Evaluation of Inertia Matching of Trans-Femoral Prosthesis based on Riemannian Distance

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***Abstract*—A method to quantify the matching of inertia property of a trans-femoral prosthesis to a given user's skill will be explored for better design of the prosthesis. Advancements in the mechanism and the control method of trans-femoral prostheses have drastically improved the gait of amputees. However, realization of a natural gait has not been investigated in detail even though such smooth gait is important to increase the amputee's activities of daily living (ADL). Inertia of the prosthesis plays an important role in natural and smooth gaits during the swing phase. We suppose that goodness of gait or easiness of walking is strongly related to effective use of the prosthesis inertia. In this paper, we will attempt to quantify matching of inertia property of the prosthesis to a given user's skill / condition from the effective use of the prosthesis inertia during gait in the swing phase. Gaits of an expert prosthesis user will be measured by changing inertia properties. Effective use of inertia property of the prosthesis in gait is evaluated by closeness of the gait to the inertia-induced motion using inertia-induced measure. In addition gait of a novice prosthesis user will be measured to compare with the expert's results.**

I. Introduction

PEOPLE whose leg was amputated above the knee use a trans-femoral prosthesis that has a mechanical knee joint. In such a prosthesis, mechanism and control method are important to guarantee safe locomotion. For instance, flexion of the knee should be prevented in the stance phase of walking, but the knee needs appropriate flexion in the swing phase to prevent stumbling and falling. From such a viewpoint, much research and development has been conducted on the knee flexion mechanism and control method to determine appropriate timing of heel contact on the floor. In particular, computer-controlled trans-femoral prostheses significantly contribute to dramatically increasing safety in walking with a prosthesis [1], [2], [3].

In order to realize higher activities of daily living (ADL) of prosthesis users, development of the prosthesis realizing easiness of walking is important. Swing phase of the prosthesis has great impact to easiness of walking / smoothness of gait. Computer-controlled prostheses increase gait smoothness in the swing phase [2], [3], [4].

In this research, we suppose that matching a prosthesis to a user according to user's skill and/or preference is important. There are several factors to be matched for instance the socket, alignment, control parameter for computer-controlled prosthesis, and inertia property. For socket matching and alignment, there are many efforts so far because they have strong impact to the safety in the stance phase as well as swing phase stability and the prosthetist and orthotist have talented with these adjustment as their basic and important skill. The control parameters should adjust after prescription of the prosthesis. In addition, there are many research studies on matching of inertia property of the prosthesis because it has great impact to the easiness of walk. For example, Czerniecki et al. pointed out the double pendulum motion of the trans-femoral prosthesis can be beneficial due to less energy consumption [5]. On the other hand, Selles et al investigated effect of the inertia properties changes on the gait with transtibial prosthesis and pointed out importance of compromise between natural motion and less torque motion[6]. Furthermore, Theroux-Jones et al. proposed a method to calculate optimal prosthesis inertial parameters based on computer simulation to reduce joint torque realizing symmetric gait to the intact side[7]. However, research on quantifying the inertia matching from the viewpoint of dynamics is not found even though such an approach is important for applying the results to design methodology of prostheses.

In this research, gait with the trans-femoral prosthesis from the viewpoint of effective use of its inertia property to establish design methodology of inertia matching of the prosthesis. The effective use of the inertia is evaluated by inertia-induced measure[8]. In our previous work, gaits were measured by changing the inertia parameters of the prosthesis by adding a weight on the leg part [9]. The results showed the strong relevance between subjective evaluation of easiness of walking and the degree of inertia-induced motion. In the inertia condition with the best subjective evaluation, the gait is closer to the inertia-induced motion. But, the paper did not deal with transient change of the degree of inertia-induced motion in time even though it can reflect user's skill for prosthesis walk.

Thus, the swing phase will be divided into three phases and then the degree of inertia-induced motion of each phase will be evaluated in this paper. The relationship between subjective evaluations of easiness of walking for each inertia

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condition and the degree of inertia-induced motion will be investigated. In the experiments, and an expert and a novice of the prosthesis walk participate. The results will be compared to investigate effect of the skill of the prosthesis walking on the degree of inertia-induced motion.

