Muscle-skeletal model of the thigh: a tool for understanding the biomechanics of gait in patients with cerebral palsy

Emiliano Pablo Ravera, Paola Andrea Catalfamo Formento, Marcos José Crespo, Ariel Andrés Braidot

Laboratory of Biomechanics, National University of Entre Ríos, 3101, Oro Verde, Entre Ríos, Argentina.

E-mail: emilianoravera@bioingenieria.edu.ar

Abstract. Cerebral Palsy represents the most common cause of physical disability in modern world and within the pediatrics orthopedics units. The gait analysis provides great contributions to the understanding of gait disorders in CP. Giving a more comprehensive treatment plan, including or excluding surgical procedures that can potentially decrease the number of surgical interventions in the life of these patients. Recommendations for orthopedic surgery may be based on a quantitative description of how to alter the properties probably muscle force generation, and how this affects the action of the muscle to determine how these muscles, impaired by disease or surgery, contributing to the movement of the segments of the limb during crouch gait. So the causes and appropriate treatment of gait abnormalities are difficult to determine because the movements generated by the muscular forces of these patients are not clearly understood. A correct determination of the etiology of abnormal patterns of the knee is the key to select the appropriate therapy, presenting a major challenge at present since there is no theoretical basis to determine the biomechanical causes of abnormal gait of these patients. The potential and necessity of using correct biomechanical models that consistently study the abnormalities becomes clear. Reinforcing and correcting a simple gait analysis and eliminating the unknowns when selecting the appropriate treatment is crucial in clinical settings. In this paper a computer muscle-skeletal model is proposed. The model represents a person's thigh simulating the six most representative muscles and joints of the hip and knee. In this way you can have a better understanding of gait abnormalities present in these patients. So the quality of these estimates of individual muscle dynamics facilitate better understanding of the biomechanics of gait pathologies helping to reach better diagnosis prior to surgery and rehabilitation treatments.

1. Introduction

Normal gait is a functional movement involving the most efficient energy transfer. Similarities between subjects are common but also small individual variations, so any deviation detected by comparing gait patterns are of great value [1].

Cerebral Palsy (CP) describes a group of developmental disorders of movement and posture causing activity restriction or disability, which are attributed to disturbances occurring in the fetal or infant brain [2]. It represents the most common cause of physical disability in modern world and within the pediatrics orthopedics units [3] [4].

Gait analysis provides great contributions to the understanding of gait disorders in CP [5]. It allows an objective and accurate quantification of the causes of gait disorders originating from
musculoskeletal abnormalities in patients with CP and it demonstrates superiority to the visual or observational analysis of movement [5] [6]. It also provides a mean for a more comprehensive treatment plan, including or excluding surgical procedures that can potentially decrease the number of surgical interventions in the life of a patient with CP [7].

One of the most common abnormalities in children with CP is crouch gait, characterized by excessive knee flexion during the terminal swing phase and initial loading response and increased dorsiflexion during the whole gait cycle. Hip flexion may be similar to the normal pattern but often presents increased flexion and internal rotation during the complete gait cycle [8] [9] [10] [11].

Spasticity or static contracture of the hamstrings, are thought to cause the excessive knee flexion in many cases. Therefore crouch gait is commonly treated by surgical lengthening of the hamstrings, typically in combination with other orthopaedic procedures with the objective of diminishing the excessive knee flexion and improving the walking pattern [9] [11] [12] [13].

When crouched gait is caused primarily by contracture of the hamstrings, surgical lengthening may improve knee extension by decreasing the passive tension in the muscles [13]. When spasticity of the hamstrings contributes to the crouched gait pattern, surgical lengthening of the hamstrings may improve knee extension by attenuating the exaggerated response of muscles to stretch [14]. However, there are patients for whom the surgical procedure did not improve the knee flexion and it is thought that the cause of crouch gait on these patients may have been different to short or slow contraction of the hamstrings (related to spastic or otherwise affected hamstrings).

Studying the length and velocity of the muscle–tendon may help to identify patients who walk with abnormally short or slow hamstrings and would possibly benefit from surgery from those who walk with normal hamstring’s length and velocity of contraction [15].

Unfortunately, it is difficult to obtain a measure of the length and velocity of the hamstrings by simple inspection of kinematic data [16]. Hence, muscle-skeletal models are needed for calculation.

The models could improve recommendations for orthopaedic surgery, based on a quantitative description of how to modify muscle force generation and how these modifications affect the action of the muscles during crouch gait [17] [18].

