Analysis of Trunk Movement in Orthotic Gait of Paraplegics

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Abstract—In an orthotic gait of paraplegics, a leg restriction and motor paralysis result in a significant loading. In this study, we address to quantify a relationship between the loading, leg restriction and motor paralysis, and analyze lumbar joint trajectories in the orthotic gait of paraplegic subjects and the ordinary and orthotic gaits of a normal subject using an inverted pendulum model. For the leg restriction, the trajectories are located anterior to an equilibrium point of the inverted pendulum, and the loading is higher due to the influence of gravity moment. Comparing the orthotic gait kinematics of paraplegic with that of normal in the horizontal plane, the lumbar joint trajectory in the paraplegic subjects was rectilinear shape, while that in the normal subject was curved in the direction to the equilibrium point. The loading is lower in the curved trajectory than in the straight trajectory because of the trade-off between gravity and inertia. These results suggest that the leg restriction and motor paralysis lead to the increase of the distance between the trunk movement and the equilibrium point of an inverted pendulum, which causes significant loading in the orthotic gait of paraplegics.

I. INTRODUCTION

To prevent secondary complications, paraplegics are trained to walk and stand wearing an orthosis [1]. The orthoses provide a functional standing and short-distance ambulation by restricting the movements of the paralyzed joints. Hip-knee-ankle-foot orthoses (HKAFO) are some of the most common types [2], [3]. In HKAFO, paralyzed trunk joints are supported by a pelvic band, and it has been used for high thoracic-injured patients. However, it is difficult for the patient to put the orthosis on by themselves, and they require a significant amount of assistance [1]. Knee-ankle-foot orthoses with a medial-single-hip joint (MSH-KAFO) are orthoses without pelvic bands [1], [4]. MSH-KAFO has been used for lower thoracic-injured patients because the paralyzed thoracic, lumbar, and hip joints are not restricted. We investigate the orthotic gait with MSH-KAFO on account of a practical advantage where they can put the orthoses on by themselves while seated in a wheelchair.

Paraplegic patients can walk only short distance due to significant loading. Several studies have approached to the problems of the loading [5], [6]. However the relationships between the orthotic gait pattern and the loading were not quantified. The loading results from leg restriction with orthoses and motor paralysis. It is important for an appropriate design of paralyzed leg joint, such as the restriction or drive, to evaluate the influences of leg restriction and motor paralysis. In this study, we address to quantify the relationship between the loading, leg restriction and motor paralysis in orthotic gait of paraplegics. First, we compare the gait kinematics of the orthotic gait of paraplegic subjects with that of the ordinary and orthotic gaits of a normal subject.

An inverted pendulum is one of the simplest models to understand human gait, and various study using the model has suggested that the movement of center of gravity is efficient during single support phase in the ordinary gait of normal [7]. The analysis of lumbar joint trajectories using an inverted pendulum model is an effective approach to specify the cause of the significant loading in the orthotic gait of paraplegics. Thus we quantitatively evaluate the relationship between the significant loading, leg restriction and motor paralysis from the analysis of lumbar joint trajectories using an inverted pendulum model.

II. GAIT MEASUREMENT EXPERIMENTS

A. Method

1) Subjects: Two paraplegic subjects 1 and 2 and normal subject 3 participated in the experiments. The injury levels of the subjects 1 and 2 were T6 and T10, while subject 3 had no disability of motor function. The subjects gave informed consent, and they agreed to participate in writing.

2) Experiments: The paraplegic subjects walked on a level floor with MSH-KAFO and crutches. Their foot joints were restricted in dorsiflexion position. The knee joints were fixed in maximum extension position. The degrees of freedom in the hip joint except the extension/flexion were fixed. The marker positions shown in Fig. 1 were measured using a 3-dimensional position measurement device OPTO-TRAK3020 (Northern Digital Inc.) at 100 Hz. The paraplegic subjects were asked to put on the MSH-KAFO and walk along the line marked on the floor as well as the usual training activities. The normal subject 3 was instructed to walk under the following conditions.

