Analysis of Inertial Motion in Swing Phase of Human Gait and Its Application to Motion Generation of Transfemoral Prosthesis*

Takahiro Wada[1], Hiroshi Sano[1], and Masahiro Sekimoto[2]

***Abstract*— The goal of this study was to develop a transfemoral prosthesis that realizes a natural and smooth gait. To realize such prostheses, the smoothness or goodness of walking in the swing phase should be investigated. So far, many studies have investigated the human gait. However, the goodness of walking has not been clearly defined. Passive walking has drawn much attention because it includes the essence of the gait with very energy-efficient walking. The human gait is thought to effectively utilize the dynamics of the body for a smooth gait. Investigating the human gait should help in building smoothly walking robots and realizing prostheses that make it easy to walk.**

The present paper proposes a method to evaluate the inertial motion of the human gait in the swing phase and a method to generate the joint motion profile of a transfemoral prosthesis that utilizes its inertial motion. First, an inertial motion index is proposed to evaluate the closeness of the given motion to the inertial motion of multilink systems. The gait of healthy participants in the swing phase was analyzed using the proposed index. The results showed that inertial motion is effectively used in the middle of the swing phase. Based on the results, a motion generation method is proposed for the knee joint of a prosthesis that utilizes the inertial motion in the middle of the swing phase. Simulation results with the method showed that a smoother change in torque was realized compared to another method that uses the measured joint motion of human gait as the desired position.

I. INTRODUCTION

Recent advancements in the mechanism and control method of transfemoral prostheses have drastically improved the gaits of amputees and realized a safer stance phase for the prosthesis. In particular, computer-controlled transfemoral prostheses have significantly contributed to dramatically increased safety when walking with a prosthesis [1][2][3][4]. To realize a higher level of activities of daily living (ADL) for prosthesis users, the development of prostheses that realizes walking easiness is important. The swing phase of the prosthesis has a great impact on the walking easiness/gait smoothness, while the function of the knee joint in the stance phase plays an important role in guaranteeing safe walking. Computer-controlled prostheses are being used to successfully increase the gait smoothness in the swing phase [2][3][4].

The goal of this study was to develop a transfemoral prosthesis that realizes a smooth natural gait. To realize such a prosthesis, the smoothness or goodness of walking in the swing phase should be investigated. Many research studies have been conducted on evaluating the gait with a prosthesis. For instance, different prostheses have been compared based on the joint angles and moments as well as the energy consumption for each joint using conventional gait analysis methodology[3][4]. However, there has been little research on quantifying the easiness of walking in terms of dynamics even though such an approach is important for applying the results to the design methodology of prostheses. On the other hand, many studies have investigated the human gait. However, the goodness of walking has not been clearly defined. Passive walking[5] has drawn much attention because it includes the essence of gait with very energy-efficient walking[6][7]. Humans are thought to effectively utilize the dynamics of the body for a smooth gait. Investigating the human gait should help in building smoothly walking robots and realizing prostheses that make it easy to walk.

The present paper proposes a method to evaluate the inertial motion of the human gait in swing phase and a method to generate the joint motion profile for a transfemoral prosthesis that utilizes its inertial motion. Sekimoto et al. proposed the inertia-induced measure as an index to quantify the inertial properties of the body dynamics[8] for analysis of skillful human motions[9]. Wada et al. applied this inertia-induced measure to transfemoral prosthesis walking [10][11]. However, the inertia-induced measure is limited in that it quantifies the closeness of the motion to the inertial motion but does not consider the effect of gravity.

The present paper proposes the inertial motion index to evaluate the closeness of a given motion to the inertial motion of multilink systems. The proposed method can consider the gravity effect because it quantifies the closeness of the measured posture to the posture of an inertial motion with gravity by using the Riemannian distance between them. The gait of healthy participants in the swing phase is analyzed using the proposed index. Based on the results, a motion generation method is proposed for the knee joint of a prosthesis that utilizes the inertial motion in the middle of the swing phase. Numerical simulations will show the effectiveness of the proposed method.

II. EVALUATION OF CLOSENESS TO INERTIAL MOTION IN MULTILINK SYSTEM

A. Evaluation of closeness of two different postures

This study considered multilink systems whose dynamics are expressed in eq.(1).

