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Gait pattern generation for a power-assist device of paraplegic gait

Takahiro Kagawa and Yoji Uno

Abstract— We address a gait pattern generation on a legged locomotor device (WPAL: Wearable Power-Assist Locomotor) for paraplegics. In the gait movement with WPAL, a backward falling is a considerable problem, and a foot-floor collision during a swing movement would induce a loss of balance. In addition, adjustability of the gait parameters, such as stride length, gait cycle and maximum hight of the toe clearance, would be required for an individual paraplegic according to the degrees of his disabilities and skills. In this paper, we propose a gait pattern generation method considering the requirements of the stability and the adjustability. First, the trajectories of toe position, horizontal hip position, and foot plantar angle are calculated using a minimum jerk trajectory with the constraints of the position and velocity at via points. Second, the desired trajectories of joint angles are determined from the calculated trajectories by inverse kinematic equations. We demonstrate that generated desired trajectories for various gait parameter values and boundary conditions were satisfied with the required stability conditions.

(a) Robotic part of WPAL (b) Gait movement with WPAL and walker

Fig. 1. A prototype system of WPAL (Wearable Power-Assist Locomotor). (a) shows the robotic equipment with 6 actuated joints consisting hip, knee and ankle joints. (b) shows walking movement with a walker. The walker is placed in front of hip body.

I. INTRODUCTION

Paraplegics with spinal cord injuries cannot keep a standing posture and walking, and they use a wheelchair for mobility. Life in a wheelchair often results in secondary complications [1]. Reconstruction of walking is desirable to improve not only locomotor functions but also physiological problems [2]. Several researches have attempted to develop gait reconstruction systems for paraplegics [3][4], no practical systems have been reported.

We are developing a legged locomotor device for paraplegic patient (WPAL: Wearable Power-Assist Locomotor, ASKA Corp.) [5][6]. Fig. 1 (a) shows a robotic system with 6 actuators assisting extension and flexion movements of hip and knee joints and plantarflexion and dorsiflexion movements of ankle joints. WPAL supports walking, sitting-to-standing and standing-to-sitting movements.

Fig. 1 (b) shows walking movement with WPAL assisting leg movements. To maintain the upright posture during walking, arm supports are required using a walker [7]. It is, however, difficult to avoid backward falling by the arm supports because the walker is placed in front of the user. In addition, a foot-floor collision during a swing movement is one of the risk factors of falling because it results in emergency stop of over current protection due to large feedback torque.

The design of an appropriate gait pattern is important to reduce the risks of falling and to maintain smooth walking. In addition, it is desirable to adjust the gait parameters, such as the stride length, gait period, according to the level of user's impairments and motor skills. In this paper, we propose a method of gait pattern generation for a legged locomotor device, considering the risks of falling and adjustability of the gait parameters. First, we formulate a condition of forward progression, i.e. not-falling backward, based on an inverted pendulum model. Next, we describe a method of a gait pattern generation based on the requirements. Finally, we report the examination results of the proposed method.

II. ANALYSIS OF BACKWARD FALLING

The loss of ballance for backward falling is one of the most serious risk factors for paraplegic gait. Since the body center of mass (COM) locates behind of the ankle position of stance leg at toe-off, sufficiently high COM velocity would be required to forward progression. To determine a boundary condition between backward falling and forward progression, we analyzed a linearized inverted pendulum model [8] during swing movement.

