Published in final edited form as:

Conf Proc IEEE Eng Med Biol Soc. 2019 July ; 2019: 1521–1524. doi:10.1109/EMBC.2019.8856444.

A Method to Quantify Multi-Degree-of-Freedom Lower Limb Isometric Joint Torques in Children with Hemiplegia

Vatsala Goyal,

departments of Biomedical Engineering and Physical Therapy and Human Movement Sciences at Northwestern University, Chicago, IL 60611, USA

Theresa Sukal-Moulton,

departments of Physical Therapy and Human Movement Sciences and Pediatrics at Northwestern University, Chicago, IL 60611, USA

Julius P.A. Dewald

departments of Physical Therapy and Human Movement Sciences and Biomedical Engineering at Northwestern University, Chicago, IL 60611, USA

Abstract

Pediatric hemiplegia, caused by a unilateral brain injury during childhood, can lead to motor deficits such as weakness and abnormal joint torque coupling patterns which may result in a loss of independent joint control. It is hypothesized that these motor impairments are present in the paretic lower extremity, especially at the hip joint where extension may be abnormally coupled with adduction. Previous studies investigating lower extremity isometric joint torques in children with spastic cerebral palsy used tools that limited data collection to one degree of freedom, making it impossible to quantify these coupling patterns. We describe the adaptation of a multijoint lower extremity isometric torque measurement device to allow for quantification of weakness and abnormal joint torque coupling patterns at the hip in the pediatric population. We also present preliminary data in three children without hemiplegia to highlight how the presence of atypical femoral bony geometry, often observed in childhood hemiplegia, can be accounted for in the Jacobian transformations and affect joint torque measurements at the hip.

I. INTRODUCTION

Pediatric hemiplegia, which can include individuals with a diagnosis of cerebral palsy (CP), is a common disability caused by a unilateral, non-progressive brain injury. Peripheral effects of this injury include motor deficits such as weakness and abnormal joint torque coupling patterns in the upper extremity [1]–[2]. It is hypothesized that lower extremity movement in this population may also be influenced by abnormal joint torque patterns, such as the coupling between hip extension and adduction reported following stroke in adulthood [3]–[5]. The hip is a critical joint for static and dynamic stability, but quantifying strength and motor coordination patterns are challenging using commercially available tools. Previous studies investigating lower extremity isometric torques in children with spastic

cerebral palsy have used handheld dynamometers in a variety of supine, prone, and sitting positions. While these methods are perhaps successful in optimizing torque generation, they have produced conflicting results about hip strength deficits [6]–[8]. Specifically, isometric hip extension strength in children with diplegia has been reported to be anywhere between 50%–90% of values from typically developing children across multiple studies [6]–[8]. Additionally, the use of handheld dynamometry limits data collection to a single degree of freedom (DOF), making it impossible to confirm abnormal joint coupling patterns and limiting the conclusions that can be made about changes in descending motor drive.

The use of multi-DOF measurement tools does offer a better approach. This paper will focus on the use of a multijoint lower extremity isometric torque (MultiLEIT) device [9] to quantify weakness and abnormal joint torque coupling patterns at the hip in children with hemiplegia. Indeed, prior studies have used this device to quantify extensor coupling patterns in the paretic lower limb of adults following hemiparetic stroke [4]–[5]. We will detail the modifications made to adapt the MultiLEIT for the pediatric population as well as describe changes to the lower limb mechanical model to account for the presence of atypical hip bony geometry reported in children with hemiplegia [10]–[13].

II. METHODS

A. MultiLEIT

The MultiLEIT, engineered by Sánchez, Stienen, and colleagues at Northwestern University, is a rigid structure comprised of two 6-DOF load cells which feed into a static mechanical model to simultaneously measure hip, knee, and ankle joint isometric torques across one leg [9]. Initially designed for adult stroke studies [4],[5],[9], the MultiLEIT has been further modified for use in the pediatric population (Fig. 1). For set-up, participants sit on a bicycle seat and are buckled into a trunk harness and pelvis supports to mechanically isolate the lower extremity from effects of trunk and pelvic movement. The trunk harness has been downsized to better fit children. The tested leg is connected to the top load cell using a thigh cuff with participant-specific cuff inserts for a secure fit. In contrast to the original design [9], the top load cell has been offset superiorly from the support bracket to increase thigh cuff position range and allow for testing of shorter legs. The tested leg is further grounded to the bottom load cell by casting the foot onto a rigid plate, providing a rigid coupling of the participant's foot. These accessories restrict participants to isometric efforts. This removes spasticity as a possible confounding factor in the recorded results. The position of the tested leg can be adjusted with the support brackets, which can move in three directions. Support brackets can also be unmounted and switched to allow for testing of both lower limbs. A motor is attached to the system such that the entire frame can rotate back into a more supine position. During testing, the non-tested leg is supported with elastic cords to prevent neural and mechanical contributions to the tested leg. The slings for the non-tested leg are attached to the ceiling, as participants are reclined 66° from vertical to maximize comfort and reduce the impact of body support on the tested leg.

