Original Article

The Effect of Spatial and Temporal Parameters on The Energy Expenditure of Orthotic Gait of Paraplegics

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Objectives: Although for those spinal cord injury (SCI) patients with paralysis of the legs but not at arms, the primary means of mobility post injury is the manual wheelchair, there are many physiological and psychological advantages to standing and walking, such as improvement in respiratory function, lower limb weight bearing and preventing osteoporosis, pressure sores prevention etc. High metabolic energy expenditure of orthotic ambulation in contrast with the manual wheelchair is the most limitation of the orthotic gait currency for SCI patients.

Methods: In the present study, the effect of the spatial and temporal parameters on the mechanical energy expenditure of the paraplegic locomotion was investigated in search for a modified gait pattern which provides lower energy Expenditure. A 3D four segment 6-DOF model of the paraplegic locomotion was developed based on the data acquired from an experimental study on a single spine cord injury subject. A Response Surface Methodology (RSM) was then performed to determine the impact of the kinematical parameters on the resulting the joint torques.

Results: According to the RSM analysis, the whole body forward inclination around the stance leg has been recognized as the most effective kinematics parameter on the total muscular effort as an index of energy expenditure during the paraplegic gait.

Conclusion: It was concluded that with a reduction on the step length accompanied with cadence increasing in a modified gait pattern, the paraplegic individuals are expected to achieve improved energy expenditure.

Keywords: Spinal cord injury, Orthotic gait, Kinematics & Dynamic Modeling, Energy expenditure

Introduction

Patients with spinal cord injury cannot activate the muscles below the level of the injury. Most of these patients use a wheelchair for mobility, but life in a wheelchair often results in a reduction of musculoskeletal functions, and it consequents such as joint contracture, muscle atrophy, spasticity, osteoporosis, and subsequent fractures of the lower extremities. Improving the musculoskeletal functions, loading the paralyzed lower extremities and expanding the relative joints' range of motion are important considerations to prevent secondary complications (1–3). So these patients are trained to walk and stand wearing an orthosis that enables standing and ambulation for short distances by restricting the movements of the paralyzed joints (spatially the knee joint). There are three major categories of accessory devices to help individuals with paraplegia to stand and walk, including mechanical orthoses, functional electrical stimulation systems, and a combination of both, i.e., hybrid orthoses. In all these devices the lower limbs are stabilized during the stance phase and some walking aids, e.g., crutches, are used to improve the stability and balance and provide a part of the propulsion required. Because of the external energy dependency for hybrid orthoses and muscle fatigue for electrical stimulation as the limitations of these systems, in this study, we focused on mechanical orthoses. A major concern in using the mechanical orthoses for ambulation of paraplegic individuals is the high energy expenditure involved. It has been

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reported that the energy Expenditure of walking of paraplegics can be as high as 3 to 9 times of that of the normal population (4, 5). There are two categories of investigators who try to improve the performance of the mechanical walking orthoses with respect to the energy expenditure, upper body loading, and user friendliness, e.g. easy donning and doffing. The first group of investigation based on trial and error efforts by adding and subtracting some links and joints to a pairs of knee-ankle-foot orthoses (KAFO), such as the reciprocating gait orthosis (RGO) or ParaWalkers, are among the most common types of orthoses (6, 7). In these two common types which have been used for high thoracic SCI, a pelvic band supports the trunk. However, it is difficult for patients to put on an orthosis by themselves and they require a significant amount of assistance. On the other hand, knee-ankle-foot orthoses with a medial-single-hip joint (MSH-KAFO) are orthoses without pelvic bands; these have the advantage that paraplegic patients can put them on by themselves while seated in a wheelchair (3, 8). MSH-KAFOs have been used with lower thoracic SCI because the paralyzed thoracic, lumbar, and hip joints are not restricted. To support their independent lives, reduced assistance is desirable. The efficiency and performance of the mechanical walking orthoses has been often evaluated using experimental tests (9, 10, 11). Such tests provide useful information on the kinematics, dynamics and energy Expenditure of the paraplegic gait when using different orthotic designs. However, it is often difficult or even impossible to achieve general conclusions from experimental data due to the small size of the test subjects available and the large interindividual differences between (12). Therefore the second group of investigation based on mathematical modeling of the paraplegic gait which can provide a better understanding of the role and significance of each individual relevant factor affecting the efficacy and performance of the orthosis. There are a few mathematical models of the paraplegic gait available in the literature. A pioneer work belongs to Tashman et al (13) who developed a 3D four segment model of the RGO-assisted paraplegic ambulation and showed that the upper body forces and body deceleration are reduced substantially if a ballistic swing could be produced using FES. Zefran et al (14) modeled the gait of a paraplegic individual when using crutches as a parallel robot and investigated the effect of different quadrupedal walking patterns on the efficiency of the gait. Greene and Granat (15) conducted a kinematical analysis on a simple 2D mathematical model of the paraplegic gait to demonstrate the importance of the ankle dorsiflexion accomplished with knee flexion for increasing the ground clearance. Kagawa et al, (16) addressed the relationship between the arm-crutch loading, leg restriction, and motor paralysis, and reported that large reaction forces occur at the crutches when the ankle and knee joints are restricted. Nakhaee et al (17) developed a simple 3D, three segments model with 5 DOFs to analyze the kinematic and dynamic behavior of the paraplegic gait and showed that an individual with low lesion level might be able to walk using appropriate mechanical orthosis and trunk maneuvers, without the need to a propulsion supply from the hands through crutches. The aim of the present study was to investigate the effects of the detail kinematical parameters of the gait pattern on energy expenditure of the paraplegic locomotion. A 3D four segment 6-DOFs model of a paraplegic subject wearing an ARGO was developed based on the data acquired from an experimental study. It was then analyzed using Response Surface Methodology (RSM) to recognize the most relevant kinematical factors affecting the total muscular efforts at the different instants of the swing phase. The results were employed to provide recommendations on how the energy expenditure can be improved during ARGO-assisted locomotion by modifying the gait pattern.