As shown in Fig. 2, a COM is regarded as a mass point and an ankle joint of a stance leg is a pivot of the inverted pendulum. The COM position related to the ankle joint position is represented by x^{com}. The equation of motion linearized around an equilibrium point ($x^{com} = 0$) is

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T. Kagawa and Y. Uno are with Department of Mechanical Science and Engineering, Nagoya University, 1 Furo-cho, Chikusa-ku, Nagoya, Japan. kagawa@nuem.nagoya-u.ac.jp

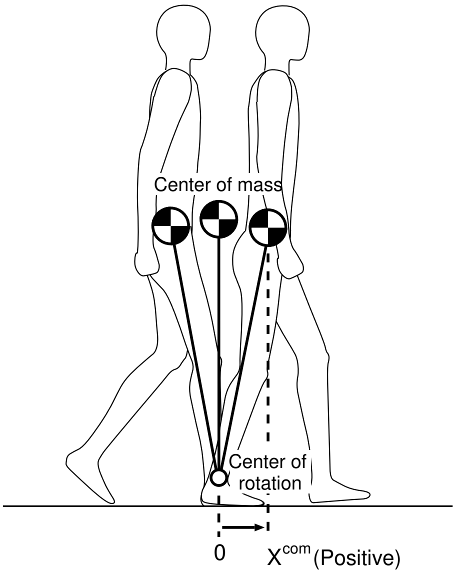


Fig. 2. An inveted pendulum model. A mass point of the inverted pendulum corresponds to a body center of mass, and a pivot corresponds to an ankle joint of a support leg.

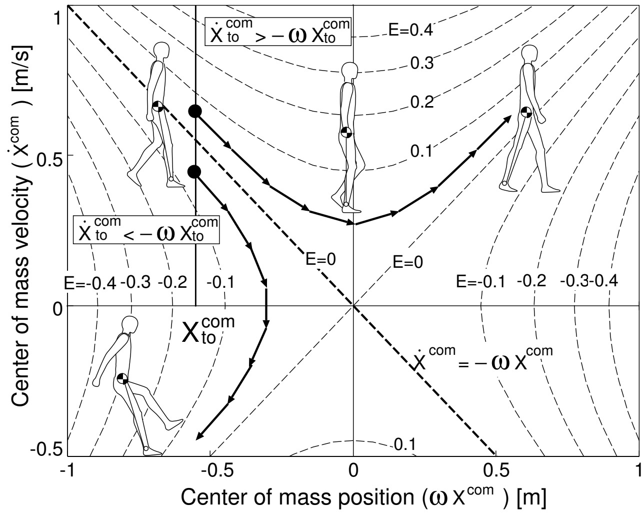


Fig. 3. Phase diagram between the position and velocity of the body center of mass, where the position represents the product of the center of mass position and natural frequency (ωx^{com}). The relationship between them is represented by a hyperbolic function described in (2). The gap of the hyperbolic curves is 0.1 J/kg. A thick line shows the critical velocity for the position in the necessary condition.

given as

$$\dot{x}^{com} = \omega^2 x^{com} \quad , \omega = \sqrt{\frac{g}{l}}, \tag{1}$$

where ω is the natural frequency, g is the gravity acceleration, and l is the length between the ankle joint and the COM position. When the ankle joint moment is zero, i.e. a ballistic movement is performed, a relationship between the COM position and velocity is given as the law of mechanical energy conservation.

$$\frac{1}{2}\left\{\dot{x}^{com}(t)\right\}^2 - \frac{1}{2}\omega^2\left\{x^{com}(t)\right\}^2 = E, \tag{2}$$

where E indicates constant mechanical energy per unit mass. The first term of (2) indicates the kinetic energy, and the second term is the potential energy. The relationship between the COM position and velocity during a ballistic movement is represented by a hyperbolic curve. We assume that the COM position is posterior to the ankle position at the time of toe off (t_{to}): $x^{com}(t_{to}) < 0$, and the velocity is progressional direction: $\dot{x}^{com}(t_{to}) > 0$. The mechanical energy must be positive value so that the COM position is going to the equilibrium point ($x^{com} = 0$). Therefore, a condition of forward progression without energy input is represented by

$$\dot{x}^{com}(t_{to}) > -\omega x^{com}(t_{to}). \tag{3}$$

Equation (3) indicates that the COM velocity must be larger as the the position locates more posteriorly for the ankle position.