The MultiLEIT uses two 6-DOF 75E20A4 load cell sensors (JR3, Inc., Woodland, CA). These sensors output raw voltages to a data acquisition unit (National Instruments, Austin, TX) connected to a graphical user interface in MATLAB (MathWorks, Inc., Natick, MA).

Raw voltages are converted to raw forces and torques in the XYZ directions using factory-specified calibration matrices for each load cell. This data is then input into a lower limb mechanical model to output joint forces and torques of the hip, knee, and ankle simultaneously. The model assumes a static system which represents the hip as a ball-and-socket joint and the knee and ankle as hinge joints. Fig. 2 shows the free body diagram for the hip joint in the sagittal and frontal planes, with the joint coordinate frame adapted from ISB recommendations [14].

Joint torques at the hip are calculated by Jacobian transformations of raw data from both load cells (Eq. 1). Inputs to the equation include thigh and shank lengths, offset matrices d_i representing the distance between the top/bottom load cells

$$\begin{aligned}
 \begin{bmatrix} H_x \\ H_y \\ H_z \\ \tau_{hx} \\ \tau_{hy} \\ \tau_{hz} \end{bmatrix} &= \begin{bmatrix} R_z(-\theta_{hip}) & \text{zeros}(3 \times 3) \\ \ell_{thigh}(3 \times 3) & R_z(-\theta_{hip}) \end{bmatrix} * \begin{bmatrix} R_x(\theta_{add}) & \text{zeros}(3 \times 3) \\ d_{top}(3 \times 3) & R_x(\theta_{add}) \end{bmatrix} * \begin{bmatrix} F_{x_t} \\ F_{y_t} \\ F_{z_t} \\ \tau_{x_t} \\ \tau_{y_t} \\ \tau_{z_t} \end{bmatrix} \\
 &+ \begin{bmatrix} R_z(-\theta_{knee}) & \text{zeros}(3 \times 3) \\ \ell_{thigh}(3 \times 3) & R_z(-\theta_{knee}) \end{bmatrix} * \begin{bmatrix} R_z(\theta_{knee} - \theta_{hip}) & \text{zeros}(3 \times 3) \\ \ell_{shank}(3 \times 3) & R_z(\theta_{knee} - \theta_{hip}) \end{bmatrix} \\
 &* \begin{bmatrix} R_x(\theta_{add}) & \text{zeros}(3 \times 3) \\ d_{bot}(3 \times 3) & R_x(\theta_{add}) \end{bmatrix} * \begin{bmatrix} F_{x_b} \\ F_{y_b} \\ F_{z_b} \\ \tau_{x_b} \\ \tau_{y_b} \\ \tau_{z_b} \end{bmatrix} \quad (1)
 \end{aligned}$$

and the knee/ankle joint coordinate frames respectively, and rotations of the joint angles from the controlled lower limb position during testing. $R_z(\theta)$ represents a rotation of hip or knee flexion angle in the sagittal plane (Fig. 2a) and $R_x(\theta)$ represents a rotation of hip adduction angle in the frontal plane (Fig. 2b). Transformed data is filtered with a 250ms moving average filter to smooth the data.

B. Experimental Protocol

To be a good candidate for this protocol, children must have a clinical presentation of hemiplegia, have ambulatory function, and have imaging on file to confirm unilateral brain injury diagnosis and injury timing. Exclusion criteria include botulinum toxin injections in the past 6 months and neuro or musculoskeletal surgery in the past year in the lower extremity to control for effects on gross motor function. Further exclusion criteria include serious comorbidities, which may make it unsafe to participate, and severe cognitive dysfunction, which may make it difficult to follow experimental instructions.