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[1]Takahiro Wada and Hiroshi Sano are with School of Information Science and Engineering, Ritsumeikan University, 1-1-1 Nojihigashi, Kusatsu, Shiga, 525-8577, Japan {twada@fc, is0069ss@ed}.ritsumei.ac.jp

[2]Masahiro Sekimoto is with Faculty of Engineering, University of Toyama, Toyama, Japan sekimoto@eng.u-toyama.ac.jp

$$H(\boldsymbol{q})\ddot{\boldsymbol{q}} + \frac{1}{2}\dot{H}(\boldsymbol{q})\dot{\boldsymbol{q}} + S(\boldsymbol{q},\dot{\boldsymbol{q}})\dot{\boldsymbol{q}} + g(\boldsymbol{q}) = \boldsymbol{\tau} \tag{1}$$

where $\boldsymbol{q} \in R^n$ and $\boldsymbol{\tau} \in R^n$ denote the joint displacement vector and generalized force vector, respectively. $H(\boldsymbol{q})$ denotes the inertia matrix, and $S(\boldsymbol{q},\dot{\boldsymbol{q}})$ denotes a skew-symmetric matrix representing the centrifugal and Coriolis forces. Vector $g(\boldsymbol{q})$ denotes the gravity term. The closeness of two given time series patterns of postures is quantified for an n DOF multilink system $\boldsymbol{q}(t)$ and $\bar{\boldsymbol{q}}(t)$, which are defined in $t \in [0, T]$.

A set of all postures of a multilink system, where one end is fixed on the inertial frame and the other ends are free, can be regarded as a Riemannian manifold when the inertia matrix of the system is utilized as the Riemannian metric [12]. A trajectory that connects the posture $\boldsymbol{q}(t)$ to another posture $\bar{\boldsymbol{q}}(t)$ on the manifold is described as $\boldsymbol{p}(s)$ by introducing the parameter $s \in [a, b]$, where $\boldsymbol{p}(a) = \boldsymbol{q}(t)$, $\boldsymbol{p}(b) = \bar{\boldsymbol{q}}(t)$. Consider the following variational problem for the manifold.

$$d(\boldsymbol{q}(t),\bar{\boldsymbol{q}}(t)) = \min_{\boldsymbol{p}(s)} \int_a^b \sqrt{\sum_{i,j=1}^{n} h_{ij}(\boldsymbol{p}) \frac{dp_i(s)}{ds}\frac{dp_j(s)}{ds}}\, ds \tag{2}$$

where h_{ij} denotes each element of the inertia matrix $H(\boldsymbol{q})$. This length can be regarded as the distance between these two points on the manifold by introducing elements of the inertial matrix into the metric; this is called the Riemannian distance[12]. The trajectory that satisfies the right-hand side of eq.(2) is referred to as geodesic. The geodesic equation can be obtained as eq.(3) by solving the variational problem on the right-hand side of eq.(2).

$$H(\boldsymbol{q})\ddot{\boldsymbol{q}} + \frac{1}{2}\dot{H}(\boldsymbol{q})\dot{\boldsymbol{q}} + S(\boldsymbol{q},\dot{\boldsymbol{q}})\dot{\boldsymbol{q}} = \boldsymbol{0} \tag{3}$$

Eq.(3) coincides with the motion of the multilink system without any external force other than the inertial force under no gravity. The distance in eq.(2) can be naturally interpreted as the distance between two postures weighted by inertia parameters. The distance is also invariant with the range defined by $s \in [a, b]$ [12].