However, the causes of and the appropriate treatment for gait abnormalities are difficult to determine because the movements generated by muscular forces of these patients are not clearly understood yet [19] [20]. Hence the explicit identification of the anomaly does not necessarily present a direct solution [21].

Due to the mechanical redundancy displayed by the human locomotor system, activities such as walking can be performed in different ways and in many combinations of muscle forces which can generate the same net joint moment. However, there are similarities in the muscle activation patterns (MAP) of different people performing the same well-learned task [22]. Experimental measurements of muscle forces during locomotion of animals showed that the MAP are in general stereotyped [23].

These consistencies suggest that the central nervous system uses specific principles for the control of individual muscle forces, and that the principles are the same for different persons. That leads to the hypothesis that the central nervous system selects optimal MAP with a set of criteria that are, as yet, unknown [22].

Assuming that the central nervous system minimizes stress on the muscles and the body, leads to the hypothesis that muscle strength may be found as the solution to an optimization problem [24]. However, it is thought that this crouch gait may be due to a variety of neuromuscular disorders, including abnormalities at the structure of hamstrings, which are not easily detectable.

A treatment plan is further complicated because there is no scientific basis to determine how the neuro-muscular-skeletal impairment contributes to the abnormal movement [25]. Abnormalities of the knee may be related to dynamic conditions such as spasticity, secondary static muscle contractures of the hip, knee and ankle, or a combination of short and spastic muscles.

A correct determination of the etiology of abnormal patterns of the knee is the key to select the appropriate therapy [26], presenting a major challenge at present since there is no theoretical basis to determine the biomechanical causes of abnormal gait of these patients [27] [28] [29].
At this point, the potential and necessity of using correct biomechanical models that consistently study the abnormalities become clear. Reinforcing and correcting a simple gait analysis and eliminating the unknowns when selecting the appropriate treatment are crucial in clinical settings [30]. In this study a computer model based on a simple rescaling method using subject-specific anthropometric data is proposed. The model is used to estimate the muscle tendon length and velocity of unimpaired subjects and patients treated for crouched gait with the objective of identifying those patients with abnormally short or slow hamstrings. It also estimates muscle forces of the six most representative muscles of the thigh using static optimization. It is believed that this will improve the understanding of gait abnormalities in patients with CP.

2. Methods

The musculoskeletal mechanical system proposed consists of the pelvis, femur, tibia and six muscles. Four of the muscles were considered two-joint muscles: Tensor Fascia Lata (TFL), Semimembranosus (SM), Sartorius (S) and Rectus Femoris (R) and two were modeled as monoarticular: Gluteus (G) and Iliac (I). The model has six degrees of freedom (DoF), three for the hip joint and three for the knee.

Estimations of the patches of origin and insertion and directions of the muscles, assumed as the linear direction between the origin and insertion, were used by the muscle-skeletal model for dynamic evaluation of the muscle behavior throughout the gait cycle. The estimations were obtained from a complete three-dimensional muscle–skeletal model of the lower limb table 1.

| Muscle | PCSA (cm²) | XO (%) | YO (%) | ZO (%) | XI (%) | YI (%) | ZI (%) |
|--------|-----------|--------|--------|--------|--------|--------|--------|
| TFL    | 5.9       | 11.4   | 26.6   | 12.5   | 3.0    | 15.2   | 20.8   |
| SM     | 28.9      | 22.21  | 18.2   | 0.0    | 21.3   | 17.7   | 3.7    |
| S      | 8.8       | 11.4   | 26.6   | 12.5   | 31.3   | 17.7   | 3.7    |
| R      | 17.1      | 11.4   | 13.3   | 6.3    | 35.2   | 0.0    | 0.0    |
| G      | 13        | ***    | ***    | ***    | 31.18(*) | 6.3(**) | 15.5(*) |
| I      | 60.8      | ****   | ****   | ****   | 31.18(*) | 8.3(**) | 21.5(*) |

(*) Our estimates.
(**) Proportions of the patient's height.
(*** ) Middle position between the anterior superior iliac spines and posterior.
(****) Position estimated 7% of the width of the pelvis behind the origin of the gluteus.