Materials and Methods

Experimental Study

Our primary purpose was to identify the simplest robotic model structure that reproduces the kinematics properties of the paraplegic gait. The geometrical and kinematical dataset needed for simulation of the model, as well as its validation, was obtained in an experimental study involving motion capturing of a paraplegic subject while wearing an Advanced Reciprocating Gait Orthosis and using two crouches.

Subject and Equipments

The paraplegic subject was a 28-year-old male, 96 kg weight, and 180 cm height, with the lesion level T10. The subjects underwent some stretch exercises for contracture of the leg joints and performed some orthotic gait trials before the measurement trials. He was informed of the objectives and procedures of the experiments and gave written consent for his participation. The mechanical orthosis which had used was an ARGO (Hugh Steeper Ltd, London, UK). ARGO is a version of HKAFOs in which the knee and ankle are locked and the hip joints are restricted to move only in the sagittal plane (Fig 1-a). Furthermore, this type of orthosis is equipped with a cable link which restricts the bilateral hip motion to a reciprocal action (Fig 1-b), so that flexion of a hip joint is coupled to the extension of the other (18).

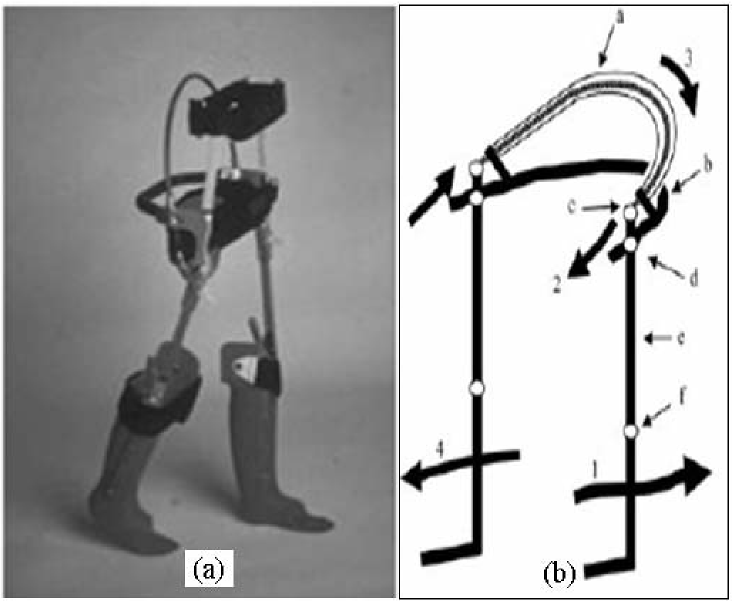

Figure 1. (a) Advanced Reciprocal Gait Orthosis. (b) Elucidation of the Reciprocal Cable functions in an ARGO structure.

During the tests, 24 reflective markers were fixed onto the subject anatomical landmarks, (Table 1). The subject was instructed to walk at his comfortable speed on a level surface, along a 7 m walking way (Fig 2). Twenty trials were recorded using a six-camera VICON motion analysis system (Oxford Metrics, UK) at 100 Hz. The 3D coordinate data of the markers was obtained in the Global Reference System (GRS), with the X, Y and Z being the coordinates in the forward/backward, left/right (media/lateral), and vertical directions, respectively.

Data Analysis

Among the recorded trials, 12 with the most complete tracking data were selected to be analyzed. The data was first filtered through a low-pass filter with a cutoff frequency of 5 HZ, to reduce the measurement noise. Also, the missing data, due to marker occlusion, was repaired using cubic spline interpolation. The 3-marker sets, attached to each of the 4 segments, were used to define a Local Reference System (LRS) for the respective segment. Each LRS was then employed to describe the rotations of the embedding segment with respect to the GRS, using Cardan sequence (X-Y-Z), and consequently the relative angles between the segments.

Figure 2. The test subject during ARGO-assisted walking

Table 1. Description, Name and the location of the markers

Description	Marker name	Placement
Left/Right Toe	L/R Toe	Center of the foot between the 2nd and 3rd metatarsals.
Left/Right Heel	L/R Heel	Posterior Calcaneus at the same height from the floor as the toe Marker on the Orthosis
Left /Right Lateral Ankle	L/R Ankle	Along the flexion/extension axis of rotation at lateral malleolus.
Left /Right Hip joint	L/R Hip	Along the flexion/extension axis of rotation at the Hip joint of the Orthosis.
Left /Right PSIS	L/R PSIS	Posterior Superior Iliac Spine.
Left /Right ASIS	L/R ASIS	Anterior Superior Iliac Spine.
T10	T10	10^{th} thoracic vertebra as the level of lesion.
C7	C7	7^{th} Cervical vertebra
Left /Right Shoulder	L/R Shoulder	Tip of the Acromion Process.
Left /Right Elbow	L/R Elbow	Lateral Epicondyle of the Humerus .
Left /Right Wrist	L/R Wrist	Centered between the Styloid Processes of the Radius and Ulna.
Left /Right Crutch	L/R Crutch	Tip of the Crutch.
Left /Right Lateral Ankle	L/R Ankle	Along the flexion/extension axis of rotation at lateral malleolus.
Top of the Head	Top-Head	On the center of top of the head.