Consequently, the closeness of the two motions $\boldsymbol{q}(t)$ and $\bar{\boldsymbol{q}}$ during $[t_1, t_2]$ can be quantified in eq.(4) by integrating (2).

$$L(t_1, t_2) = \frac{1}{t_2 - t_1}\int_{t_1}^{t_2} d(\boldsymbol{q}(t),\bar{\boldsymbol{q}}(t))dt \tag{4}$$

B. Evaluation of Closeness to Inertial Motion

The inertial motion pattern during walking with a prosthesis was considered. In the present study, the motion of a multilink system, as given in eq.(1), without any external force/torque is referred to as inertial motion, which is described by eq.(5).

$$H(\boldsymbol{q})\ddot{\boldsymbol{q}} + \frac{1}{2}\dot{H}(\boldsymbol{q})\dot{\boldsymbol{q}} + S(\boldsymbol{q},\dot{\boldsymbol{q}})\dot{\boldsymbol{q}} + g(\boldsymbol{q}) = \boldsymbol{0} \tag{5}$$

The following is a method to evaluate the closeness of a given measured joint motion to the inertial motion developed from the measured posture and velocity at a given moment. Suppose that the measured data $\boldsymbol{q}_m(t)$, $\dot{\boldsymbol{q}}_m(t)$, $t \in [0, T]$ are given. Here, the initial value problem of eq.(5) with the initial value $\boldsymbol{q}(t_i) = \boldsymbol{q}_m(t_i)$, $\dot{\boldsymbol{q}}(t_i) = \dot{\boldsymbol{q}}_m(t_i)$ is solved, where t_i is a certain time in $[0, T]$. The solution of the initial value problem $\boldsymbol{q}_{IM}(t)$, $t \in [t_i, t_i + \Delta w]$ $(t_i + \Delta w \le T)$ is the inertial motion of the given multilink system, where Δw is a certain time window that is determined later. The closeness of the inertial motion to the given measured motion $\boldsymbol{q}_m(t)$ in the time window $t \in [t_i, t_i + \Delta w]$ is quantified as follows. As shown in Fig.1, the closeness of the two postures $\boldsymbol{q}_m(t)$ and $\boldsymbol{q}_{IM}(t)$ is defined as eq.(6) using the Riemannian distance based on eq.(2).

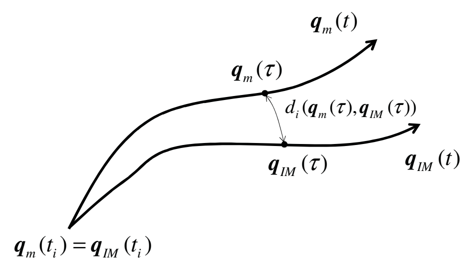


Fig. 1. Schematic image of closeness evaluation between measured trajectory and trajectory of inertial motion as calculated by initial value problem.

$$d_i(\boldsymbol{q}_m(t),\boldsymbol{q}_{IM}(t)) = \min_{\boldsymbol{q}(s)} \int_a^b \sqrt{\sum_{j,k=1}^{n} h_{jk}(\boldsymbol{q}) \frac{dq_j(s)}{ds}\frac{dq_k(s)}{ds}}\, ds \tag{6}$$

where $\boldsymbol{q}(s)$ $(s \in [a, b])$ satisfies $\boldsymbol{q}(a) = \boldsymbol{q}_m(t)$ and $\boldsymbol{q}(b) = \boldsymbol{q}_{IM}(t)$. The geodesic trajectory of the solution of eq.(6) coincides with (3) and is obtained by solving the boundary value problem with $\boldsymbol{q}(a) = \boldsymbol{q}_m(t)$, $\boldsymbol{q}(b) = \boldsymbol{q}_{IM}(t)$. The total closeness during the given time window $t \in [t_i, t_i + \Delta w]$ is defined as eq.(7).

$$L(t_i, t_i + \Delta w) = \frac{1}{\Delta w}\int_{t_i}^{t_i+\Delta w} d_i(\boldsymbol{q}_m(t),\boldsymbol{q}_{IM}(t))dt \tag{7}$$

This $L(t_i, t_i+\Delta w)$ is referred to as the inertial motion index because a smaller $L(t_i, t_i+\Delta w)$ means that the given motion $\boldsymbol{q}_m(t)$ is closer to the inertial motion. $L = 0$ means that $\boldsymbol{q}_m(t)$ completely coincides with the inertial motion in the given time window.

The inertial motion index can be determined for every evaluation interval Δt [s] in the time domain $[0, T]$ by dividing the time region into N parts. This means that the inertial motion index is evaluated every $t_i = i \times \Delta t$ $(i = 0, 1, \cdots, N-1, N(= T/\Delta t))$. In general, the time window Δw and evaluation interval Δt can be chosen independently. For the sake of simplicity, $\Delta t = \Delta w$ was assumed in this study.

