Function and Disability Journal ISSN: 2588-6304
Original Article

A Comparison of Mechanical Leg Impedance of Lower Limb Joint in Cerebral Palsy Crouch Gait Children with Healthy Match Group

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Article Info

Received: 2018/07/16
Accepted: 2018/08/28
Published Online: 2018/09/27

DOI: 10.30699/fdisj.1.3.1.48

How to Cite This Article

Pourhoseingholi E, Farahmand B, Bagheri A. A Comparison of Mechanical Leg Impedance of Lower Limb Joint in Cerebral Palsy Crouch Gait Children with Healthy Match Group. Func Disabil J. 2018; 1 (3) :48-57

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ABSTRACT

Background and Objective: Cerebral palsy (CP) is the most common form of upper motor neuron lesions in the children, leaded to joints deviation and different patterns of walking, like the crouch gait. Therefore, these joints devastation as well as the compensatory mechanisms were altered the mechanical impedance and gait insufficiency in these patients. The aim of this study was to compare mechanical impedance of joints and limb (skeletal and muscular components) at different sub phase of the stance phase of gait in Cerebral palsy crouch gait children.

Methods: Twenty five children with spastic diplegia crouch gait Cerebral palsy and twenty five healthy controls walked at a self-selected comfortable speed. Kinetic and kinematic data were measured and analyzed during the loading response, the mid-stance, the terminal stance and the pre-swing sub phases of gait.

Results: According to our study the Cerebral palsy crouch gait group showed a significant decrease in the leg impedance but increase in the joint impedance during the early to mid stance phase. However, during the terminal stance the leg impedance was increased as a result of more contribution of the muscular component to achieve sufficient impedance, which required increasing the muscular demands for keeping the body posture against collapse.

Conclusion: The results of current study depicted that the Cerebral palsy crouch gait group relied more on the muscular than the skeletal components to achieve the required leg impedance. In addition, the more flexed hip and knee were increased the lever-arm length of the Ground Reaction Force (GRF) vector at the joint centers, thus increased the joint moments.

Keywords: Cerebral palsy, Crouch gait, Impedance, Inverted pendulum, Joint impedance

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Introduction

Crouch gait is a major sagittal plane deviation in Cerebral palsy (CP), characterized by excessive ankle plantarflexion and excessive knee flexion throughout the early to late stance phase (Miller, 2018; Young, Rodda, Selber, Rutz, & Graham, 2010).

The main reasons for this deviation are the poor neuromuscular control, altered in the joints impedance, impaired balance, altered the muscle physiology, skeletal deformities, excessive moments and forces in the muscles or the passive structures, restricted joint range of motion and limited hip, knee, and ankle range of motion, leaded to the major musculoskeletal limitations (Mélo, Guimarães, & Israel, 2017; Miller, 2018).

Nowadays, one of the most treatment programs in this deviation was focused on improving these musculoskeletal limitations, especially the mechanical impedance (Ma et al., 2019; Miller, 2018; Wren, Rethlefsen, & Kay, 2005).

The mechanical impedance incorporates information of both the kinematics and kinetics of the lower extremity joints in these patients (Trumbower, Krutky, Yang, & Perreault, 2009; Wang et al., 2015) and plays an important role in the control, the postural stability and the joint dynamics. This parameter was defined as the ratio of the GRF and the shortening of the lower extremity (leg length), incorporated to support the body against collapse during gait (Chavez-Romero, Cardenas, Rendon-Mancha, Vernaza, & Piovesan, 2015; Farley & Morgenroth, 1999; Trumbower et al., 2009).

The mechanical leg (hip and knee) impedance includes skeletal component, related to the forces transmitted through the joints, bones, and muscular component (Farley & Morgenroth, 1999; Wang et al., 2015), as well as the joint impedance, provided predominately moments at the joints (Rouse, Gregg, Hargrove, & Sensinger, 2012; Wei et al., 2009). However, for the ankle impedance there is no need to contribute the muscular component and it only depends on the ankle angle and moments which do not support the limb against collapse in gait (Chavez-Romero et al., 2015; Davis & DeLuca, 1996; Galli, Rigoldi, Brunner, Virji-Babul, & Giorgio, 2008; Lark, Buckley, Bennett, Jones, & Sargeant, 2003).