Fig. 3 shows a phase portrait of COM position and velocity. The hyperbolic curves (dashed lines) indicate the contour lines of the mechanical energy (the interval is 0.1 J/kg), and they correspond to the ballistic COM trajectories. A dashed-thick line is the boundary of forward progression area represented by (3). The mechanical energy is large when the COM velocity is large and the COM position is near the equilibrium point. In addition, the energy is zero when a state is on the boundary line.

In order to consider the condition of forward progression, we focus on the area of the second quadrant ($x^{com} < 0$ and $\dot{x}^{com} > 0$) as a states at toe-off. In the area that the mechanical energy is lower than that on the boundary line, i.e. $E < 0$, the COM position cannot reach the equilibrium point, and turn to the backward movement. When the COM state locates over the boundary, the trajectory pass through the equilibrium point and enable to perform the forward progression without energy input. The condition indicates that the leg joints should be controlled to ensure the sufficient COM velocity before toe-off.

For applying this condition for gait pattern generation, the measurement of the COM is difficult because the movements of the trunk, head, and arms cannot be measured and controlled. We assume the hip joint position as the COM position because the COM position locate near the hip joint during standing and walking [9].

III. Gait Pattern Generation

A. Gait Phases

A user handles a walker to maintain an upright posture during walking with WPAL. An arm movement is required to displace the walker anteriorly after a leg swing movement. During the arm movement, movements of leg joints should be stopped because the base of support should be sufficiently large. In addition, the condition forward progression must be satisfied before toe off. Therefore we define three gait phases for one

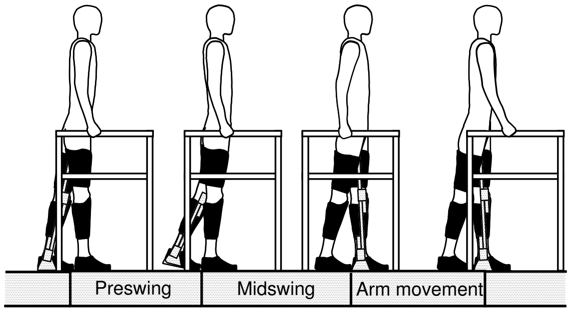
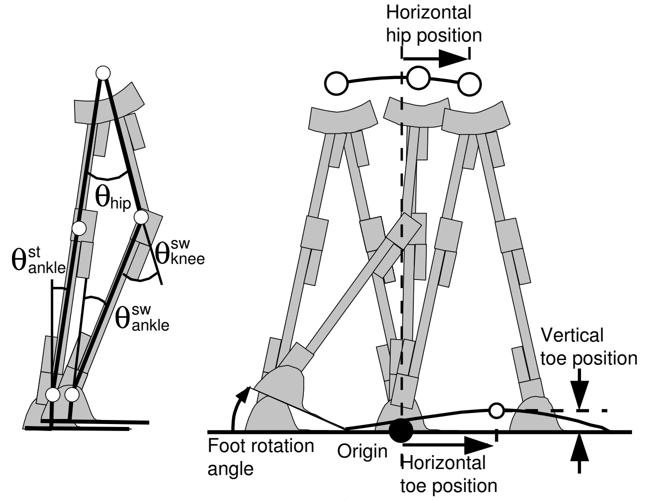


Fig. 4. Gait phases for gait pattern generation.



(a) Joint degrees of freedom (b) Variables of trajectory generation based on the toe clearance and not falling condition.

Fig. 5. (a) Four joint values are specified for gait pattern generation. (b) Variables of the external coordinates to be satisfied with the requirements.

step movement as ‘preswing’, ‘midswing’, and ‘arm movement’ shown in Fig. 4.