III. Evaluation of Inertial Motion in Human Gait

A. Experimental methods

Experiments on normal walking were performed to analyze the inertial motion patterns.

1) Experimental apparatus: A straight and flat path with a length of 7.5 m was used. To measure the positions of the lower extremities and orientation of the trunk, 13 reflective markers were attached to the greater trochanter, knees, ankles, heels, fifth metatarsals, and toes of both feet as well as the back. The three-dimensional positions of the markers were measured using a motion capture system (motion analysis) with eight cameras at 100 Hz(Fig.2).

Fig. 2. Experimental appratus

2) Participants and procedure: Four males who gave informed consent and with no neuromuscular disorders or functional limitations in their lower extremities participated in the experiments. Their mean (SD) age was 22 (SD 0.4) years. Their mean (SD) body height and weight were 1.71 (SD 0.04) m and 67.6 (SD 15.0) kg, respectively.

The participants were instructed to walk along the predetermined straight path from the start point at their own selected speeds after an auditory cue. There were no instructions for the walking velocity or stride. Three walking trials were recorded for each subject after some practice walks.

3) Analysis method: Fig.3 illustrates the link segment model of walking in the swing phase. For the sake of simplicity, the motions of the ankle and knee of the stance leg were omitted to reduce degrees of freedom of the system. Link 1 denotes the upper and lower legs. Link 2 denotes the upper leg of the other side. Link 3 denotes the lower limb including the foot. Link 0 represents the whole trunk. Link 0 was assumed to be vertical throughout the walking. Angle q_1 denotes the ankle joint of the stance side, and the ankle joint was assumed to be fixed to the floor. The dynamics of the link segment model is given by eq.(1).

B. Experimental results

Fig.4 illustrates the joint angles from the toe off to the heel contact of a foot. The blue lines illustrate the measured joint angles in the experiments. The red lines illustrate the joint angles in the inertial motion that was calculated as solutions to the initial value problem. Each figure has ten red lines because the swing phase was divided into ten regions, and the inertial motions were calculated for all regions. As shown in the figure, joint q_1 appeared to be almost linear, which

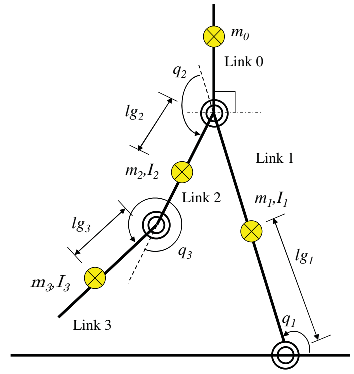


Fig. 3. Link segment model of swing phase walking

coincided with its inertial motion. For q_3, the knee angle demonstrated large flexion in the middle and was almost fully extended just before the end of the swing phase and then flexed a bit to the heel contact. Note that inertial parameters such as the mass, moment of inertia, and position of the center of mass for each segment, which are required to calculate the inertial motion index, were estimated using a regression model of Japanese athletes[13].

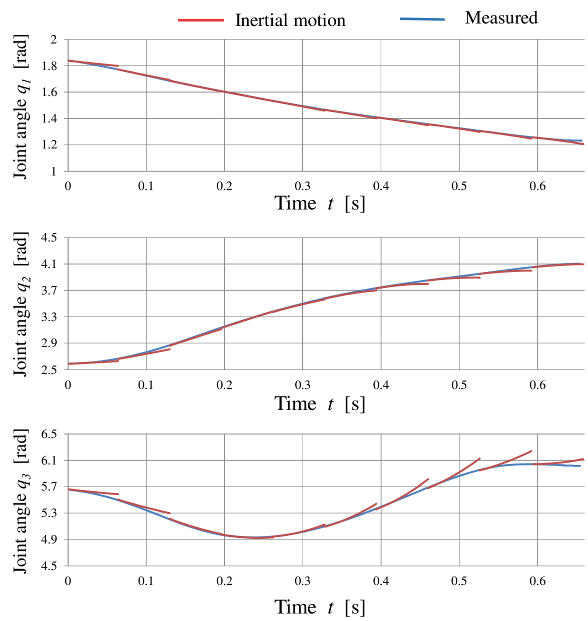


Fig. 4. Time series of joint angles

C. Inertial motion index

As an example, Fig.5 illustrates the results of the inertial motion index with the knee joint angle and its angular velocity of a participant. The inertial motion index was large at the beginning and end of the swing phase while relatively small around the middle of the swing phase.