Therefore, in an extended lower extremity, the leg impedance depends only on the skeletal component. While as the lower extremity goes to a flexed posture, the dependence on the muscular component will be increased, and the dependence on the skeletal component was decreased (Wang et al., 2015).

Therefore, it seems, the mechanical lower extremity impedance would be a valid parameter in interpreting the potential of adjusting these components in the different sub phase of gait. This impedance is an important parameter in predicting the potential of deficiency progression and treatment response as well as improving the design and fabrication of the lower limb orthosis to cope with the needs of CP with crouch gait (Farley & Morgenroth, 1999; Wang et al., 2015).

In our knowledge, no study interpreted the mechanical impedance and its component as an index in the stance phase sub phases in CP and healthy children.

Therefore, the aim of this study was to compare mechanical impedance of joints and limb (skeletal and muscular components) at different sub phase of the stance phase of gait in Cerebral palsy crouch gait children.

Materials and Methods

Subjects

Twenty five children with a crouch gait cerebral palsy (age: 10±1.2 years; height: 132.2±11.81cm; mass: 32.0±8.8 kg), and Twenty five age-matched healthy controls (age: 10±1.1 years; height: 133.1±12 cm; mass: 33.4±9.2kg) participated in this study. Using Gross Motor Function Classification System (GMFCS) every CP participants were graded 1–3 in this system. All CP crouch gait have slightly limited passive range of motion (ROM) of the hip and ankle (Wang et al., 2015). Muscle strength evaluations depicted moderate to normal and mild to normal muscle tone without any pain and without leg length discrepancy, serious muscle contracture, joint deformity or other pathology which might affect gait and/or cognitive function. The healthy controls were free from any musculoskeletal, neurological or cardiovascular disorders. This study was approved by Iran university ethical committee . All the subjects and their parents/guardians were informed of the procedure and provided written informed consent prior to the study, including enrolment and data collection.

Protocol

Subjects walked bare foot at a self-selected normal place on a 10-meter walkway. Retro reflective markers were used for tracking the motions of the body segments in a Helen Hays manner. The acquisition of the kinematic data was made using Qualysis Motion Capture System (Qualisys workstation AB, Gothenburg, Sweden 2013), with six ProReflex cameras. The external forces' data was collected with Kistler force platforms (Kistler Holding AG, Winterthur, Switzerland, Model 9286B) which was utilized for obtaining the ground reaction forces (GRF) (Wang et al., 2015).

To calculate the joint torques, an inverse dynamics analysis was performed using the kinematic data and the ground reaction forces from the force platforms.

Before starting tests, subjects were allowed to walk on the walkway several times for better adaptation with the experimental environment.

Ten successful trials were captured in order to obtain reliable data from cameras and force plate.

As routine, we assumed the anatomical reference position for the ankle as zero; positive angle position for dorsiflexion and positive ankle torque for plantar flexion torque.

Data analysis

In current study, the ankle, knee and hip function were analyzed in detail depicting the angular positions, velocities and torques over time. The gait phases were determined using the force plate and marker data. Spatial-temporal parameters, walking speed, stride length normalized to LL, stride time, cadence and step width, were also obtained. Calculation of the impedance of every joint requires three major kinetic and kinematic variables: joint angle, joint angle rate of change and joint torque. Inverse dynamic model with GRF and kinematic data measurement, angular motions and internal moments of limb joints were calculated (Ancillao et al., 2017; Chien, Lu, Liu, Hong, & Kuo, 2014; Lark et al., 2003; Mélo et al., 2017). All the joint moments' calculation was normalized to body weight before starting test (BW .The rotational movements of each joint were described according to sequence (z-x-y) with the positive x-axis directed anterior, the positive y-axis right and the positive z-axis to the superior (Chen, Hsieh, Lu, & Tseng, 2011).