The preswing is defined as a movement phase from the heel off of the swing leg to the toe off. Since, the COM moves forward to satisfy the condition of forward progression, the ankle joint of the stance leg moves toward dorsiflexion position. In addition, the foot of swing leg is inclined and the knee and hip joints moves extension direction. The midswing phase corresponds to the duration from toe off to foot contact. In the midswing phase, the swing leg moves forward and the ankle of stance leg continues to the dorsiflexion movement so that the movement distance of the foot is equal to the given parameter value of the stride length. The arm movement phase corresponds to the duration from foot contact to heel off of the next swing movement. The arm movement with a walker is performed in this phase, and the leg joints are not moved.

B. Variables for gait pattern generation

Fig. 5 (a) shows the joint degrees of freedom for design of the gait pattern, where the knee joint of stance leg is fixed at maximum extension position because large joint torque is required for small joint movement. The variables for gait pattern generation are ankle angle of stance leg θ^{st}_{ankle}, hip angle θ_{hip}, knee angle of swing leg θ^{sw}_{knee} and ankle angle of swing leg θ^{sw}_{ankle}.

The condition of (3) should be satisfied to reduce the risk of backward falling. In addition, the trajectory of the toe position during the midswing phase is also important to avoid the foot floor collisions. For these requirements, it is difficult to determine the joint angles directly. Therefore, we determine the trajectories of the horizontal hip joint position and vertical toe position are calculated to satisfy the requirements. To adjust the stride length, the trajectory of horizontal toe position is determined. In addition, the time series of inclination angle of swing foot is calculated to determine the 4-dimensional joint angles. These variables are shown in Fig. 5 (b).

C. Trajectories Generation

Minimum jerk trajectory [10] is used to determine the positions and angles because of the simple calculations of a smooth trajectory. The minimum jerk trajectory $x(t)$ is given by the fifth order polynomial function.

$$x(t) = a_5t^5 + a_4t^4 + a_3t^3 + a_2t^2 + a_1t + a_0. \quad (4)$$

The position, velocity, and acceleration at the start point ($t = 0$) are x_0, $\dot{x}_0$, and $\ddot{x}_0$, and that at the end point ($t = t_f$) are x_f, $\dot{x}_f$, and $\ddot{x}_f$. When velocity and acceleration at the start and end points are stationary, a_1 and a_2 are zero, and other parameter values are determined by the following equation.

$$\begin{pmatrix} a_3 \\ a_4 \\ a_5 \end{pmatrix} = \begin{pmatrix} t_f^3 & t_f^4 & t_f^5 \\ 3t_f^2 & 4t_f^3 & 5t_f^4 \\ 6t_f & 12t_f^2 & 20t_f^3 \end{pmatrix}^{-1} \begin{pmatrix} x_f - x_0 \\ 0 \\ 0 \end{pmatrix}. \quad (5)$$

The various requirements, such as not-falling backward and maximum height of swing toe, are considered by constraints of the minimum jerk trajectory. As the constraints are represented as the position and velocity of a via-point at time t_{via}: (x_{via}, $\dot{x}_{via}$), the minimum jerk trajectory with the constraints is given as follows,

$$x(t) = \begin{cases} a_5t^5 + a_4t^4 + a_3t^3 + x_0 & (t \le t_{via}) \\ \\ a_5t^5 + a_4t^4 + a_3t^3 + x_0 & \\ +\frac{1}{24}\pi_1(t - t_{via})^4 & (t > t_{via}) \\ +\frac{1}{120}\pi_2(t - t_{via})^5 & \end{cases}. \quad (6)$$

The parameters of a_3, a_4, a_5, π_1 and π_2 are determined according to boundary conditions by the following equations,

$$\begin{pmatrix} a_3 \\ a_4 \\ a_5 \\ \pi_1 \\ \pi_2 \end{pmatrix} = A^{-1} \begin{pmatrix} x_{via} - x_0 \\ x_f - x_0 \\ \dot{x}_{via} \\ 0 \\ 0 \end{pmatrix}, \quad (7)$$