The swing phase was divided into three phases Lines A and B denote the timing of the maximum knee flexion and maximum knee joint velocity, respectively, and these lines were used to divide the swing phase into three phases.

Let us divide the swing phase into three phases of the beginning, middle, and end using the timing of the maximum knee flexion and maximum knee joint velocity. Fig.6 shows the inertial motion index of all participants when calculated for the three phases. The index values at the middle of the swing phase were the smallest, so the inertial motion was greatest in the middle phase as well as the final phase. It should be noted that Euclidian norm at the final phase seems to be the significantly larger than the other phases from the visula differences in q_3. This difference demonstrated the importance of use of appropriate metric rather than Euclidian norm even though its computation cost is much higher.

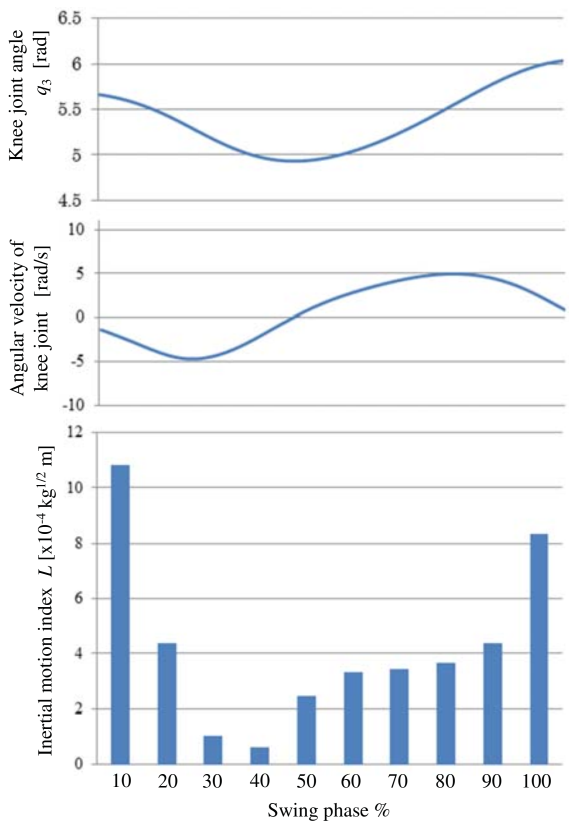


Fig. 5. Inertial motion index

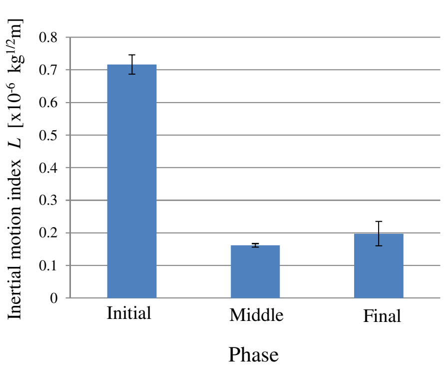


Fig. 6. Inertial motion index divided into three phases

IV. Generation method of natural knee motion

A. Overview

A motion generation method is proposed for the prosthetic knee joint that is based on the inertial motion. The previous section showed that a motion close to the inertial motion appears at the middle and the end of the swing phase. Thus, the proposed method utilizes the inertial motion of the knee joint around the middle of the swing phase as its desired angle. This is based on the hypothesis that the active use of the inertial motion in mid-swing increases the smoothness of walking.

A schematic image of the proposed method is shown in Fig.7, where the stance phase is divided into three subphases. In the middle, the inertial motion that ends at point Q is utilized as the desired angle. Then, the desired trajectory from point Q and the heel contact is generated by another inertial motion. Point P is not dealt with explicitly in this method and treatment of initial phase expressed later.