(1) Ordinary gait with the stride length of subject 1
(2) Orthotic gait with the stride length of subject 1
(3) Ordinary gait with the stride length of subject 2
(4) Orthotic gait with the stride length of subject 2

In the orthotic gait conditions, the normal subject 3 put on MSH-KAFO as well as the paraplegics use. In the ordinary gait conditions, pressure sensors were attached on their toes and heels to detect gait phases (stance and swing phases).

3) Analysis: The measured position data in the orthotic gait of the paraplegic and normal subjects was filtered using a low-pass filter with cutoff frequency of 5 Hz, and the gait
phases were detected from the velocity of the toe position. The position data in the ordinary gait of normal subject 3 was filtered with cutoff frequency of 10 Hz because of the short gait cycle. Their gait phases were detected from foot pressure. One-cycle position data was separated based on the timing of left toe-off, and 15 gait patterns were picked up at random and analyzed. The gait parameters of cycle time, stride and velocity for the conditions were calculated from the one-cycle gait patterns. We address the lumbar joint trajectories to examine the differences of trunk movements among the conditions. A mean trajectory was calculated by applying 5-th order Fourier series approximation to averages of the 15 trajectories.

B. Results

1) Gait parameters: Table I shows the gait cycle, stride and gait velocity in the experimental condition. In the orthotic gait, the gait cycle is larger and the velocity is lower than in the ordinary gait of normal.

2) Lumbar joint trajectories: Although we show the results only in the large stride conditions, the result in the small stride conditions corresponded to the following results. The mean trajectories of the lumbar joint in the frontal plane are shown in the left column of Fig 2. The maximum peaks of the vertical displacement are found during single support phases in all the conditions. The peaks of lateral displacement in the orthotic gait of paraplegics are found during double support phases. While those in the orthotic gait of normal are found at heel-on, and those in the ordinary gait of normal are found during single support phases. The vertical and lateral displacements were the smallest in the ordinary gait of normal, while those were the largest in the orthotic gait of paraplegics.

The mean trajectories in the horizontal plane are shown in the right column of Fig 2. Relating to the phase in which the lateral peak appear, the direction of the lateral displacement at heel-on was support leg side in the orthotic gait of paraplegics. There was little displacement in the orthotic gait of normal, and the direction was swing leg side in the ordinary gait of normal. In the single support phases, the trajectories in the paraplegics were rectilinear shape, while those in normal were curved. These differences might relate to the significant loading in the orthotic gait of paraplegics. Particularly, force acting on crutches shows a maximum peak in the single support phase [5]. In the next section, the relationship between the loading and the kinematical differences for the conditions are analyzed using an inverted pendulum model.

III. ANALYSIS USING INVERTED PENDULUM MODEL

A. Method

1) Analysis of lumbar joint trajectories in single support phase: As shown in Fig. 3 (a) and (b), the lumbar joint trajectories are mapped to the coordinates of an inverted pendulum. First, measured lumbar joint trajectories were shifted so that the origin is equal to the foot joint position. When the foot joint position is defined as the center of rotation, an

![Image](OPTOTRAK.png)

Fig. 1. Experimental environment.

**Table I**

Gait parameters in the experimental conditions.

<table>
<thead>
<tr>
<th>Sub.</th>
<th>Gait cycle [s]</th>
<th>Stride [cm]</th>
<th>Velocity [m/s]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sub. 1</td>
<td>3.62 (0.13)</td>
<td>73.6 (9.0)</td>
<td>75.9 (5.6)</td>
</tr>
<tr>
<td>Sub. 3(1)</td>
<td>1.32 (0.04)</td>
<td>76.5 (4.5)</td>
<td>77.9 (4.5)</td>
</tr>
<tr>
<td>Sub. 3(2)</td>
<td>2.80 (0.19)</td>
<td>77.4 (5.0)</td>
<td>76.3 (6.6)</td>
</tr>
<tr>
<td>Sub. 3(3)</td>
<td>2.29 (0.15)</td>
<td>49.9 (7.0)</td>
<td>48.8 (5.0)</td>
</tr>
<tr>
<td>Sub. 3(4)</td>
<td>1.25 (0.06)</td>
<td>45.8 (1.5)</td>
<td>45.8 (2.1)</td>
</tr>
<tr>
<td>Sub. 3(5)</td>
<td>2.72 (0.17)</td>
<td>46.1 (3.8)</td>
<td>45.4 (3.5)</td>
</tr>
</tbody>
</table>