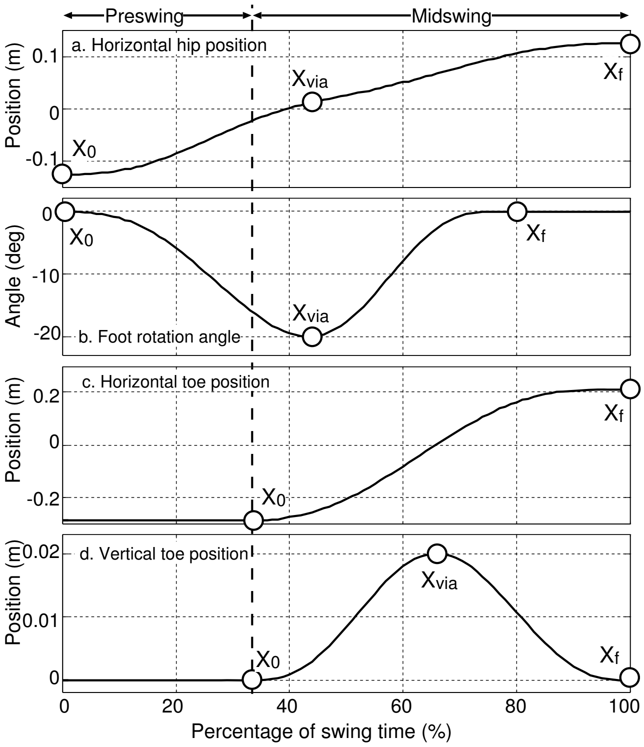


Fig. 6. Generated gait patterns for various parameter values and initial and terminal conditions.

$$A = \begin{pmatrix} t_{via}^3 & t_{via}^4 & t_{via}^5 & 0 & 0 \\ t_f^3 & t_f^4 & t_f^5 & \frac{(t_f - t_{via})^4}{24} & \frac{(t_f - t_{via})^5}{120} \\ 3t_{via}^2 & 4t_{via}^3 & 5t_{via}^4 & 0 & 0 \\ 3t_f^2 & 4t_f^3 & 5t_f^4 & \frac{(t_f - t_{via})^3}{6} & \frac{(t_f - t_{via})^4}{24} \\ 6t_f^2 & 12t_f^3 & 20t_f^4 & \frac{(t_f - t_{via})^2}{2} & \frac{(t_f - t_{via})^3}{6} \end{pmatrix}. \tag{8}$$

A typical trajectories of the hip and toe positions and the foot angle are shown in Fig. 6, where the swing time indicates the sum of the time of preswing and midswing phases. In this pattern, the time of preswing phase (toe-off) is 33 % of the swing time. The trajectories of horizontal toe position is determined by a minimum jerk trajectory without constraint, and other trajectories are determined by that with the constraints of via points. As the start postions X_0 of the trajectories are given, the end position is determined from X_0 and the stride length S which is 0.5 m in the typical trajectories. Parameters of via points: x_{via}, $\dot{x}_{via}$, and t_{via} are specified on the basis of the requirements and user-specific adjustments. The swing movement time is also user-specific parameter.

The hip position and the foot angle start at the beginning of the preswing phase, and the horizontal and vertical toe positions start at beginning of the midswing phase (toe-off). The time of the via point of hip position and foot angle is 43 % of the swing time. The end time of the foot rotation is 80 % of swing time because the maximum knee flexion could be reduced. The end position of the hip is given as $X_f = X_0 + S/2$ and that of foot angle is zero for foot-flat contact with the ground. The hip position and velocity at the via point are set by trial and error so that the state at toe-off is satisfied with the condition of forward progression for not only walking but also transfer movements between standing and walking. The via point position is 0.02 m, and velocity is equal to the mean velocity. The via point position of foot angle is determined by trial and error to reduce maximum knee flexion, and the velocity at the via point is zero.

The toe position is stationary during preswing phase and, moves during the midswing phase. The via-point position of the vertical trajectory is set to the parameter value of the maximum toe height, and the velocity is set to be zero. The time of the via point is set to be 66 % to reduce the maximum flexion angle of the knee.