First, thigh and shank lengths are measured to input into the Jacobian (Eq. 1), and then participants are helped into the Multi-LEIT. The controlled position of the leg in the sagittal

plane is 30° hip flexion, 30° knee flexion, and 10° ankle plantarflexion (Fig. 2a). Previous adult stroke studies positioned the tested leg in the frontal plane at 10° hip abduction [4]–[5]. Because children with hemiplegia may have added joint tightness in this direction, the tested leg is instead positioned at 5° hip adduction for comfort (Fig. 2b). Distances between the center of the top and bottom load cells and the center of the knee and ankle joints respectively are measured to input into the Jacobian (Eq. 1). The MultiLEIT frame is then rotated back in supine to optimize isometric torque generation at the hip.

To evaluate strength of the lower limb in this position, maximum voluntary torques (MVTs) are collected first.

$$WR = \frac{\tau_{\max/ND} - \tau_{\max/D}}{\tau_{\max/ND} + \tau_{\max/D}} \quad (2)$$

Participants are instructed to generate MVTs in hip flexion/extension or hip abduction/adduction, one direction at a time. Feedback is displayed on a monitor by a rotating dial, with counter-clockwise rotation associated with hip flexion/abduction and clockwise rotation associated with hip extension/adduction. The outcome measure, a weakness ratio, is quantified by comparing single task MVTs between the non-dominant (ND) and dominant (D) legs to compare strength between the two (Eq. 2). Ratios closer to -1 indicate a weaker non-dominant limb and ratios closer to 1 indicate a weaker dominant limb. Participants are then asked to perform a dual task to quantify movement limitations that may arise from abnormal joint torque coupling patterns. The dual task combines one of three submaximal levels of hip extension (20%, 40%, 60%) with maximal hip abduction torque generation [4]–[5], a combined movement that is outside the typical coupling pattern of hip extension and adduction observed in pediatric hemiplegia. Feedback is displayed on a monitor with the same dial, which moves vertically to a target in the submaximal hip extension DOF and uses the rotating dial to display maximal hip abduction torque generation. The outcome measure is hip abduction torque, normalized to the max, during submaximal hip extension generation.

C. Advancements to the Lower Limb Model

Children with hemiplegia may have significantly increased femoral anteversion (FA) in their paretic limb [10]–[11]. The formal definition defines femoral version “as the angle between an imaginary transverse line that runs medially to laterally through the knee joint and an imaginary transverse line passing through the center of the femoral head and neck” [12], with anteversion being an excessive angle that rotates the femoral neck forward (Fig. 3a). At birth, FA is present in infants between 30-40°, and this angle progressively decreases during typical development [13]. Conversely, factors such as delayed independent walking, lower limb hypertonicity, and abnormal joint loading strategies can contribute to excessive FA past infancy. As a compensatory mechanism to prevent hip dislocation, children with increased FA may internally rotate the femur of their paretic limb to maintain the femoral head within the pelvis acetabulum [10]. This increase in hip internal rotation may alter the orientation of the hip joint coordinate frame, which can be represented in the mechanical model by multiplying $R_y(\theta_{IR})$, a y-rotation of hip internal rotation angle, to $R_x(\theta_{add})$ in the Jacobian (Eq. 1).

III. RESULTS

To validate this method for the pediatric population, we evaluated both the dominant (D) and non-dominant (ND) lower limbs of three female participants without hemiplegia, ages 6.1 years, 13.2 years, and 13.9 years. We also explored the effects of FA on outcome measures. The mean FA angles of the participants were 5° for the dominant leg and 8° for the non-dominant leg. A previous study using a biplanar X-ray technique quantified a mean of 25° from a sample of ambulant children with CP [15]. Thus, we took the difference between the sample mean and the mean FA angle in both limbs to calculate a resultant compensatory hip internal rotation angle of 20° for the dominant leg and 17° for the non-dominant leg as an input to $R_j(\theta_{IR})$.

The participants had comparable strength between legs, with a maximum average difference of 0.063 in the hip flexion direction (Table 1). The addition of FA into the model only slightly altered the weakness ratios, and it did change the direction with the maximum average difference of 0.057 in the hip abduction direction (Table 1). All participants were able to successfully complete the dual tasks, supporting the hypothesis that children without hemiplegia would not demonstrate abnormal coupling patterns (Fig. 4). The addition of FA into the model on average decreased the hip abduction torque generated, especially for the 60% hip extension submaximal level for the dominant leg.