We consider the leg length (LL as the length between the anterior superior iliac spine and the medial malleolus. Legs behave like compression springs so that during the first half of the ground contact phase, LL decreases while the GRF increases, and during the second half of the ground contact phase, LL increases while the GRF decreases (Wang et al., 2015).

In the search for general principles underlying agile gaits, biomechanics have modeled the body as a linear mass spring supporting a point mass equivalent to body mass. The impedance of this spring, typically referred to as 'leg spring', was determined from the relationship between the magnitude of the GRF and the distance between the centre of mass (CM) and the centre of pressure (COP) on the ground (Wang et al., 2015).

In current study each of the lower limbs was modeled as a non-linear spring. Therefore, the leg impedance, Imp leg (t), at time t could be calculated as the gradient of the Fe(t) vs. Le(t) curve as follows (Farley & Morgenroth, 1999; Wang et al., 2015).

Imp Leg (t) =\frac{d}{dt}\frac{F_e}{L_e}= F1e(t) L1e (t)

F1e(t) = dFe(t)dt

L1e(t)= dLe(t)dt

Resemblance, the joint impedance of the ankle, Imp\ A(t), knee, Imp\ k(t), and hip, Imp H (t), at time t were calculated as the slope of the moment, M(t), vs. angle (α (t)) curve of the joint as follows:

Impj(t)=\frac{d}{dt}\frac{M}{\alpha}= M1e(t)\alpha1 (t)

M1t) = dM(t)dt

\alpha1e(t)= d\alpha(t)dt

Before being used to calculate the leg and joint impedance, Low-pass filter (50 ms) reduces the adverse effects of high frequency noise, Fe(t), Le(t), \alpha(t) and M(t) to smooth the leg and joint impedance (Chien et al., 2014; Rosenbaum et al., 2007).

The leg impedance was formed from two ingredients of skeletal and muscular components as follows:

Imp leg(t) = Imp leg-muscle(t) + Imp skeletal-leg(t) = Imp(t)sin2α(t) +Imp leg(t)cos2α(t);

Imp leg-muscle (t) = Imp muscle (t) sin α (t); Imp skeletal-leg (t) = Imp skeletal (t) cos α (t);

Imp muscle (t) = Imp leg (t) sin α (t); I skeletal (t) = I leg (t) cos α (t);

where α(t) is the angle between the longitudinal axis of the leg and the line joining the COP of the GRF; Imp skeletal-leg(t) is the skeletal component in Imp leg (t); Imp muscle-leg(t) is the muscular component in Imp leg (t); Imp skeletal (t) is the skeletal impedance; and Imp(t) is the muscular impedance (Wang et al., 2015).

A simplified leg model, has depicted the skeletal and muscular components of the leg impedance. Because of the minimal effects of the ankle joint position on α(t) the foot segment was not included in the model which helps us to have a simpler model. The skeletal component transmitted the load bone through its proximal and distal joints, related to the alignment between the bones, i.e., joint angles, instead of deformity or shape of the bones. When the limb is in completely extended position (α(t) = 0) lower limb only relies on the skeletal component.

As the angle (α) was increased into flexion, the lever arm length was increased, leading to the muscular component dependence of the leg impedance (Wang et al., 2015).

Calculating the ratio of the muscular and skeletal components also helped to assess their relative contribution. The means and standard deviations of kinematic parameters during the gait cycle for each subject were obtained. In both single and double support, the angle, moment and impedance at the hip, knee and ankle in the sagittal plane, COP and GRF data were measured for each limb during the gait cycle (Farley & Morgenroth, 1999; Wang et al., 2015).

Moreover by connecting the COP, line of the GRF of the limb and the hip joint center, we could evaluate the leg and the joint impedance as the slope of the force vs. Effective GRF (Fe(t)). It was considered as a force applied to the leg previously modeled as a spring, defined as the component of the GRF of the limb along the line joining the COP and the hip joint center (Wang et al., 2015).