![Orthotic gait in paraplegic subject 1](Orthotic_gait.png)

![Ordinary gait in normal subject 3](Ordinary_gait.png)

Fig. 2. Lumbar joint trajectories of the large stride conditions in the frontal plane (left column) and in the horizontal plane (right column). The solid lines denote the trajectories in single support phase, and dashed lines denote those in double support phase. Circles and squares indicate the positions at toe-off and heel-on, respectively.
approximation error appears between the trajectories and the spherical surface. Next we estimate an appropriate position of the center of rotation with respect to the progression position \( y_0 \) by minimizing mean squared error. An equation of spherical surface is given by

\[
x(i)^2 + (y(i) - y_0)^2 + z(i)^2 = L^2.
\]

(1)

\( x(i), y(i), z(i) \) is the lateral, progressional and vertical position of the lumbar joint, and \( i \) indicates a time series number, and \( L \) is a radius of sphere. \( y_0 \) is given by differentiating (1) with respect to time.

\[
y_0 = \frac{\sum_{i=1}^{N} \dot{y}(i) \{ x(i) \ddot{x}(i) + y(i) \ddot{y}(i) + z(i) \ddot{z}(i) \}}{\sum_{i=1}^{N} \ddot{y}(i)}.
\]

(2)

Finally we obtain the mean trajectory as well as Section II-A.3 from the origin-shifted trajectories. As shown in Fig. 3 (b), the degrees of freedom of an inverted pendulum are defined as rotation angle \( \theta \) and \( \psi \) related to \( Y \) and \( Z' \) axis, respectively. The generalized coordinates is defined as \( q = [\theta, \psi]^T \). Then, the equation of motion is given by

\[
\tau = M(q) \ddot{q} + V(q, \dot{q}) + G(q),
\]

(3)

where \( \tau = [\tau_\theta, \tau_\psi]^T \) is torque acting on the generalized coordinates, \( M \) is inertial matrix, \( V \) and \( G \) are Coriolis’s - centrifugal term and gravity term. The dynamics parameters of the mass and link length are set to \( M = 68 \text{ kg}, \ L = 1.0 \text{ m} \). A position \( q \), velocity \( \dot{q} \) and acceleration \( \ddot{q} \) are calculated from the mean trajectory using inverse kinematical equations, and torque \( \tau \) is calculated from (3). Because the magnitude of the torque relates to the loading of gait, we introduce the following function as a criterion of the loading,

\[
C = \int_0^{T_f} (|\tau_\theta| + |\tau_\psi|) dt,
\]

(4)

where, \( T_f \) indicates the movement duration of the single support phase.

2) Shapes of lumbar joint trajectory: To examine the differences between the orthotic gait of paraplegics and normal, we analyze the influences of a trajectory direction angle \( \delta \) at heel-on shown in Fig. 3 (a), which represents a shape of the trajectory. To calculate the trajectory from \( \delta \), we apply the minimum jerk model known as a good approximation of smooth human movements [8]. In this model, the trajectory can be obtained from the parameters consisting with positions, velocities and accelerations at start (toe-off) and end (heel-on) points. To generate the trajectory corresponding to \( \delta \), we calculate the velocity at end point from \( \delta \) and tangential velocity at end point, and other parameters are estimated by the mean trajectory.