II. Evaluation of Swing Phase Gait with Trans-femoral Prosthesis Based on Effective Use of Inertia Property

A. Link Segment Model of Swing Phase Gait with Prosthesis

Motion in the swing phase with the prosthesis is modeled by Fig.1. Assume that motions of the ankle and the knee of the intact end can be ignored. Link 1 denotes the upper and the lower legs. Link 2 denotes the upper leg of the amputated end and the socket part. Link 3 denotes the knee joint, the lower leg, and the foot part of the prosthesis. Link 0 represents the human's whole trunk. Assume that link 0 is vertical through walking. Angle q_1 denotes the ankle joint of the intact end and it is assumed that the ankle joint is fixed at the floor.

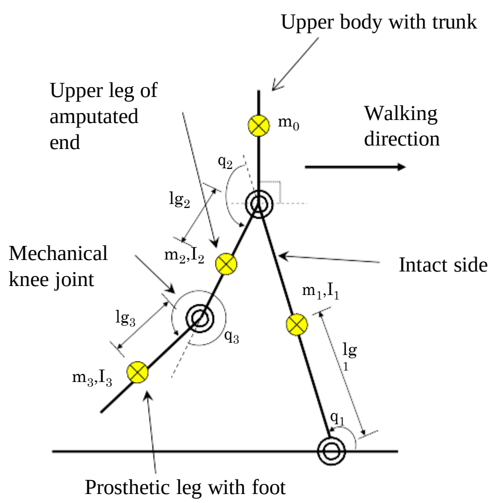


Fig. 1 Four link model of prosthetic leg walking.

B. Index to Evaluate Closeness to Inertia-Induced Motion

Let us consider to quantify goodness of the gait from the viewpoint of effective use of the prosthesis inertia property. Now, the motion derived by inertial force is formulated. Dynamics of n DOF multi-body system including model in Fig.1 can be represented by the Lagrangian form in eq.(1).

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

where $\boldsymbol{q} \in R^n$ denotes joint angle of the mechanism. Matrices $H(\boldsymbol{q}) = [h_{i,j}]$ and $S(\boldsymbol{q},\dot{\boldsymbol{q}})$ denote inertia matrix and skew symmetric matrix related to centrifugal and Colioris force, respectively. Vectors $\boldsymbol{g}(\boldsymbol{q})$ and τ represent gravitational force and external joint torque, respectively. Now, consider to regard effective use of the inertia property as closeness of the given gait to that of inertia-induced motion.

The set of all postures of the system can be regarded as Riemannian manifold [10]. A length of the trajectory connecting given two postures from $\boldsymbol{q}(a) = \boldsymbol{q}_a$ to $\boldsymbol{q}(b) = \boldsymbol{q}_b$ on the manifold can be defined as eq.(2).

$$L = \int_a^b \sqrt{\sum_{i,j=1}^{n} h_{i,j}(\boldsymbol{q})\dot{q}_i(t)\,\dot{q}_j(t)}\;dt \qquad (2)$$

where component of inertia matrix $h_{i,j}(\boldsymbol{q})$ is regarded as Riemannian metric [10]. Trajectory minimizing eq.(2) is called geodesic and its length $d(\boldsymbol{q}_a,\boldsymbol{q}_b)$ given in eq.(3) is called Riemannian distance.

$$d(\boldsymbol{q}_a,\boldsymbol{q}_b) = \inf \int_a^b \sqrt{\sum_{i,j=1}^{n} h_{i,j}(\boldsymbol{q})\dot{q}_i(t)\,\dot{q}_j(t)}\;dt \qquad (3)$$

Equation of geodesic obtained by solving optimization problem of eq.(3) is given by eq.(4) [10].

$$H(\boldsymbol{q})\ddot{\boldsymbol{q}} + \frac{1}{2}\dot{H}(\boldsymbol{q})\dot{\boldsymbol{q}} + S(\boldsymbol{q},\dot{\boldsymbol{q}})\dot{\boldsymbol{q}} = 0 \qquad (4)$$

This equation coincides with dynamic equation with only inertial force but without external joint torque and gravitational force. This equation represents law of inertia of the multi-body system. Namely, trajectory from $\boldsymbol{q}_a$ to $\boldsymbol{q}_b$ by only inertial force without any external torque can be described by equation of geodesic eq.(4) and the length of eq.(2) is minimized in this case.