The effective LL (Le(t)) defined as a length of the leg spring is defined as the distance between the hip joint center and the COP. Therefore, the leg Imp (t) =d GRF effective (t) d effective length (t) =Fe (t) Le (t)

Fe (t) = d GRF effective (t) DT

Le (t) = d Length effective (t) DT

Resemblance, the joint impedance of the ankle (IA (t)), knee (IK (t)), and hip (IH (t)) at time t were calculated as the slope of the moment (M (t)) vs. angle (θ (t)) curve of the joint as follows (DeVita & Hortobagyi, 2000): We hypothesized that α (t) is the angle between the longitudinal axis of the leg (shank and tight with some degree angle) and the line joining the COP of the GRF ;Imp leg skeletal (t) is the skeletal component in Imp leg (t); Imp muscle leg (t) is the muscular component in Imp leg (t); Imps (t) is the skeletal impedance; and Imp m (t) is the muscular impedance. By increasing angle (α) toward flexion, the lever arm length related to the muscular component. The ratio of the muscular and skeletal components was also help to assess the relative contribution of the two muscular and skeletal components. In both single and double support, the angle, moment and impedance at the hip, knee and ankle in the sagittal plane, COP and GRF data were measured for each limb during the gait cycle (Wang et al., 2015). We averaged the data from both limbs for each trial for and in each subject.

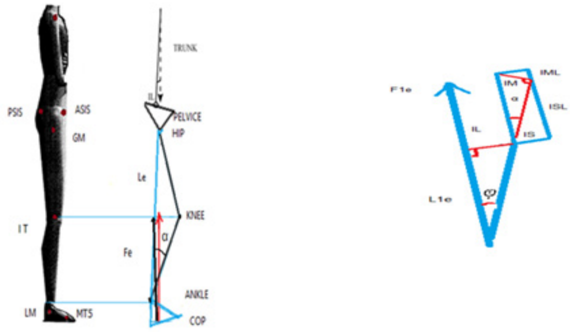

Figure 1. The mechanical and joint impedance

Statistical Analysis:

For each of the calculated variables independent t-tests were performed to compare each of the time-averaged values over the sub-phases between the CP and control groups. The gait speed relation with all impedance related variables was analyzed by finding correlation between the gait speeds as a confounding factor. All significance levels were set at α=0.05. All the statistical analyses were performed using SPSS 19.0 (SPSS Inc., Chicago, Illinois, USA).

Results

The result of current study showed that the speed of walking in these patients were 0.81(0.07) and in healthy match group were 1.22(0.11). However, a sig-

Table 1. Means (standard deviations) of the time-averaged angles and moments of the hip, knee and ankle in the sagital plane sub-phases of the stance phase of gait for children with CP and the health group.

	Loading response		Mid stance		Pre swing	
Hip	Angle (degree)	Moment (Nm)	Angle (degree)	Moment (Nm)	Angle (degree)	Moment (Nm)
Health	23.53(0.93)	0.02(0.01)	18.11(3.28)	0.002(0.005)	8.85(2.52)	0.005(0.0002)
CP	45.11(1.70)	0.05(0.08)	12.48(38.33)	0.005(0.003)	-9.20(12.5)	0.009(0.004)
P-value	0.000	0.000	0.000	0.000	0.000	0.000
Knee						
Health	12.45(0.18)	0.016(0.007)	15.21(0.38)	0.013(0.003)	12.62(0.46)	0.001(0.0003)
CP	35.82(1.11)	0.02(0.008)	32.27(1.25)	0.015(0.001)	32.27(1.51)	0.003(0.002)
P-value	0.000	0.000	0.000	0.000	0.000	0.000
Ankle						
Health	0.004(0.001)	0.005(0.001)	1.43(0.10)	0.008(0.004)	1.50(0.005)	0.008(0.002)
CP	0.0285(0.02)	0.007(0.003)	-3.39(0.16)	-0.01(0.01)	-8.41(0.006)	0.007(0.003)
P-value	0.00	0.000	0.000	0.000	0.000	0.000

nificantly increase was observed in this parameters in healthy match group than CP group ($P<0.05$). Furthermore, a strong correlation between impedance of all joints and speed with high Pearson coefficient ($P>0.05$) was observed. Comparison of Pearson correlation in CP and health groups showed a significant relationship between speed of walking and joint / leg impedance. According to the findings of the present study during the loading response, the mid stance and pre-swing sub phases at the hip knee flexion and ankle plantar flexion significantly increased in CP comparing to the healthy group. Furthermore, the hip, knee extensor and dorsiflexion moment at these sub phases significantly increased in CP compared to the healthy group (Table 1).