IV. DISCUSSION

This paper detailed the device used in a protocol to measure joint weakness and abnormal joint torque coupling patterns at the hip in children with hemiplegia. Previous approaches quantifying lower extremity isometric joint torques restricted measurements to a single DOF by using handheld dynamometers and presented inconsistent results due to testing in various anti-gravity and gravity-neutral limb positions [6]–[8]. One study further determined that tracking hip strength changes over time is difficult with these methods [16]. The additional equipment and set-up time required using the MultiLEIT are offset by the simultaneous multi-DOF and high-resolution measurements recorded by this system. The true novelty of the approach is the ability to measure joint torques in multiple DOF concurrently, which allows us to quantify the abnormal joint torque couplings that have only been previously described in children with hemiplegia. The use of a static mechanical model also limits off-axis contributions to joint torque measurements. Improving measurement resolution may make the distinction between joint torque coupling patterns, and by inference descending motor control, more detectable in the pediatric population. The system could further offer a sensitive measure of torque coupling pattern changes to evaluate the impact of current or novel therapeutic interventions.

We showed preliminary data in three individuals without hemiplegia to highlight how the addition of femoral anteversion may affect joint torque measurements at the hip. Prior studies have investigated the effects of considering this skeletal malalignment on hip moment arm lengths and kinematic models of gait in children with CP. The researchers showed that generic models overestimated moment arm lengths in hip flexion/extension and abduction/adduction DOFs [17]. They also found that inclusion of FA reduced error in gait

models by more accurately defining joint coordinate frames based off of bony geometry [18]. Given the significant differences in femoral geometry in cerebral palsy and the differences seen in the outcome measures for this method, further development of the static mechanical model may be warranted.

V. CONCLUSIONS

This paper summarized the methods involved in quantifying weakness and abnormal joint torque coupling patterns at the hip in children with hemiplegia, which can reveal changes in neural control used to drive lower extremity movement in this population. This has the potential to increase current understanding of lower extremity discoordination and impact the rehabilitation treatments for pediatric hemiplegia.

ACKNOWLEDGMENT

We would like to thank Sabeen Admani for her assistance with technical issues and Nayo Hill for her guidance.

Research supported by NIH grants R01HD039343 and R01NS058667 to Julius P. A. Dewald

REFERENCES

- [1]. Sukal-Moulton T, Krossschell KJ, Gaebler-Spira DJ, and Dewald JPA, Motor Impairments Related to Brain Injury Timing in Early hemiplegia. Part II, Neurorehabilitation and Neural Repair, vol. 28, no. 1, pp. 2435, 2014.
- [2]. Sukal-Moulton T, Krossschell KJ, Gaebler-Spira DJ, and Dewald JPA, Motor Impairment Factors Related to Brain Injury Timing in Early hemiplegia, Part I, Neurorehabilitation and Neural Repair, vol. 28, no. 1, pp. 1323, 2014.
- [3]. Brunnstrom S, Movement therapy in hemiplegia: a neurophysiological approach. Hagerstown: Harper Row, 1970.
- [4]. Sánchez N, Acosta AM, Lopez-Rosado R, Stienen AHA, and Dewald JPA, Lower Extremity Motor Impairments in Ambulatory Chronic Hemiparetic Stroke: Evidence for Lower Extremity Weakness and Abnormal Muscle and Joint Torque Coupling Patterns, Neurorehabilitation and Neural Repair, vol. 31, no. 9, pp. 814826, 2017.
- [5]. Sánchez N, Acosta AM, Lopez-Rosado R, and Dewald JPA, Neural Constraints Affect the Ability to Generate Hip Abduction Torques When Combined With Hip Extension or Ankle Plantarflexion in Chronic Hemiparetic Stroke, Frontiers in Neurology, vol. 9, 2018.
- [6]. Wiley ME and Damiano DL, Lower-Extremity strength profiles in spastic cerebral palsy, Developmental Medicine & Child Neurology, vol. 40, pp. 100107, 1998.
- [7]. Eek MN, Tranberg R, Zgner R, Alkema K, and Beckung E, Muscle strength training to improve gait function in children with cerebral palsy, Developmental Medicine & Child Neurology, vol. 50, no. 10, pp. 759764, 2008.
- [8]. Thompson N, Stebbins J, Seniorou M, and Newham D, Muscle strength and walking ability in Diplegic Cerebral Palsy: Implications for assessment and management, Gait & Posture, vol. 33, pp. 321325, 2011.
- [9]. Sánchez N, Acosta AM, Stienen AHA, and Dewald JPA, A Multiple Degree of Freedom Lower Extremity Isometric Device to Simultaneously Quantify Hip, Knee, and Ankle Torques, IEEE Transactions on Neural Systems and Rehabilitation Engineering, vol. 23, no. 5, pp. 765775, 2015.
- [10]. Miller F, Cerebral Palsy. New York, NY: Springer Science Business Media, Inc., 2005.
- [11]. Robin J, Graham HK, Selber P, Dobson F, Smith K, and Baker R, Proximal femoral geometry in cerebral palsy, The Journal of Bone and Joint Surgery, vol. 90, pp. 13721379, 2008.