Sekimoto et al. have defined closeness of the movement given two points $[\boldsymbol{q}_a, \boldsymbol{q}_b]$ to the inertia-induced motion based on the Riemannian distance or the degree of inertia-induced motion as eq.(5) and have called inertia-induced measure [8].

$$R_N = \frac{L - d(\boldsymbol{q}_a,\boldsymbol{q}_b)}{d(\boldsymbol{q}_a,\boldsymbol{q}_b)} \qquad (5)$$

Sekimoto et al. have successfully analyzed skillfulness of reaching movement of human arm using this measure [8]. In this paper, we regard eq.(5) as quantity of effective use of inertia property, thus, examine whether the index can represent goodness of gait with prosthesis.

In calculation of eq.(5), L is calculated from eq.(2) numerically using measured data in the experiments. The Riemannian distance $d(\boldsymbol{q}_a, \boldsymbol{q}_b)$ is calculated by solving boundary value problem associated by eq.(4) with boundary condition $\boldsymbol{q}(a) = \boldsymbol{q}_a$ and $\boldsymbol{q}_b(b) = \boldsymbol{q}_b$, then calculated by eq.(2) along the obtained trajectory, say, geodesic. Please note that the angular velocity of the start and the end of the gait cannot be specified in advance but is determined after the calculation of the boundary value problem.

Swing phase is divided into three sub phases and then degree of inertia-induced motion in each sub phase is calculated to investigate effect of the inertia property of the prosthesis in each phase. Three phases are defined as shown in Fig.2 based on gait analysis method[11]. Phase I called Initial swing is defined as the duration from the start of the swing phase or toe off to feet adjacent. The feet adjacent is the time when the swinging leg passes the stance phase leg, and the two fee are side by side. Phase II called midswing is defined as the duration from the feet adjacent to tibia vertical. Phase III called terminal swing is defined as the duration from the tibia vertical to the end of swing phase or heel contact.

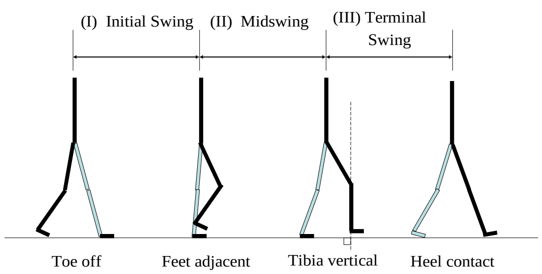


Fig. 2 Definition of phases divided by knee joint angle behavior

III. EXPERIMENTS WITH EXPERT USER[9]

A. *Experimental Method*

Fig.3 shows a participant of the experiments with a prosthesis. Fig.4 shows an experimental scene. A modular type of prosthesis leg is used in this research. A knee joint 3R95 (Otto Bock) is utilized because it is easier to add weight due to its light original weight (0.35kg). In addition, it employs the relatively simple control method such as a simple damping control. The damping coefficient can be changed by a mechanical lever in 8 levels. A foot part 1C40 (Otto Bock) and a IRC type socket that are used in the subject's daily life is used in the experiments. A male of 45 years old with left thigh amputation participated in the experiments. He uses the prosthetic leg for 11 years in his daily living. He uses C-Leg (Otto Bock) as the knee joint of the prosthesis daily.

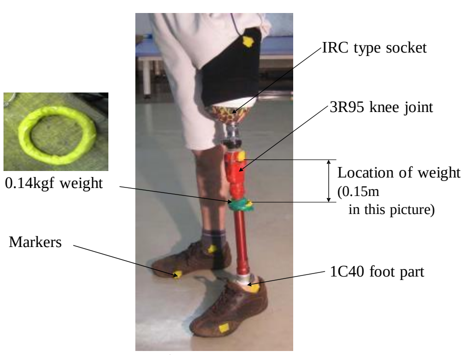


Fig. 3 Participant of experiments with prosthesis leg.