According to the findings of the present study the skeletal impedance, during the loading response was significantly decreased in CP group in comparison with the healthy group. However, the muscular impedance was significantly increased in CP group compared to the healthy group. Moreover, during this phase the leg impedance at the hip, knee and ankle joints were significantly reduced in CP group comparing to the healthy group. According to the findings of the present study, the skeletal impedance, during the mid stance significantly decreased in CP group compared to the healthy group. However, the muscular impedance significantly increased in CP group comparing to the healthy group. During this phase the leg impedance in the hip joint significantly increased in CP group compared to the healthy group and the knee and ankle impedance significantly decreased (Table 2).

As the finding of the present study showed, both skeletal and muscular impedance, during the terminal stance significantly increased in CP group than healthy group. During this phase the leg impedance in the hip and knee joints significantly decreased in CP group compared to the healthy group and the knee and ankle impedance significantly decreased. According to the findings of the present study, both skeletal and muscular impedance, during the pre-swing significantly increased in healthy group in comparison with CP group. In addition, during the pre-swing, the leg impedance, at the hip and knee joints was significantly reduced in CP group comparing to the healthy group. This impedance

Table 2. Means (SD) of the leg impedance, and skeletal and muscular components of it in Cerebral Palsy Crouch Gait Children compared to the health group in sub-phases of stance during walking.

Impedance	Loading response	Mid stance	Pre swing
Skeletal			
Health	14.71(0.43)	15.00(0.24)	3.96(0.13)
CP	7.39(0.39)	9.19(0.3)	2.05(0.52)
P-value	0.000	0.000	0.000
Muscular			
Health	13.2(0.39)	0.58(0.002)	1.11(0.13)
CP	8.23(0.43)	-0.96(0.006)	0.48(0.14)
P-value	0.000	0.000	0.000

Table 3. Means (SD) of the lower limb impedance in children with CP and the health group in sub-phases of stance during walking.

Limb Impedance	Loading response	Mid stance	Pre swing
Hip			
Health	0.0024(0.003)	0.0013(0.000)	0.0006(0.0001)
CP	0.001(0.004)	0.0019(0.000)	0.0004(0.0008)
P-value	0.000	0.000	0.000
Knee			
Health	0.0051(0.000)	0.001(0.0003)	0.0006(0.000)
CP	-0.002(0.0004)	0.003(0.0004)	0.0001(0.000)
P-value	0.000	0.000	0.000
Ankle angle(degree)			
Health	0.002(0.0016)	0.017(0.004)	-0.057(0.009)
CP	-0.002(0.000)	-0.019(0.003)	-0.014(0.02)
P-value	0.000	0.000	0.000

significantly increased at the ankle joint (Table 3).

Discussion

The aim of this study was to investigate the mechanical leg impedance and its components in Cerebral Palsy crouch gait children comparing to the healthy match group.

Our experimental and analytical methods and results of this study were presented for the estimation of the mechanical impedance of lower limb and the leg in these patients (muscular and skeletal component).

According to the findings of the present study, during the loading response the hip and the knee flexion, ankle plantar flexion significantly increased in CP crouch gait group comparing to the healthy group. Furthermore, the hip, knee extensor and dorsiflexion moments significantly increased, confirming the results of a study on CP spastic diplegia by Wang et al. (2015).