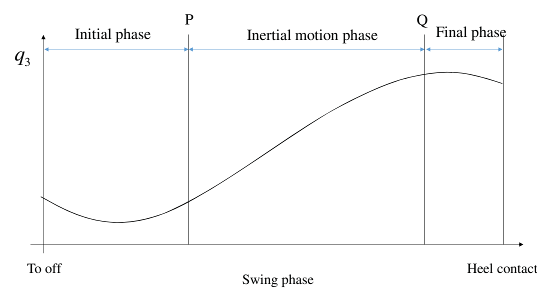


Fig. 7. Schematic image of motion generation

B. Method to determine inertial motion around middle phase

Let the joint angle and angular velocity of the knee joint be $q_3(t_Q)$, $\dot{q}_3(t_Q)$. Given the position of Q, the inertial motion as given by eq.(5) that reaches $q_3(t_Q)$, $\dot{q}_3(t_Q)$ is calculated as follows. Time-reversal symmetry is satisfied when the inertial motion described in eq.(5) is considered: that is, without any friction. With this feature, time-reversal integration of eq.(5) can be represented as follows by introducing the reversed time t^-, which is defined as $t^- = -t$.

$$
\begin{aligned}
\frac{d}{dt^-}\begin{bmatrix} \boldsymbol{q} \\ \dot{\boldsymbol{q}} \end{bmatrix} &= \frac{d}{dt}\frac{dt}{dt^-}\begin{bmatrix} \boldsymbol{q} \\ \dot{\boldsymbol{q}} \end{bmatrix} = -\frac{d}{dt}\begin{bmatrix} \boldsymbol{q} \\ \dot{\boldsymbol{q}} \end{bmatrix} \\
&= \begin{bmatrix} -\dot{\boldsymbol{q}} \\ H(\boldsymbol{q})^{-1}\{\frac{1}{2}\dot{H}(\boldsymbol{q})\dot{\boldsymbol{q}} + S(\boldsymbol{q},\dot{\boldsymbol{q}})\dot{\boldsymbol{q}} + \mathbf{g}(\boldsymbol{q})\} \end{bmatrix} \quad (8)
\end{aligned}
$$

Because eq.(8) satisfies the time-reversal symmetry, the solution of this equation can be followed by reversing the time axis.

Suppose that the measured $q_1(t)$ and $q_2(t)$ are given. Suppose that the data are approximated by the third degree polynomial in time as $q_1^m(t)$ and $q_2^m(t)$. Substituting them into eq.(8) yields eq.(9).

$$\begin{bmatrix} \frac{dq_3}{dt^-} \\ \frac{d^2 q_3}{(dt^-)^2} \end{bmatrix} = \begin{bmatrix} -\dot{q}_3 \\ [0\ 0\ 1]\, H(\tilde{\boldsymbol{q}})^{-1}\{\frac{1}{2}\dot{H}(\tilde{\boldsymbol{q}})\dot{\tilde{\boldsymbol{q}}} + S(\tilde{\boldsymbol{q}}, \dot{\tilde{\boldsymbol{q}}})\dot{\tilde{\boldsymbol{q}}} + \mathbf{g}(\tilde{\boldsymbol{q}})\} \end{bmatrix} \tag{9}$$

where $\tilde{\boldsymbol{q}} = [q_1^m, q_2^m, q_3]^T$.