Modeling

Kinematics Analysis

Considering the motion constraints imposed by the lesion and ARGO, a robotic model was developed for kinematics and dynamic analysis of ARGO-assisted gait (Fig 3). The four-segment model consists of the HAT segment representing head, arms and trunk (upper part of the lesion point), the middle segment representing pelvis and insensible part of the trunk (under the lesion point), and tow leg segments representing thighs, shanks and feet. Initially, there are 7 DOFs associated with the model: 3 DOFs between the stance leg and the ground (rotations around Z, X, and Y axes, i.e., $\theta_1, \theta_2, \theta_3$, respectively), 2 DOFs between the two legs and the pelvis (rotations around the single pin joints at the hips of the stance and swing legs, i.e., θ_p, θ_6, respectively), and 2 DOFs between the pelvis and the trunk (one in each of the sagittal and frontal plans, i.e., θ_4 and θ_5, respectively). However, considering the function of the reciprocal link of the ARGO, rotations of the two hip joints are coupled:

$$\theta_6 - \theta_3 = 2 \times \theta_p \qquad (1)$$

This constraint reduces the independent DOFs of the model to 6. These rotational DOFs of the model are controlled by 6 independent torque actions: 3 between the stance leg and the ground (τ_1, τ_2, τ_3), one between each leg and the pelvis ($\tau_p = -\tau_6$); due to the action of the reciprocal cable), and 2 between the pelvis and the trunk (τ_4, τ_5). The torques, that act on the 3 DOFs of the stance foot, are actually the resultants of all forces and moments that are applied to the upper body, due to its force interactions with the ground through crutches. In fact, they represent all the force/moment effects produced by the upper body for balancing and propulsion, as well as the partial body weight transfers to the ground.

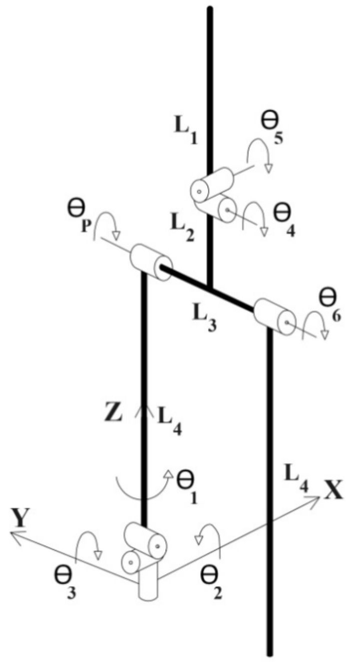

Figure 3. A four segment model, including trunk (L1), pelvis (L2 & L3), and left and right legs (L4) was used to describe the paraplegic ambulation.

The sufficiency of the proposed model to simulate the ARGO-assisted paraplegic gait was evaluated in two steps. At first, 3-marker sets were used to define a Local Reference System (LRS) for each of the segments. Each LRS was then employed to describe the rotations of the embedding segment with respect to the GRS, using common Cardan sequence (X-Y-Z). The relative angles between the segments were then obtained. In the next step, a forward kinematics analysis was performed to evaluate the sufficiency of the 6 DOFs considered for the model. The trajectories of a number of end effectors were obtained, based on Denavit-Hartenberg approach, and compared with those of the experiments. The end effectors considered, included C7, L-heel and R-heel (Table 1).

Dynamic Analysis

The modified Newton-Euler formulation for multi-body branched systems, considering the Anthropometric properties of the model (19), was used to obtain the model's nonlinear equations of motion as follow:

$$\tau_i = F_i\,(\theta_j, \dot{\theta}_j, \ddot{\theta}_j), \qquad i, j = 1, \ldots, 7 \qquad (2)$$

Where τ_i and θ_i represent the torque and angle of the i^{th} joint, respectively. According to the kinematics analysis an inverse dynamic analysis was also conducted to obtain the joint toques during a representative swing phase.

RSM Analysis

Response Surface Methodology (RSM) is a collection of statistical and mathematical techniques useful for developing, improving, and optimizing processes (20).The most extensive applications of RSM are in the particular situations where several input variables potentially influence some performance measure or quality characteristic of the process. Thus performance measure or quality characteristic is called the response. The input variables are sometimes called independent variables, and they are subject to the control of the scientist or engineer. The field of response surface methodology consists of the experimental strategy for exploring the space of the process or independent variables, empirical statistical modeling to develop an appropriate approximating relationship between the yield and the process variables, and optimization methods for finding the values of the process variables that produce desirable values of the response.