- [12]. Cibulka MT, Determination and Significance of Femoral Neck Anteversion, *Physical Therapy*, vol. 84, pp. 550558, 2004.
- [13]. Gulan G, Matovinovic D, Nemec B, Rubinic D, and Ravlic-Gulan J, Femoral Neck Anteversion: Values, Development, Measurement, Common Problems, *Collegium Antropologicum*, vol. 2, pp. 521527, 2000.
- [14]. Wu G, Siegler S, Allard P, Kirtley C, Leardini A, Rosenbaum D, Whittle M, DLima DD, Cristofolini L, Witte H, Schmid O, and Stokes I, ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motionpart I: ankle, hip, and spine, *Journal of Biomechanics*, vol. 35, no. 4, pp. 543548, 2002.
- [15]. Massaad A, Assi A, Bakouny Z, Sauret C, Khalil N, Skalli W, and Ghanem I, Three-dimensional evaluation of skeletal deformities of the pelvis and lower limbs in ambulant children with cerebral palsy, *Gait & Posture*, vol. 49, pp. 102107, 2016.
- [16]. Dyball KM, Taylor NF, and Dodd KJ, Retest reliability of measuring hip extensor muscle strength in different testing positions in young people with cerebral palsy, *BMC Pediatrics*, vol. 11, 2011.
- [17]. Scheys L, Campenhout AV, Spaepen A, Suetens P, and Jonkers I, Personalized MR-based musculoskeletal models compared to rescaled generic models in the presence of increased femoral anteversion: Effect on hip moment arm lengths, *Gait & Posture*, vol. 28, no. 3, pp. 358365, 2008.
- [18]. Scheys L, Desloovere K, Spaepen A, Suetens P, and Jonkers I, Calculating gait kinematics using MR-based kinematic models, *Gait & Posture*, vol. 33, pp. 158164, 2011.

Fig. 1.

The MultiLEIT device, with modifications for younger participants including a bracket to raise the top load cell, an elastic ceiling sling to test in a more supine position, and smaller straps to restrict trunk motion.

Fig. 2.

A free-body diagram representation of the right lower limb in the system, highlighting the hip. Load cell coordinate frames are drawn in blue and the hip joint coordinate frame is drawn in green. (a) View of the leg position in the sagittal plane, (b) View of the leg position in the frontal plane.

Fig. 3.

Femoral anteversion (FA) and its effects on the lower limb. (a) The top image shows typical anteversion at 10°, while the bottom image shows excessive anteversion at 25°(b)

Individuals with FA may internally rotate the femur as a compensatory mechanism, leading to increased hip internal rotation.

Fig. 4.

The effect of femoral anteversion (FA) on dual task data for three control individuals. The x-axis is hip extension submaximal level and the y-axis is hip abduction torque normalized to its maximum. The blue bars represent the dominant leg and the green bars represent the non-dominant leg. Light-colored bars are data from the original model (Eq. 1) and the dark-colored bars are data from the FA model. The FA model decreases normalized hip abduction values for both legs at all submaximal levels.

TABLE 1

AVERAGE WEAKNESS RATIOS FOR 3 PARTICIPANTS

	Original Model	FA Model
Hip Abduction	0.051 ± 0.06	0.057 ± 0.03
Hip Adduction	0.007 ± 0.05	-0.035 ± 0.11
Hip Flexion	0.063 ± 0.09	0.019 ± 0.10
Hip Extension	-0.008 ± 0.04	-0.013 ± 0.07