The position of the patch ($\vec{p}_{\text{origin}}$, $\vec{p}_{\text{insertion}}$), origin and insertion of the six muscles, were calculated using axis systems embedded on and linked to the movement, $[\vec{u}, \vec{v}, \vec{w}]_{\text{segment}}$, of the
pelvis and the leg respectively. Subject-specific anthropometric measures, $W_{\text{segment}}$, such as pelvis width and knee width were used for calculation. The origin of the muscles was then determined using equations (1) and (2).

$$
\mathbf{p}_o = \mathbf{u}_{\text{pelvis}} - W_{\text{pelvis}} \mathbf{w}_{\text{pelvis}} \cdot \mathbf{W}_{\text{pelvis}} \tag{1}
$$

$$
\mathbf{p}_{\text{origin}} = \mathbf{p}_{\text{hip}} + \mathbf{p}_o \tag{2}
$$

Similarly, the position of the insertions was calculated from the position of knee joint. The calculation used a weighted displacement linked to the movement of the leg. In this case, the weighting factor was the value of knee width of the patient (3).

$$
\mathbf{p}_i = \mathbf{u}_{\text{leg}} - W_{\text{leg}} \mathbf{w}_{\text{leg}} \cdot \mathbf{W}_{\text{knee}} \tag{3}
$$

Finally (4),

$$
\mathbf{p}_{\text{insertion}} = \mathbf{p}_{\text{knee}} + \mathbf{p}_i \tag{4}
$$

Kinematic data from thirty trials from unimpaired subjects (10–25 years of age, 1.50–1.94 m in height and 49–105 kg in mass) and twenty trials from patients with CP that presented crouch gait (8–23 years of age, 1.08–1.83 m in height and 28–79 kg in mass) were used for calculation. Data from both, healthy and CP populations were provided by the Gait and Movement Laboratory at FLENI Institute for Neurological Research. All subjects analyzed were older than eight years, then it is considered that everyone had a mature gait pattern [33].

2.1. Calculation of the length and velocity of shortening of hamstrings

Since changes in length and velocity of the semimembranosus and the semitendinosus muscles are similar during the gait cycle, semimembranosus was considered as representative of medial hamstrings and used for this analysis [13].

Hamstrings length is then calculated as the distance between muscle position of origin and insertion (as line segments [34]). Shortening velocity is calculated as the first derivative with respect to time of the length of the muscle [13] [17], using the 25-parameter coefficients proposed by Savitsky and Golay [35]. Finally, the length of the semimembranosus is normalized to the length of the femur [17].

2.2. Static optimization model

The static optimization problem is given by (5) where the activation and / or muscle strength are calculated for each sample independently [22]. Thus, muscle forces are constrained by two physiological limitations: muscles may provide only contraction forces and these forces are limited to a maximum [36] [22].

$$
\text{Min } G(f_i^{(M)}) \text{ subject to } C \cdot f = r \text{ } y_M(i) \geq 0 \text{ } i \in \{ 1, \ldots, n^{(M)} \} \tag{5}
$$

Where $C \cdot f = r$ is given by the net total torque of the six muscles actuating on the joints of the knee and hip (6) [36] [37]. Among the objective functions used in static optimization in gait, the use of polynomials criteria stands up (7).
These describe physiological conditions only if they are complemented by restrictions that prevent individual muscle forces from exceeding their physiological maximum when the external loads increase [24].

\[ G(\vec{f}^{(M)}) = \sum_{i=1}^{n} \left( \frac{f_i^{(M)}}{N_i} \right)^p \]  

(7)

In this paper we used a polynomial optimization function of order \( p = 3 \) [38] [31]. Where \( f_i^{(M)} \) represents the amplitude of force of the \( n^{(M)} \) muscles involved in the model and \( N_i \) is the normalization factors of the function, in this case the Muscle physiological cross-sectional areas (PSCA). The physiological behavior shown by this criterion is to minimize muscle fatigue [22].

3. Results

3.1. Model length and velocity of shortening of hamstrings

A \( t\)-Student analysis (\( p < 5\% \)) shows that the proposed muscle-skeletal model presented a better differentiation between patients and healthy populations in areas of increased hamstrings activation during walking.

Figure 2. Semimembranosus length. Normal (line), crouch gait (dash) and area of difference \( p > 95\% \) (gray bars) [30].

Figure 2 shows the mean and standard deviation of hamstring length for patients and healthy subjects. The figure shows larger differences between groups in the area of final swing and initial stance. Mean while there is a high probability of similarities (\( p > 95\% \)) between groups during the rest of stance phase.

Figure 3. Semimembranosus shortening velocity. Normal (line), crouch gait (dash) and area of difference \( p > 95\% \) (gray bars) [30].
Figure 3 shows the mean and standard deviation of the velocity of the hamstrings for both groups. The *t*-Student test showed a statistically significant difference (*p* > 95%) between populations in a large part of the gait cycle. The difference between groups is more notorious during middle swing, when the hamstrings reached the highest velocity of shortening.