In the present study the response variable of interest is the energy Expenditure and the independent variables are the kinematical parameters of the gait pattern. They included all independent DOFs of the model, except for the swing leg hip flexion, θ_6, and the whole body axial rotation around the stance leg, θ_1, which were assumed fixed to keep the step length unchanged. For each of the design variables, i.e., $\theta_2, \theta_3, \theta_4, \theta_5$, a range of variation was considered, based on the kinematical results of the repetitive experimental tests. The energy expenditure was characterized based on the idea that the most efficient gait is obtained when a given distance is travelled with the lowest muscular effort. So, we used the total muscular effort (TME) index to characterize the energy expenditure. However, considering the fact that instead of individual muscles, the joint torque generators- which reflect the resultant effect of contributing muscles- were represented in our model, we used the sum of squared torques of all joints as an indicative of the TME index:

$$\text{TME index} \sum_{i=1}^{6} \tau_i^2 \qquad (3)$$

where τ_i represents the torque of the i^{th} joint which can be obtained from the equation (1). In order to simplify the RSM analysis, the joint torques was found assuming static conditions (i.e., $(\dot{\theta}_j = \ddot{\theta}_j = 0)$. Considering the 4 design parameters, 2^4=16 analyses were performed at each critical instant of the swing phase to obtain the relationship between the independent variables and the response function (TME). The results of the RSM analysis were expressed using linear regression equations (20):

$$f(\theta) = \beta_1 + \beta_2.\theta_2 + \beta_3.\theta_3 + \beta_4.\theta_4 + \beta_5.\theta_5 \qquad (4)$$

Where $f(\theta)$ represents a response function, and β_i s are called the regression coefficients which reflect the impact of each of the independent variables on the response function.

Results

Kinematics and Dynamic Analysis

The mean values of the spatial and temporal gait parameters obtained from the kinematics analysis compared with the normal one (21-22) in Table 2. This comparison showed that the stride length measured in the present study is slightly shorter than the normal walking. Cadence is almost 25% of that of normal gait and walking speed in the paraplegic gait is 20% of the speed of normal one. In the length of stance phase comparison, the higher percentage of the gait spent in the stance phase of impaired gait compared to that during normal walking reflected the faster walking speed of normal gait. The range of hip motions in the sagittal plan is almost similar. The lateral hip displacement in the frontal plane is greater in the paraplegic than for normal gait, presumably due to the need to tilt the orthosis for ground clearance of the swing leg. However, the vertical movement of the hip was not increased when walking in an ARGO.

Table 2. Spatial and Temporal Gait Parameters in the paraplegic gait and the normal one

Item	Experimental data for Paraplegic Gait (mean-Val)	Normal Gait (mean-Val) [ref]
Stride Length (m)	1.2045	1.4 [21]
Cadence (step/s)	0.48	1.93 [21]
Walking Speed (m/s)	0.28	1.37 [21]
Length of stance phase (% Gait Cycle)	66.34	60 [22]
Length of Double support phase (% Gait Cycle)	34.03	…
Hip angle in the sagittal plane-Flexion (deg)	17.75	20 [22]
Hip angle in the sagittal plane-Extension (deg)	36.45	30 [22]
Hip angle in the frontal plan-abduction (deg)	0.81	…
Hip angle in the frontal plan-adduction (deg)	10.6	…
Lateral hip displacement (m)	0.125	0.05 [21]
Vertical hip displacement (m)	0.058	0.05 [21]