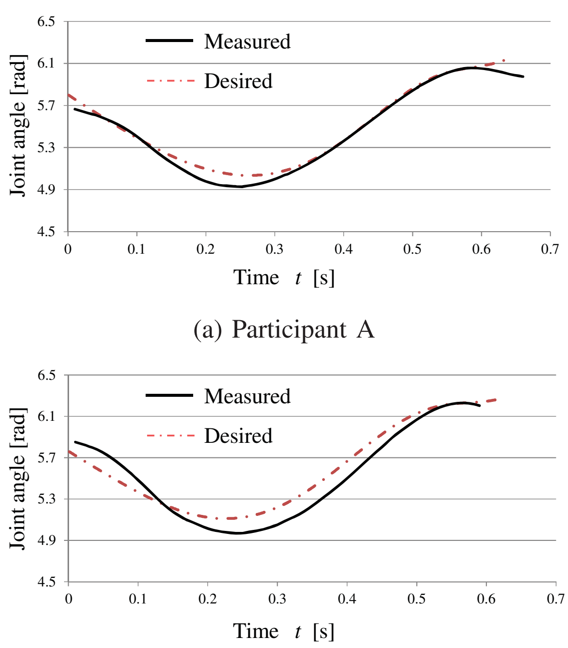
With this method, candidates of the knee motion are calculated using the time-reversal integration of eq.(9) with various initial values around point B of Fig.5. Point Q is determined in order to maximize the time duration when the solution of difference between the time-reversal integral and the measured $q_3(t)$ at each moment is small.

C. Trajectory generation in final phase

The knee motion trajectory from point Q to the heel contact is generated by solving the forward initial value problem of eq.(1) with the initial values of $q_3(t_Q) = q_3^m(t_Q)$ and $\dot{q}_3(t_Q) = \dot{q}_3^d(t_Q)$, where t_Q denotes the time when the knee is at point Q. Assume that the knee joint is designed to keep its fully extended position $q_3 = 2\pi$ once it is reached though it is not guaranteed.

D. Generated motion of knee joint

Fig.8 shows the knee joint motions $q_3^d(t)$ generated by the proposed method using $q_1^m(t)$ and $q_2^m(t)$ of two participants with prostheses for which the inertia parameters are given in TableI. The generated knee motion was similar to or not very far from that of the measured knee motions of the participants even when only inertial motion was used.



(a) Participant A

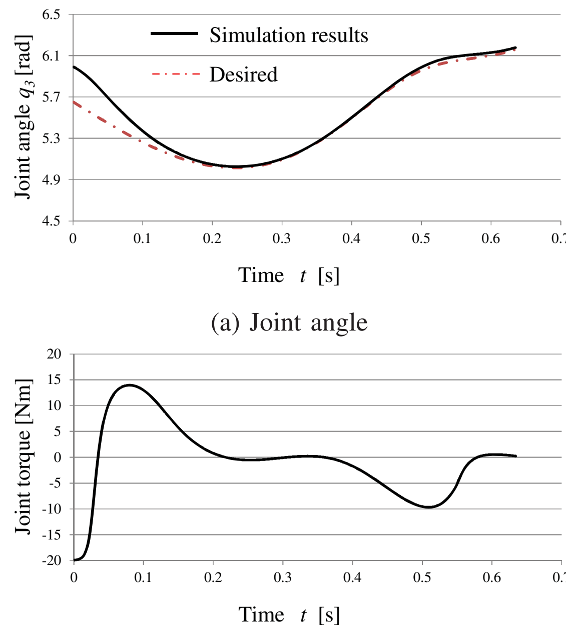
(b) Participant B

Fig. 8. Generated knee joint motion

TABLE I

INERTIA PARAMETERS OF LOWER LEG OF PROSTHESIS

m_3 [kg]	l_3 [m]	l_{g3} [m]	I_3 [kgm2]
1.0	0.501	0.425	0.0238



(a) Joint angle

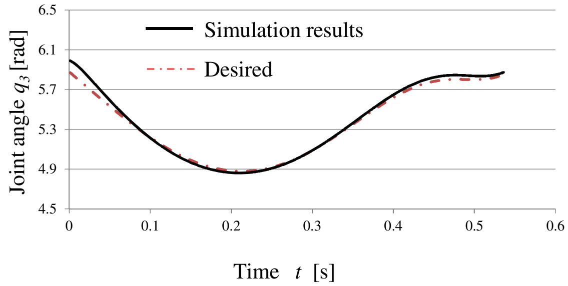
(b) Joint torque

Fig. 9. Knee joint angle and torque in numerical simulation (Participant A)