3.2. Static optimization model

Figure 4 shows the forces of each of the six major muscles crossing the hip and knee joints for two gait cycles of three healthy subjects. It is possible to see that the patterns of force show a relationship between different trials of the same subject but there are not common patterns among different subjects. Force patterns found are similar to electromyographic recordings of the respective muscles reported in the literature [39].

![Figure 4. Module force patterns of two gait cycles of three subjects. Subject 1 (black line), 2 (gray line), 3 (dash).](image)

The net joint moments at the hip obtained using articulated Link Segment Model (LSM) and recalculated by estimating the torque of the individual muscle forces simulated by the model presented in this work are shown in figure 5.

![Figure 5. Net muscle moments of the hip in global coordinate system.](image)
Figure 6. Net muscle moments of the knee in global coordinate system.

Figure 6 shows net joint moments of knee using articulated Link Segment Model (LSM) and recalculated values by estimating the torque to the individual muscle forces simulated by this model.

It is possible to see that the net joint moments for both joints calculated using the proposed model are very similar to the calculated using LSM. The small differences seen may be due to the filtering process applied to muscle forces prior to the reconstruction of torque. Also, the proposed model does not consider the effect of meniscus and ligaments, which may have influenced the differences found at the knee.

4. Discussion

The hamstrings length curves (figure 2) allowed for an optimal discrimination between groups (patients and healthy subjects) especially considering the differences at the end of swing and initial stance. However, there is a high probability of similarity between groups ($p > 95\%$) during mid and late stance. These findings were expected since previous work [40] showed that hamstrings exhibit little activity during mid and late stance while they are stretched to their maximum length at the end swing and initial stance.

In the case of velocity of shortening (figure 3) the difference between groups was more notorious during middle swing, when the hamstrings reached the highest velocity of shortening [40].

It is possible to see that the use of hamstrings length as a standalone parameter may not be sufficient for studying this group of patients, since some of them developed crouch gait, with spastic but not short muscles. This is consistent with previous results [12] that showed that from a population of patients with crouch gait, 30% had short and spastics muscles while 35% had spastic but not short hamstrings.

In the present study, 40% of patients walked with short and spastics hamstrings and 45% presented only spastic hamstrings. The minor differences in percentage may be due to the relatively small sample size used in this study, though it is sufficient to outline the trend.

In the evaluation of a movement that requires muscles submaximal efforts, the use of optimization problems makes it possible to predict: (i) the reciprocal coactivation of monoarticular antagonists, (ii) synergistic co-activation of a two-joint articulation with their antagonists, (iii) simultaneous activation of muscles crossing the same joint, and (iv) the strong relationship between force and muscle activation as seen in figure 4 [31].

It also notes that achieved high concordance reconstruct the resulting net moments acting on the joints, leaving with the conclusion that the forces estimated by this model consistently approximates the actual forces acting. In this model, we observed that muscle forces for each person have a stereotyped MAP as reported in the literature [22] [23].
5. Conclusion
Patients with crouch gait normally walk with excessive flexion of the knee and hip [10] [12] [16]. It is thought that this abnormality may be due to a variety of neuromuscular disorders, including pathology of the hamstrings, which are not easily detectable. One possible solution is the implementation of biomechanical models.

The main obstacles faced by these models are that they are anatomically and physiologically incomplete, they may provide insufficient accuracy of the parameters and the appropriate validation remains a challenge.

This paper presents a model with six muscles and six DoF for the estimation of parameters of interest in the biomechanics of cerebral palsy gait such as length and velocity of shortening of hamstring and muscle strength with high concordance with existing models [24] [22] [23] and muscle activation.

A multi segment model was proposed in this study based on rescaling and adjusting for each patient using subject-specific anthropometric data. The model allowed for the discrimination between subjects with short, slow and normal hamstrings. It has the potential to provide information about muscle forces involved in walking. However, for a correct discrimination of forces in cerebral palsy patients further analysis regarding the most appropriate objective functions to use will be required since it is unlikely that minimization of metabolic cost would represent a real scenario for those patients. Ultimately the quality of these estimates of muscle dynamics facilitate better understanding of the biomechanics of gait pathologies helping to reach better diagnosis prior to surgery and rehabilitation treatments.

The main limitation of the proposed model is that the patient must not have undergone any orthopedic intervention in the leg which might affect the origin and insertion of the hamstrings.

The main advantage of the model, on the other hand, is that it does not require costly and time consuming magnetic resonance images for estimation of anatomical parameters.

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