The joint angles (θ_{ankle}^{st}, θ_{hip}, θ_{knee}^{sw}, θ_{ankle}^{sw}) are calculated from the trajectories using an inverse kinematics transformation, where the parameters of length of body segments are specified for individuals.

IV. EXPERIMENTS

Aiming to confirm the feasibility of our method, the gait patterns are calculated for various parameter values, initial and terminal conditions. We compared two patterns changed the stride length to 0.4 m and the maximum toe height to 0.04 m with the typical pattern shown in Fig 6. For changing the initial and terminal posture, transfer movements between standing and walking are calculated. We confirm the smoothness of the trajectory and whether the requirements are satisfied.

To validate the proposed method, examinations for paraplegics were performed at Fujita Health University. The examinations were a part of the clinical examinations of WPAL approved by the Institutional Review Board of Fujita Health University. Five paraplegic patients participated in this experiments. They gave consent for their participations, and walked with the WPAL and a walker under the supervision of physical therapists and medical doctors. During the walking, auditory cues were given at the timings of the phase transitions. The parameter values of the stride length, the maximum toe height and the duration of the phases were specified for individual subjects before the examinations. The gait movements were recorded by a video camera at 24 fps.

V. RESULTS

Fig. 7 shows the calculated gait patterns for changing the parameter values and the initial and terminal postures. The left figures are joint trajectories during swing movement and the right figures are stick pictures in which the time interval is 10 % of swing movement time and the thick lines show the initial and terminal posture. The joint angles of all patterns show smooth