These more flexed postures of the hip, knee and the ankle plantarflexion following the loading response were contributed to the reduction in the leg impedance (skeletal component) and increased the need for the muscular components, as well as the joint impedance, by more sharing these impedances' parameters between the muscles, skeletal components and joints. However, in this time, the leg impedance was related to the muscular component more than the joint impedance (DeVita & Hortobagyi, 2000; Farley & Morgenroth, 1999; Trumbower et al., 2009). Therefore, the lack of necessary impedance, provided by the joint and skeletal components, increased the extensive muscle weakness and reduced the sufficient stability; therefore the risk of falling will dramatically increase (Trumbower et al., 2009; Wang et al., 2015).

The importance of considering these two impedances (joint and leg) becomes clear while the contra lateral limb was in the pre-swing sub phases and the body weight had to transfer to another limb, which increased the need for ipsi-lateral stability (Farley & Morgenroth, 1999; Trumbower et al., 2009). Hence, the safe and smooth transfer of the body weight and acceptable regulation of the leg and joint impedance between two reciprocal limbs during this period are vital for gait stability in these patients, placing further load on the flexed joint and leading to decrease strength, as well as, increasing joint impedance (DeVita & Hortobagyi, 2000; Lin, Guo, Su, Chou, & Cherng, 2000; Trumbower et al., 2009; Wang et al., 2015). However, in the health group, increased in the joints impedance provided sufficient stability in this sub phases (Wang et al., 2015).

According to the findings of the present study during the mid stance the hip, knee flexion, ankle plantar flexion and the hip and knee extensor and dorsiflexion moment significantly increased in CP group compared to the healthy group, confirming the results of other studies (Hicks, Schwartz, Arnold, & Delp, 2008; Trumbower et al., 2009). During this phase, CP group also relied on the muscular comparing to leg impedance to ensure the smooth progression of the body over the stationary foot, which had to be based on sufficient stability of the lower limb (Farley & Morgenroth, 1999; Trumbower et al., 2009).

Furthermore, the leg impedance increased by increasing the knee impedance. This strategy was used in the mid to terminal stance to prepare the limb for transferring the body weight to the contra lateral limb as well as preventing an undesired toe clearance of the contra lateral limb (Farley & Morgenroth, 1999; Lin et al., 2000; Wang et al., 2015).

According to the findings of the present study, during the pre-swing sub phase the hip, knee flexion, ankle plantar flexion, as well as, the hip and knee extensor and dorsiflexion moments significantly increased in CP in comparison with the healthy group, confirmed in a study on CP spastic diplegia (Wang et al., 2015). In addition, the leg impedance significantly reduced during the pre-swing (push off) sub phase, however the joint impedance increased at the hip and knee. It seems that reduction in the hip and knee impedance, as well as, increase in the ankle joint impedance in the CP group is a compromise for producing a push-off force, confirmed by other studies (Farley & Morgenroth, 1999; Lin et al., 2000; Wang et al., 2015). This more flexed hip and knee increases the lever-arm length of the GRF vector at the joint centers, which was placed further load on the joint by increasing the joint impedance (DeVita & Hortobagyi, 2000; Lin et al., 2000; Wang et al., 2015). Therefore, these patients exposed to a sudden increase in the GRF, the extra forces were needed to improve balance and reduced the risk of collapse.

This time, increasing the leg impedance as a strategy, may help the knee joint from becoming collapse, which was produced by the GRF, may lead to the larger stride length and the smoother body progression (Wang et al., 2015). Therefore, within the weight transferring from the stance limb to the contra lateral limb, the leg impedance and the resultant walking stability during this period would not to be enough for the safe and smooth walking (Wang et al., 2015). It reveals, the lower part of leg will be under variety of loads, throughout the stance phase (Farley & Morgenroth, 1999).

Additionally, muscles co-contractions, spasticity, contracture and joint deformity, would also change the lower leg impedance during the stance phase in these patients (Ficanha, Ribeiro, & Rastgaar, 2015; Rouse et al., 2012; Wang et al., 2015).

At last, joint impedance compare to lower limb impedance seems to be a creative part of our research and may help to describe complicated association between joint, skeletal and muscular components of lower extremity (Farley & Morgenroth, 1999; Wang et al., 2015). In addition, Stiffness-related variables may be considered as important indexes in prediction of treatment outcome and a more complete assessment in therapeutic planning as well as decision-making in prescription of suitable lower limb orthosis or occupational therapy planning like Botox injection, exercise training, casts or surgery in children with CP crouch gait.