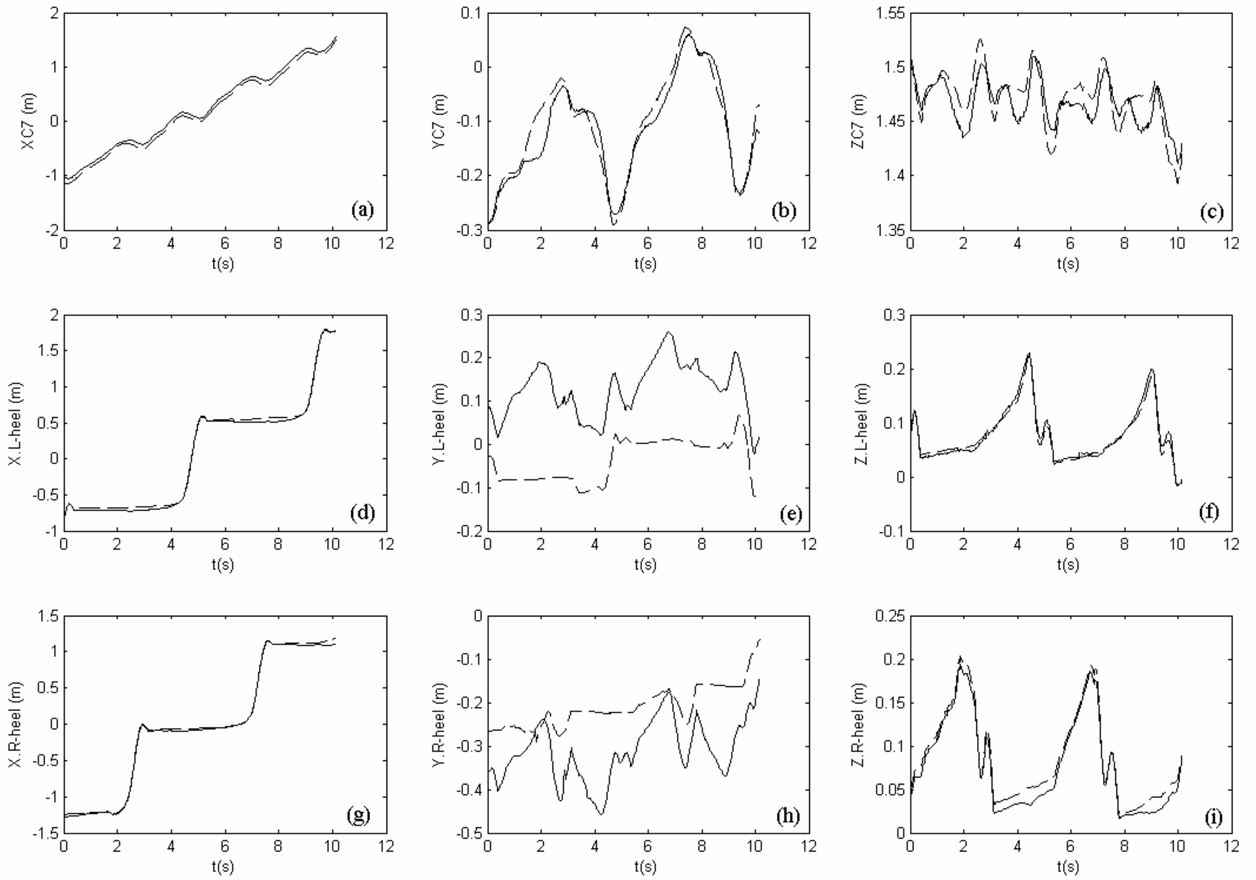

Figure 4. Comparison of the 3D trajectories of the end effectors predicted by the model (solid lines) and obtained from the experiment (dash lines) considering the lower trunk segment as the reference: C7 (top), L-heel (middle), R-heel (bottom)

The results of the forward kinematics analysis of the model, in comparison with the corresponding results of the experiment, are illustrated in figure 4, considering the pelvis segment as the reference. In general, there was a good agreement between the predictions of the model for the 3D trajectories of the end effectors examined and the data obtained from the experiment. For C7, all the X, Y, Z coordinates, and for L-heel and R-heel, the X and Z coordinates of the 3D trajectories were nearly identical in the modeling and experimental results. The largest differences were found for the Y coordinates of the heels, which represent the lateral movements of the legs.

Also stick figures of the robotic model in a representative swing phase in the sagittal, frontal and the horizontal plan are shown in the Fig 5. The lateral hip displacement in the frontal plane for ground clearance of the swing leg was shown in the fig 5-b. The moments of the actuators in the robotic model during a representative swing phase are shown in the Fig 6. Just one comparable torque in the hip joint of the normal walking (dash line) (20), indicates that the boundary values but not the shape of the curves in the normal and the modeled paraplegic walking are approximately the same (Fig 6-d). Comparison between the torques magnitude showed that the biggest magnitude belong to the whole body forward inclination over the stance leg, τ_3, (Fig 6-c) and the smallest one belong to the whole body axial rotation around the stance leg, τ_1, (Fig 6-a).

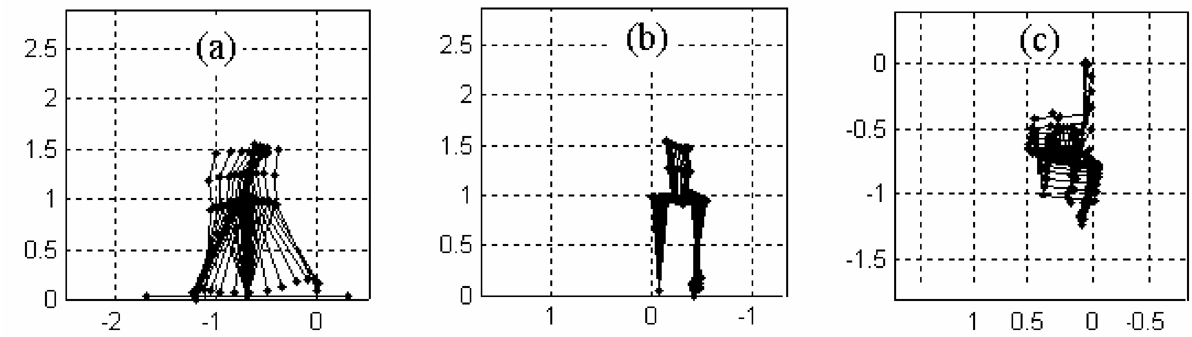

Figure 5. Stick figures of the 4 rigid body segments (trunk, pelvis, two legs) based on the 3D coordinate measurement during a representative swing phase: (a) Sagittal plan, (b) Frontal plan, (c) Horizontal plan

RSM analysis

Considering the results of the kinematical analysis of the paraplegic gait, five critical instants of the swing phase were found to appear at about 5%, 25%, 50%, 75% and 95% of the total swing phase duration. The ranges of variation of the independent variables, i.e., θ_2, θ_3, θ_4, θ_5, considered for the RSM analysis obtained assuming a wider domain (to at least 10 degrees) than that of the real values appeared in the kinematical analysis according to the experimental data for each critical instant of the swing phase. The results of the RSM analysis, indicating the relationship between the independent variables and the total muscular effort (TME) index, as the response functions, are illustrated in the following equations (5-9).