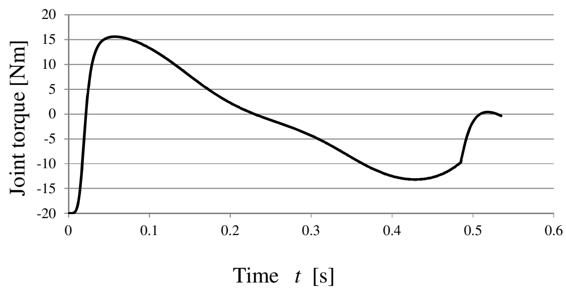
E. Simulation of motion control of knee joint

A numerical simulation was performed to show the effectiveness of the proposed motion generation method. A simple PD feedback controller of the knee joint angle was used with the desired angle of the knee joint $q_3^d(t)$, which was generated by the proposed method offline. Note that the feedback gain was set a low value because it is thought that a large gain affects thigh motion with small deviation from the desired trajectory and it should be avoided to increase comfort. This low gain yields larger difference existed at the beginning of the phase.

Figs.9 and 10 illustrate the knee joint motion and joint torque of two participants with the proposed method as example. These figures show that the desired motion and the simulation results agreed with each other. For participant A, small torque region appears in the middle of the swing phase. Fig.11 demonstrate the stick figure of the resultant gait by the proposed method. For comparison, the numerical simulation of the PD feedback that uses the measured knee joints angle $q_3^m(t)$ as the desired motion is performed. Fig.12 shows the joint torque of the knee joint of the participants as an example with this method. Comparing the joint torque in the proposed method and another candidate demonstrates that the proposed method realized smoother changes in the torque through the swing phase.



(a) Joint angle



(b) Joint torque

Fig. 10. Knee joint angle and torque in numerical simulation (Participant B)

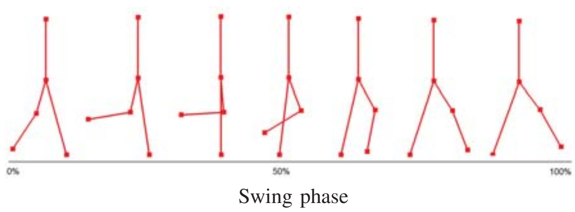
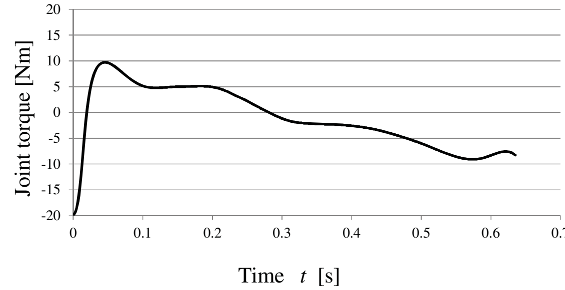
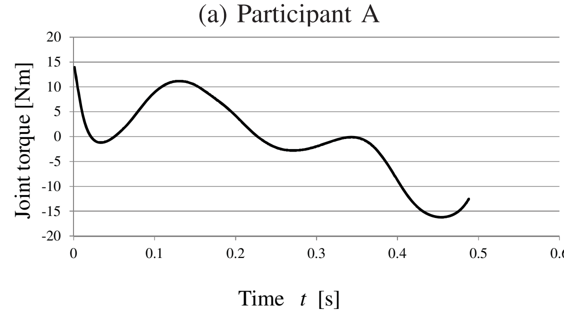


Fig. 11. Stick pictures of gait simulation results



(a) Participant A



(b) Participant B

Fig. 12. Knee joint torque by another method that uses the measured knee joint angle as the desired motion

V. Conclusions

The proposed inertial motion index is a new method to quantify the closeness of a given motion to its inertial motion in multilink systems and is based on the Riemannian distance in the configuration manifold. Analysis of the inertial motion in the human gait using the proposed index showed that the inertial motion is effectively used in the middle of the swing phase. A method for motion generation of the knee joint that utilizes the inertial motion in the middle swing phase was proposed. A smooth gait was generated with this method. The torque of the prosthetic knee joint actuator changed more smoothly with the proposed method than when the measured joint motion of the human gait was used to determine the desired motion. It is expected that smoother change in the knee joint torque of the prosthesis affect the thigh motion less, and it leads to ease of walking.

Future work will involve experimental verification of the proposed motion generation method. Subjective rating of the easiness of walking with a prosthesis using the proposed control method will be investigated.

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