Further research is needed to define the use of this parameter in providing planning the practice strategy in CP children with different levels of joint and leg impedance. Larger, randomized clinical trials are necessary to examine the impact of this parameter. Besides, the current study was limited to couch gait cerebral palsy and this parameter may be used in determining therapeutic decision-making of other paretic diseases.

Conclusion

Current study depicted that the leg and joint impedance and related variables could be used successfully as an index for distinguishing the stability, predicting the risk of falling and supporting the body against collapse in CP crouch gait. Additionally, the results of current study could be used as a guide for comparing the results from patients suffering from other form of spasticity like stroke or multiple sclerosis.

Acknowledgements

The authors wish to express their sincere gratitude to Miss Afsane Mohammad pour for their assistance throughout this study.

Conflict of Interest

The authors declared that they have no competing interests.

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Function and Disability Journal ISSN: 2588-6304
مقالۀ پژوهشی

مقایسه امپدانس مکانیکی اندام در کودکان فلج مغزی با کودکان سالم

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اطلاعات مقاله

تاریخ وصول: ۱۳۹۷/۰۴/۲۵
تاریخ پذیرش: ۱۳۹۷/۰۶/۰۶
انتشار آنلاین: ۱۳۹۷/۰۷/۰۵

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چکیده

زمینه و هدف: فلج مغزی یکی از شایع‌ترین آسیب‌های اعصاب مرکزی در کودکان است، که منجر به تغییرات مفصلی و الگوهای مختلف راه‌رفتن، مانند کراچ گیت است. بنابراین، به‌دلیل ایندر نتیجه، مفاصل همراه با مکانیسم‌های جبرانی، امپدانس مکانیکی مفاصل را تغییر داده که خود عاملی در جهت ناکارآمدی گیت در این افراد می‌شود. هدف این مطالعه، مقایسه امپدانس مکانیکی مفاصل و اندام‌ها (مؤلفه‌های اسکلتی و عضلانی) در ساب‌فازهای مختلف ایستایی در گیت بیماران فلج مغزی همراه با کراچ گیت بود.

روش کار: ۲۵ بیمار مبتلا به فلج مغزی با کراچ گیت و ۲۵ فرد سالم با سرعت انتخابی در محیط آزمایشگاه مورد بررسی قرار گرفتند. داده‌های سینتیک و سینماتیک در ساب‌فازهای پاسخ به‌بارگذاری، میانه ایستایی و پیش نوسان اندازه‌گیری و آنالیز شد.

یافته‌ها: بنابر نتایج این مطالعه امپدانس اندام در مبتلایان به فلج مغزی همراه با کراچ گیت در ساب‌فازهای ایستایی نسبت به گروه سالم کاهش معنی‌دار یافت، در حالی‌که امپدانس مفاصل از ابتدا تا میانه ایستایی افزایش معنی‌دار داشت. هرچند، در ساب‌فاز پیش نوسان امپدانس اندام در اثر دخالت بیشتر مؤلفه عضلانی برای رسیدن به امپدانس کافی افزایش یافت. در نتیجه نیاز به فعالیت عضلانی، به‌منظور حفظ پوزیشن بدن و جلوگیری از کولاپس افزایش یافت.

نتیجه‌گیری: در مبتلایان به فلج مغزی همراه با کراچ گیت، راه‌رفتن بیشتر با تکیه بر مؤلفۀ عضلانی نسبت به مؤلفه اسکلتی، به‌منظور امپدانس کافی در اندام است. به‌علاوه، قرار گرفتن زانو و هیپ در فلکشن بیشتر باعث افزایش طول بازوی اهرمی بردار نیروی عکس‌العمل زمین در مراکز مفاصل شد و در نتیجه ممان در مفاصل افزایش یافت.

واژه‌های کلیدی: فلج مغزی، کراچ گیت، امپدانس، پاندول معکوس، امپدانس مفصل