$$(TME)_{\%5} = 9.2863 + 2.0339\theta_2 + 5.2602\theta_3 + 0.3112\theta_4 + 0.2708\theta_5 \qquad (5)$$

$$(TME)_{\%25} = 5.5667 + 1.5652\theta_2 + 3.5019\theta_3 + 0.2120\theta_4 + 0.1759\theta_5 \qquad (6)$$

$$(TME)_{\%50} = 3.2071 + 1.2544\theta_2 + 1.4477\theta_3 + 0.1423\theta_4 + 0.1728\theta_5 \qquad (7)$$

$$(TME)_{\%75} = 6.7791 + 1.2982\theta_2 + 3.4017\theta_3 + 0.2457\theta_4 + 0.1588\theta_5 \qquad (8)$$

$$(TME)_{\%95} = 15.822 + 2.181\theta_2 + 7.352\theta_3 + 0.305\theta_4 + 0.295\theta_5 \qquad (9)$$

The constant coefficients of the linear regression equations (Eq. 4) of the TME index reflect the impact of each of the independent variables on these functions at each of the five critical instants of the swing phase. These impacts were elucidated in the Fig 7. Results of the Fig 7 indicate that the TME index is most sensitive to θ_3 and then θ_2 independent variables, i.e., the whole body forward inclination and lateral tilting, respectively. However, while the effect of θ_2 is almost consistent during the entire swing phase, θ_3 affects the TME differently in different instants. In particular, the θ_3 has its lowest impact on the instant 50% and the more greeter distance from this instant the more greeter effect to the TME. The trunk kinematical variables, i.e., θ_4, and θ_5, were found to affect the TME index with much lower impacts. This suggests that the total muscular effort (TME) during paraplegic gait will be decrease with the lower variation of the independent variable θ_3 around the instant 50% symmetrically. In the paraplegic gait due to the knee restriction to the full extension status, reduction on the θ_3 will be reflected to the shorter step length.