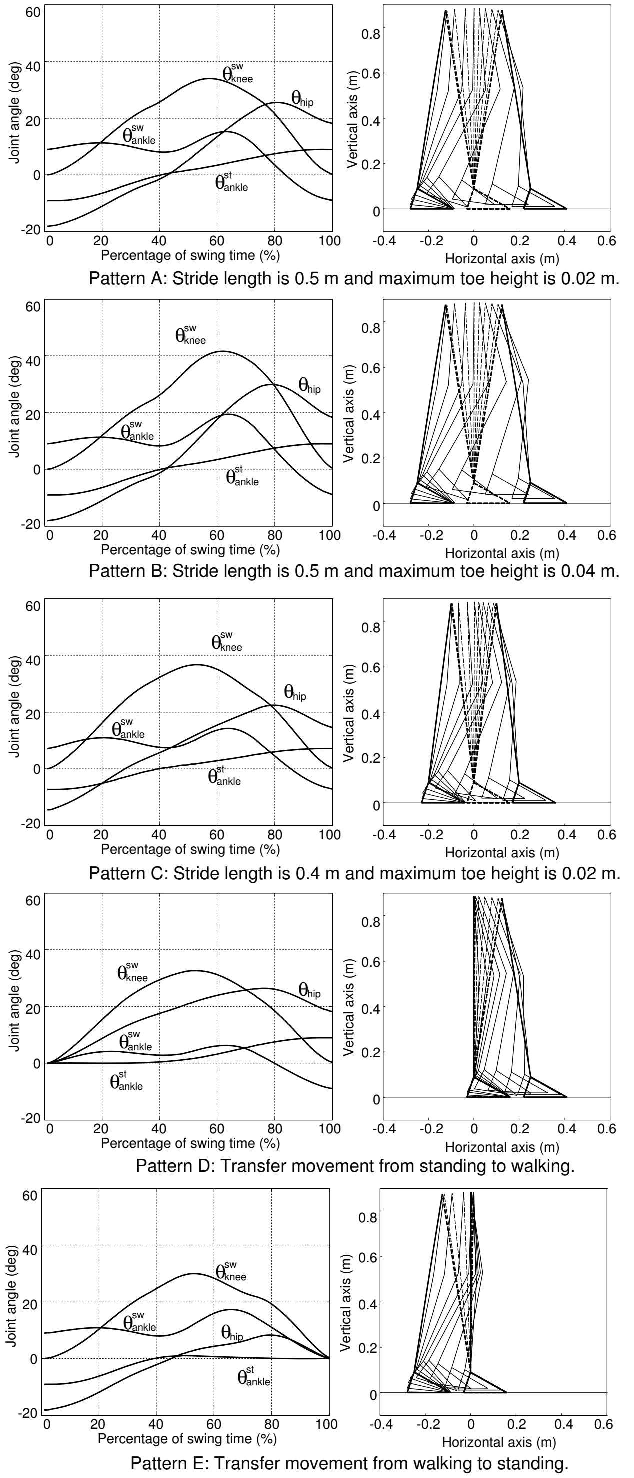


Fig. 7. Generated gait patterns for various parameter values and initial and terminal conditions. Pattern A, B and C are gait movements, and Pattern D and E are transfer movements between standing and walking.

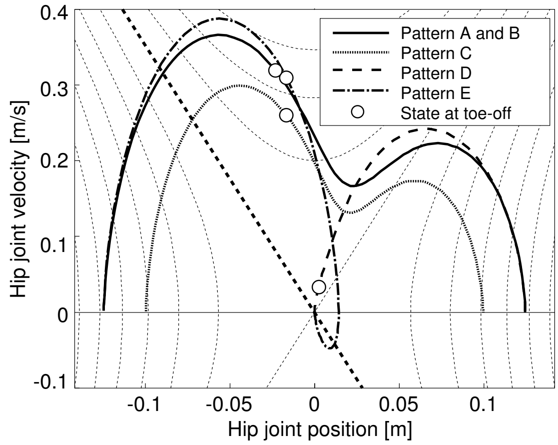


Fig. 8. Phase diagram of hip joint position and velocity when the swing time is 1.33 s. The energy gap of the contour lines is 0.02 J/kg The states (0, 0) is the that of a quiet standing. White circles indicate the states at toe-off.

time series patterns for changing the parameters and postures.

Pattern A is the typical trajectory shown in Fig. 6. In the preswing phase (from 0 % to 33 %), The ankle joint of the stance leg move toward dorsiflesion, and hip and knee joints of the hind leg are flexed for the forward proggresion of hip. In the midswing phase, the hip, swing knee and swing ankle joints show the peaks around the maximum toe height (66 %). At the end of the midswing phase, the knee joint angle of swing leg is maximum extension position and the toe and heel positions are contact on the ground.

Pattern B is a gait pattern in which the maximum height of the toe position is change to 0.04 m. The maximum flexion angle of swing knee and hip increase due to larger vertical displacement of the toe. Pattern C is that in which the stride length is change to 0.4 m. The amplitude values of the hip and ankle joints decrease because the initial and terminal postures are changed. Pattern D shows a transfer movement from standing to walking. The initial posture is set to standing posture and the stride length is set to a half of the stride length of gait, and other parameter values are the same of Pattern A. Pattern E shows the transfer movement from walking to standing. The parameter values are the same of Pattern A except to the terminal posture and stride length. The ankle joint is moved toward the neutral position in the preswing and is stationary in midswing phase so that the pattern is satisfied with the condition of forward progression.

The Fig. 8 shows the hip position and velocity on the phase portrait. The trajectories correspond to the gait patterns shown in Fig 7, where the swing movement time is 1.33 s. The trajectories of Pattern A and B is the same because the hip trajectory is not depend on the maximum toe height. The states at toe off for all trajectories are satisfied with the condition of forward progression. Although trajectory of Pattern E is in the area of backward falling around the equilibrium point, little risk of the falling backward exists because the hip joint is sufficiently inside of the base of support.