Fig. 4 Experimental Scenes.

Table I shows experimental conditions in inertia and walking velocity. There are four conditions in inertia property by adding a weight to the lower leg of the prosthesis as shown in Fig.3 in order to investigate effect of inertia property of the prosthesis on gait. Originally, no weight is attached to the lower leg. The participant walked several times to get used to walking with the prosthesis. Then, a prosthetist and orthotist of one of the authors added a 0.14kgf weight at the lower leg of the prosthesis and the best position of the weight was determined as 48% of the length of the lower leg from the knee joint by trial and error with participant's comment. Then, conditions 100% that means on the ankle joint and 0% that means on the knee joint were made for comparison by changing the position of the weight.

A digital camera EX-F1 (CASIO) was set in perpendicular to the sagittal plane and recorded gaits. In the experiments, the frame rate of the camera was set as 210Hz. Markers made by color tapes attached to the hip joint of the amputated end and the knee joint, the ankle and the toe of the prosthesis and the ankle and the toe of the intact end as shown in Fig.3. Joint angles were calculated by digitizing the video images using motion analysis software DIPP-MOTION (DITECT corp.).

The participant was instructed to walk in the most comfortable velocity determined by his subjective evaluation. In addition, he was asked to walk in slower and faster velocities in order to investigate effect of walking velocity on gait and inertia induced motion. In the experiments, resultant mean durations of swing phase in slow, normal, and fast conditions were 0.49s (SD 0.045), 0.44s (SD 0.022), and 0.40s (SD 0.019), respectively. The numbers from 1 through 12 in Table I denote the order of the experiment for each condition.

TABLE I
EXPERIMENTAL CONDITIONS

		Walking velocity		
		Normal	Slow	fast
Location of additional weight from knee joint [%]	None	1	2	3
	48	4	5	6
	100	7	8	9
	0	10	11	12

B. Identification of Inertia Property of Prosthesis and Human Body

The thigh, the lower leg, and the foot module can be separated since a modular type of the artificial leg is used as mentioned before. Thus, the parameters of each part are identified separately. Mass of each part is measured by an electronic balance. Assume that the center-of-gravity of each part is located on the line connecting two joints on the link. Thus, the center of gravity is measured from equilibrium of moment using the electronic balance and a lever. The moment of inertia is measured by hanging it at the one end and by measuring frequency of vibration by swinging the part as a pendulum (Fig.5).

Fig. 5 Measurement of Moment of Inertia.

On the other hand, inertia of the human body is calculated as follows. Inertial parameters such as mass, moment of inertia, and location of center of gravity of the intact end and the other body including trunk can be calculated based on cadaver data's proportion by Clauser [12]. In addition, inertia parameters of the upper leg in the amputated end are similarly estimated based on the Clauser's method. Assume that shape of the upper leg can be approximated by a circular truncated and inertia property is estimated using Clauser's proportion. As a result, the estimated inertia parameters in the model of Fig.1 calculated by the derived method are given in Table II.

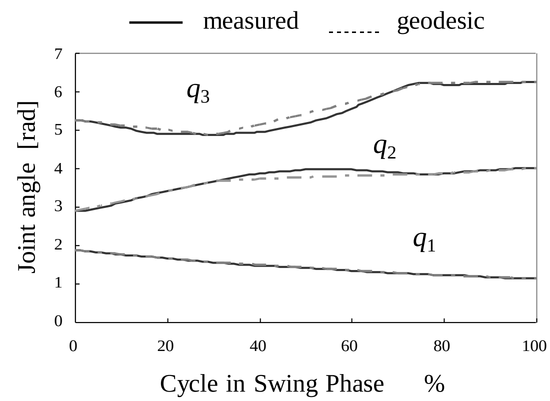
TABLE II
ESTIMATED INERTIAL PARAMETERS

Link #	Weight location	m_i [kg]	Lg_i [m]	I_i [kgm^2]	L_i [m]
0		43.91			
1		9.150	0.422	0.701	0.78
2		5.370	0.132	0.680	0.38
3	None	1.208	0.234	0.0163	0.40
	0	1.348	0.210	0.0231	
	48		0.225	0.0173	
	100		0.241	0.0168	