First, we calculate a vector specifying the velocity direction in state space \( u = [u_\theta, u_\psi]^T \) from \( \delta \). The relationship between the velocity direction in the horizontal plane and that in the state space is represented by

\[
\begin{bmatrix}
-\sin \delta \\
\cos \delta
\end{bmatrix} =
\begin{bmatrix}
-Ls_\theta c_\psi & -Lc_\theta s_\psi \\
0 & -Lc_\psi
\end{bmatrix}
\begin{bmatrix}
u_\theta \\
u_\psi
\end{bmatrix},
\]

(5)

where \( s_\theta, \psi \) and \( c_\theta, \psi \) indicate the sine and cosine functions of \( \theta \) and \( \psi \). \( u \) is obtained by solving (5). Next, the state velocity \( \dot{q}_i \) is obtained from

\[
\dot{q}_i = \frac{|v_f|}{|v_i|} u,
\]

(6)

where \( |v_f| \) is a tangential velocity at heel-on in mean trajectory, and \( |v_i| \) is a tangential velocity obtained from velocity direction \( u \). A minimum jerk trajectory is calculated from the boundary conditions, and the relationship between the loading and the shape of trajectory is examined on the basis of (4).

B. Results

1) Analysis of mean trajectory in single support phase: Fig. 4 (a) and (b) show the lumbar joint trajectories in the left single support phase. Initial positions (right toe-off) in the ordinary gait are located behind the equilibrium point and the peaks of the lateral displacement are located near the equilibrium point. In the orthotic gaits of paraplegics and normal, the initial positions are located anterior to the equilibrium point. Integrated absolute torque is shown in Fig 4 (c). The results show that the loading is the lowest in the ordinary gait of normal and it is the highest in the orthotic gait of paraplegics. In addition, not only the lateral torque but also the progressional torque show high values with respect to large lateral and vertical displacements in the orthotic gait. Resisting the gravity moment acting in the anterior direction, the torque acts on the posterior direction.

2) Analysis of direction angle \( \delta \): Fig 5 (a) shows the trajectories depending on the value of \( \delta \) and the mean trajectory. The minimum jerk trajectories calculated with direction angle of mean trajectory \( \delta_{MT} \) consist with the mean trajectories. The trajectories of paraplegic and normal are curved according to the value of \( \delta \). Fig. 5 (b) shows the integrated absolute torque corresponding to the trajectories shown in Fig. 5 (a). The results of the mean trajectories and the minimum jerk trajectories are well consisting. \( \delta \) of the trajectory, in which torque is lowest, corresponded to the \( \delta \) of the mean trajectory in the orthotic gait of normal. The torque is lowest when \( \delta \) is equal to 0 deg in both paraplegics and normal trajectories. Then the trajectory curved in the direction to the equilibrium point.
**IV. DISCUSSION**

The lumbar joint trajectories of the ordinary gait of normal are located near the equilibrium point of the inverted pendulum. They are explained as the ballistic movement with sufficiently large velocity at toe-off. Due to the leg restriction, the trajectories are located anterior to the equilibrium point during the single support phase so that the center of mass is inside the base of support consisting with the contact points of a stance foot and crutches. However their loading is higher due to the influence of gravity moment. The anteriorly located trajectory in paraplegic gait results from insufficient velocity at toe-off because of the foot-joint restriction. To improve the orthotic system, a mechanism pushing on the ground during the double support phase may be required for the sufficient velocity at toe-off.

Comparing the lumbar joint trajectories of paraplegics with that of normal, the trajectories in the paraplegics were rectilinear shape, while those in normal was curved in the direction to the equilibrium point. The loading is lower in the curved trajectory than in the straight trajectory, where the gravity moment increases according to the distance between the trajectory and the equilibrium point, while the moment of inertia increases due to distance of trajectory. These results suggest that the leg restriction and motor paralysis lead to the increase of the distance between the trunk movement and the equilibrium point of an inverted pendulum, which causes significant loading in the orthotic gate of paraplegics.

**REFERENCES**


