWOFI (yz plane).

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451 List of tables included in the manuscript listed in order of appearance.

452 Table 1. Lower extremity mass distribution and scaled TWOFI PCSAs using shape adductor

453 brevis factor (f) (Friederich and Brand, 1990)

Total Body Mass (BM) 7.26 kg			f = 0.036	Friederich PCSA (cm ²)	TWOFI scaled PCSA (cm ²)
Limb	%BM	Mass (kg)	Pectineus	9.03	0.321
Thigh	9.45	0.69	Adductor Brevis	11.52	0.410
			Adductor Longus	22.73	0.809
Leg	4.2	0.30	Adductor Magnus Minimus	25.52	0.908
			Adductor Magnus Middle	18.35	0.653
Foot	1.35	0.10	Adductor Magnus Posterior	16.95	0.603
Full leg	1.5	1.09	Gracilis	3.73	0.133

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455 Table 2. Muscle Force and Moment Contribution at equilibrium configuration.

	λ (Stretch)	Total Force (N)	X-Moment (Nm)	Y-Moment (Nm)	Z-Moment (Nm)
Pectineus	1.710	45.84	0.2834	0.3328	0.0738
Adductor Brevis	1.523	15.01	0.1353	0.1342	-0.0165
Adductor Longus	1.422	1.35	0.0138	0.0028	-0.0006
Adductor Magnus Minimus	1.392	1.83	0.0189	0.0013	-0.0014
Adductor Magnus Middle	1.316	0.84	0.0085	-0.0013	-0.0012
Adductor Magnus Posterior	1.229	0.33	0.0031	-0.0030	-0.0007
Gracilis	1.357	0.33	0.0040	0.0016	-0.0010

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460 Table 3. Muscle Force and Moment Contribution on Grade 3 dislocation and Grade 4
461 dislocation.

Grade 3 Dislocation	λ (Stretch)	Total Force (N)	X-Moment (Nm)	Y-Moment (Nm)	Z-Moment (Nm)
Pectineus	2.009	2967.04	17.4055	20.1682	4.6945
Adductor Brevis	1.658	105.68	0.9486	0.8630	-0.0649
Adductor Longus	1.588	14.46	0.1479	0.0152	0.0052
Adductor Magnus Minimus	1.560	20.52	0.2135	-0.0082	0.0041
Adductor Magnus Middle	1.414	3.49	0.0359	-0.0083	-0.0026
Adductor Magnus Posterior	1.281	0.71	0.0068	-0.0071	-0.0011
Gracilis	1.417	0.77	0.0093	0.0032	-0.0020
Grade 4 Dislocation	λ (Stretch)	Total Force (N)	X-Moment (Nm)	Y-Moment (Nm)	Z-Moment (Nm)
Pectineus	2.153	2907.02	7.1963	12.1199	-0.3056
Adductor Brevis	1.729	79.65	0.4442	0.4848	-0.0833
Adductor Longus	1.667	11.78	0.0693	-0.0121	0.0175
Adductor Magnus Minimus	1.640	16.62	0.0974	-0.0394	0.0314
Adductor Magnus Middle	1.460	2.15	0.0162	-0.0075	0.0020
Adductor Magnus Posterior	1.299	0.39	0.0033	-0.0040	0.0001
Gracilis	1.432	0.42	0.0044	0.0014	-0.0009

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