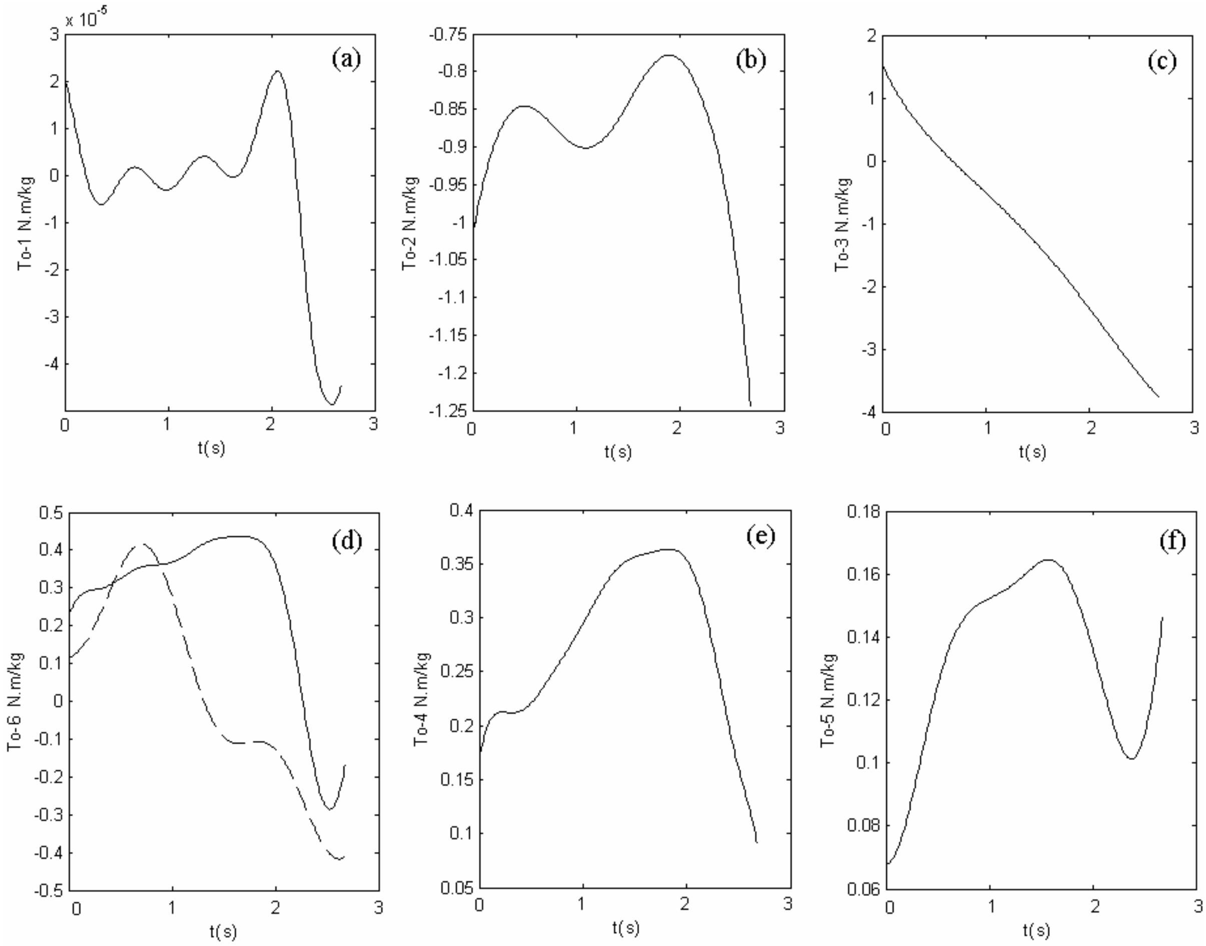

Figure 6. The torques related modeled paraplegic gait during a representative swing phase (solid lines) compared with the normal walking just in the hip joint (dash line). (a) whole body axial rotation around the stance leg, τ_1 , (b) whole body lateral tilting over the stance leg, τ_2 , (c) whole body forward inclination over the stance leg, τ_3 , (d) hip flexion of the swing leg, τ_6 , (e) trunk anterior bending, τ_4 , (f) trunk lateral bending, τ_5

Discussion

The efficacy and performance of the mechanical walking orthoses and the associated gait patterns has been often evaluated using experimental tests (23-26). Such tests provide assessments of the spatial and temporal characteristics of the gait, e.g., speed, stride length and cadence, using motion capture systems. They also provide estimations of the energy expenditure via measurement of the oxygen uptake and heart rate. However, pure experimental examinations are often unable to provide a deep understanding of the factors affecting the system's performance and cannot help to obtain optimal designs for orthosis and/or gait pattern. This is mainly due to their inadequate research methodology in view of the large variance in populations who use a particular orthosis, the significant impact of heterogeneity within each population, and the resultant small number of potential subjects available to participate in research that would meet strict inclusion and exclusion criteria required to minimize this heterogeneity (27). Mathematical modeling of the gait pattern has long been used as a research tool to provide a better understanding of the mechanics of human locomotion. It provides a detailed description of the role and significance of each individual relevant factor affecting the kinematics, dynamics, and energy expenditure of the gait. This is obviously of great importance for the interpretation of the experimental data, which normally reflects the results of a combination of several determinants, e.g., subject's conditions, orthotic design, and gait pattern. However, the most attractive feature of Mathematical modeling in gait biomechanics may be its capabilities for parametric studies, as well as synthesis of new motion patterns, which are not possible to be achieved experimentally (28).

The previous modeling studies of the paraplegic gait have been often concerned with the efficacy and performance of the mechanical walking orthoses in combination with the FES systems (13). However, the use of such hybrid devices for ambulation of paraplegics has not yet evolved practically, due to several limitations, e.g., long term infection of implanted electrodes, fast muscles fatigue of surface electrodes, long don and doff times, high set up and maintenance costs, etc (27). Similarly, the hybrid systems based on the external electric or pneumatic actuators are not yet in practical use due to the need to external energy sources, long don and doff times, and high maintenance costs (29). In the present study, we focused on ARGO mechanical orthoses which are routinely used by paraplegic individuals for locomotion (30). We studied the kinematics and dynamic behavior of the ARGO assisted paraplegic gait in detail in search for an improved gait pattern which provide lower energy expenditure. For the first time, the RSM was employed to explore the relationship between the detailed kinematical parameters, considered as independent variables, and the total muscular efforts, defined as response functions. Then by identification of the impacts of each kinematical parameter, practical guidelines were revealed to obtain a more advantageous gait pattern. The kinematical results of this study for ARGO assisted paraplegic gait are in general agreement with the data reported in the literature, in spite of the significant effect attributed to the heterogeneity of the test subjects. Similar to our study, other researchers (31-33) have found lateral tilting over the stance leg, at the beginning of the swing phase, and extension of the trunk and pelvis, in the mid swing. However, considering the results of our RSM analysis, the whole body forward inclination θ_3 has a significant effect on the energy expenditure of the paraplegic gait, spatially in the more distant points from the 50% instant point of the representative swing phase (Fig 7). Due to the direct relation between the step length and the whole body forward inclination (θ_3), in the paraplegic gait with the knees blocked, it was concluded that the shorter step length will cause the lower energy expenditure in the swing phase but in the complete cycle of walking to compensate the reduction of the walking speed due to the reduction of step length the cadence must be increased. This conclusion is in good agreement with our comparison of spatial and temporal parameters between the paraplegic gait and the normal one, as the result of the table 2 the stride length measured in the present study is slightly shorter than the normal walking but Cadence is almost 25% of that of normal gait and walking speed in the paraplegic gait is 20% of the speed of normal one. Thus it appeared that a low cadence contributed more than a reduced stride length to the slower speed of those walking using an ARGO compared to non-impaired gait.