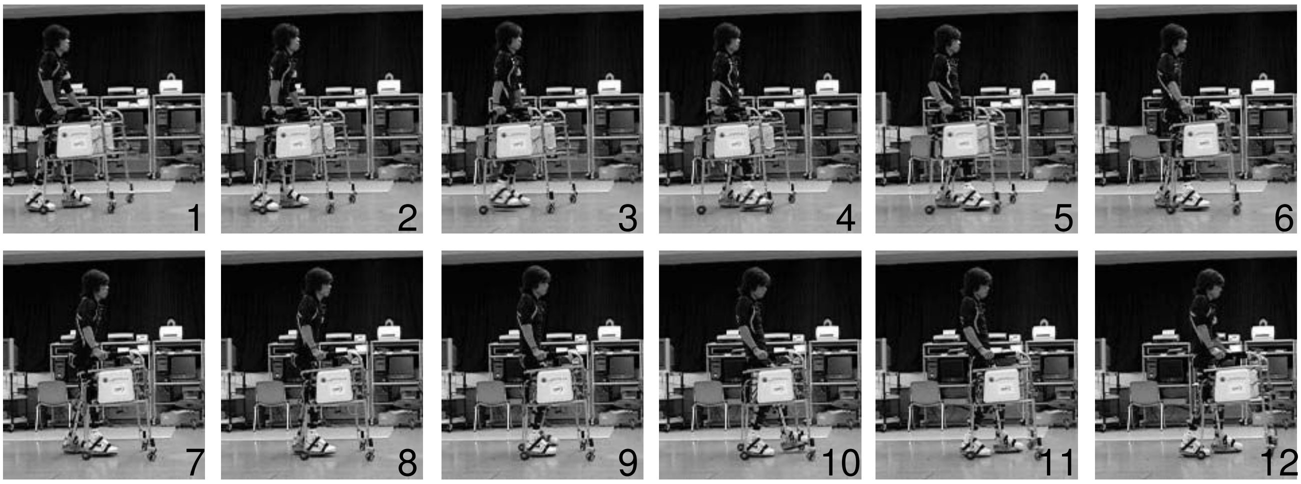


Fig. 9. Pictures of two step movement of a paraplegic subject with WPAL and a walker. The time interval of the picture is 333 ms. Pictures 1 and 7 are the motion in the preswing phase, pictures 2, 3, 4, 8, 9 and 10 are the motion in the midswing phase, and pictures 5, 6, 11 and 12 are the motion in the arm movement phase.

Fig. 9 shows the gait movement of a typical paraplegic patient using WPAL and a walker. For this subject, the stride length was 0.47 m and maximum height of toe was 0.02 m. The duration of the swing was 1.33 s, and the duration of the arm movement phase was 0.67 s. The subject walked smoothly during about 30 min and 145 m without a break, and there was no falling accident. In addition, we confirmed sufficient toe clearance (Fig. 9–3) and smooth heel contact (Fig. 9–4, 5). We confirmed that the other subjects could also walk with WPAL.

VI. DISCUSSION

In this study, we propose a gait pattern generation method of a legged locomotor device for paraplegics. The following requirements are considered in the gait pattern generation: (1) not-falling backward, (2) sufficient foot-floor clearance during swing movement, and (3) adjustablity of stride length and gait period. To avoid a backward falling, a condition for forward progression is formulated on the basis of an inverted pendulum model. Four variables of the external coordinates are determined by the minimum jerk trajectories so that the requirements are satisfied. The joint angles are calculated by inverse kinematics from the variables.

It was confirmed that the proposed method could provide smooth patterns and could apply not only the gait patterns but also the transfer movements between standing and walking. We demonstrated that the legged locomotor following the trajectories of the proposed method enabled the paraplegics to walk smoothly. These results suggests that the proposed method is effective for legged locomotor control of paraplegics.

Some parameter values of the gait pattern should be determined automatically by machine learning and/or a human interface based on user's intension. In the future works, we will design the framework of an online pattern generator including the machine learning and the human inteface for the cooperations of human and machine.

VII. ACKNOWLEDGMENTS

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