In spite of the interesting observations discussed, care must be taken in making general conclusions from the results, in view of the several limitations involved in the methodology of the present study. The experimental data basis was obtained from a single individual with a specific level of injury and wearing a particular type of orthosis. The model was also simple and included a limited number of rigid segments and rotational DOFs. However, in spite of these limitations, the essential kinematic and dynamic characteristics of the ARGO-assisted paraplegic gait are thought to could be reproduced by our model. The basic structure of the model, with four segments for representation of the trunk, pelvis, and two legs, is consistent with our test configuration and might be also applied to a wide range of SCI individuals wearing other types of

KAFOs and HKAFOs. The 6 DOFs assumed for the joints of the model correspond to the basic motion restrictions imposed by the ARGO, i.e., sagittal plane reciprocal motion of the hip joints. However, the results of the forward kinematics analysis (Fig 4) suggest that the lateral rigidity of the hip joints of the orthosis might be inadequate to completely restrict the subject's joints from frontal plane motions, i.e., adduction and abduction, in practice (32). Unlike the close agreement we found between the modeling and experimental trajectories of the two heels in the sagittal plane, the relatively large lateral shifts observed in the experimental results were not consistent with the model's predictions (Fig 4). This suggests that for a more realistic simulation of the ARGO-assisted paraplegic gait, it might be necessary to include a lateral DOF with appropriate stiffness at the hip joints of the model. However, such stiffness data is not yet available in the literature and needs to be generated from experimental examinations. Obviously, if other types of HKAFOs are to be modeled, more changes in the model's rotational DOFs would be required. Another major limitation of the present study arises from the fact that the equations of motion were solved assuming static conditions. This simplification was necessary for our RSM analysis in order to keep the number of the design variables within an affordable range. As a result, however, our model ignores the dynamic effects of the joints' motions. In particular, the momentum produced by the body axial rotation around the stance leg is not reflected in our model. This effect, however, is not expected to be significant in ARGO-assisted paraplegic gait, in which the flexion-extension motion of the hip joints is controlled by the ARGO's cable link. Finally, we used the results of the RSM analysis to provide information on the muscular effort costs resulted from different maneuvers involved in the paraplegic gait. The muscular effort was characterized in terms of the sum of squared joints torques, in view of the fact that the individual muscles were not included in our skeletal model and their resultant effect was represented by joint torque generators. Such a simplification is only a rough estimation of the muscular effort and cannot reflect the limited capacity of the muscles for force generation. Moreover, it is well documented in the biomechanical literature that the force produced by a muscle depends upon the muscle length and velocity, according to Hill's modified model of muscle contraction (19). The similarity of our RSM results for TME index is thought to be resulted from not considering such details in the estimation of the muscular effort costs. Further work is going on to develop a more sophisticated musculoskeletal model of the paraplegic gait, so that a more accurate assessment of the muscular effort and energy cost could be obtained.

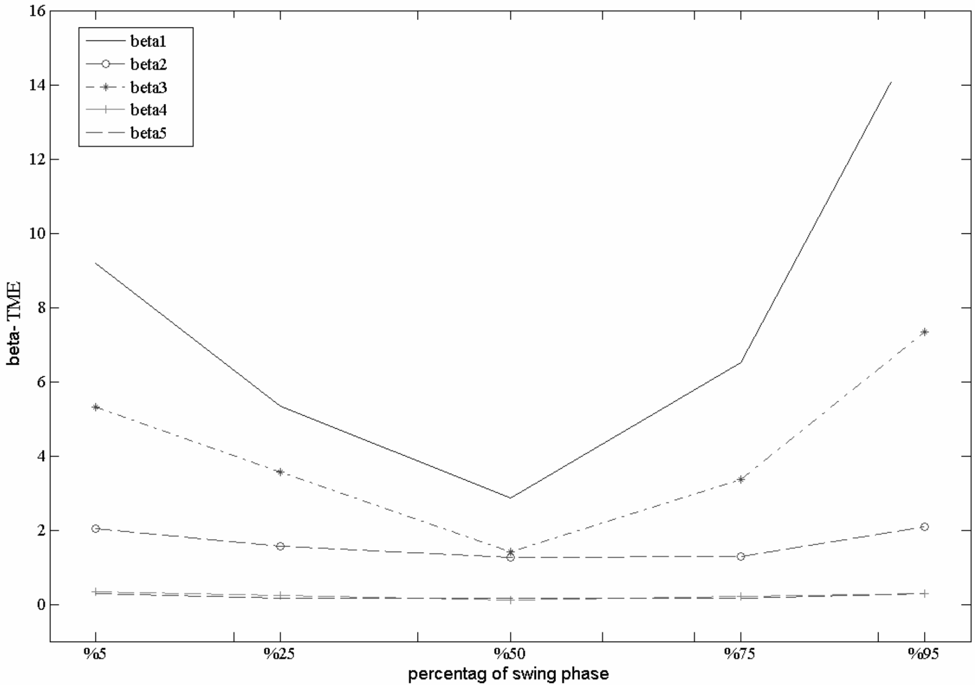

Figure 7. The constant coefficients of the linear regression equations (Eq. 4) of the TME index on these functions at each of the five critical instants of the swing phase

Conclusions

According to the RSM analysis, the whole body forward inclination θ_3 has been recognized as the most effective kinematics parameter on the total muscular effort as an index of energy expenditure during the paraplegic gait, spatially in the more distant points from the 50% instant point of the representative swing phase. Due to the direct relation between the step length and the whole body forward inclination (θ_3), in the paraplegic gait with the knees blocked, it was concluded that the shorter step length will cause the lower energy expenditure in the swing phase but in the complete cycle of walking to compensate the reduction of the walking speed due to the reduction of step length the cadence must be increased. So with a reduction on the step length and cadence increasing in a modified gait pattern, the paraplegic individuals are expected to achieve improved energy expenditure. However, further work, based on more sophisticated musculoskeletal models and large sample size experimental tests are needed to verify these conclusions.

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