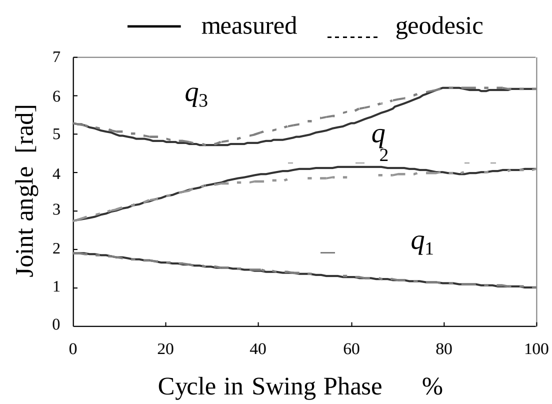
IV. EXPERIMENTAL RESULTS

A. Examples of Joint Angles

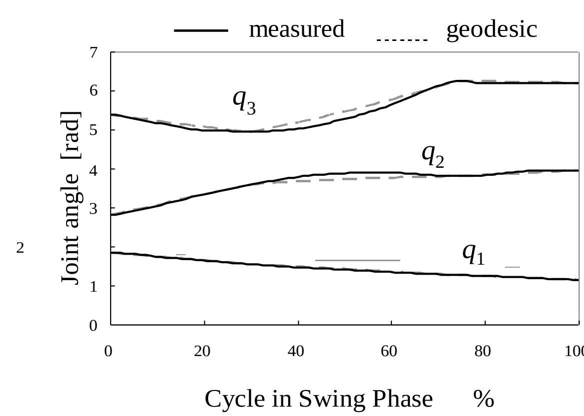
Fig.6 shows examples of joint angles in the swing phase in 48% and 100% weight location conditions. Geodesic curves that present inertia induced motions are also plotted in the figure. The geodesic was calculated by dividing the swing phase into three phases based on the q_3's extreme values as explained in Fig.2. The horizontal axis represents time normalized by the duration of the swing phase. As seen from these figures, measured joint angles are close to the inertia induced motion in three conditions. In particular, ankle joint angle q_3 is almost same as the geodesic for all conditions. There are some discrepancies between joint angles q_1 and q_2 and their geodesic curves in middle stage of the swing phase. Other major difference or tendency is not found from these figures.



(a) 48% weight location with normal velocity



(b) 48% weight location with fast velocity



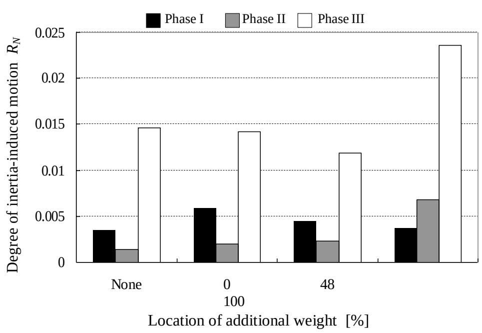
(c) 100% weight location with normal velocity

Fig. 6 Examples of joint angles with inertia induced motions.

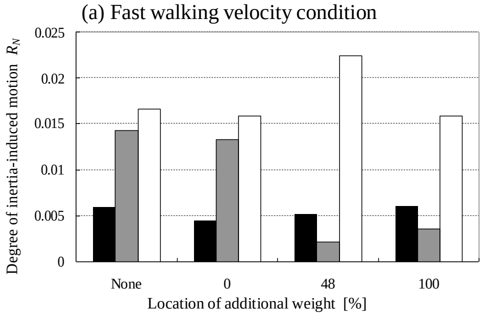
B. Degree of Inertia-induced Motion

Fig.7 shows the degree of inertia-induced motion calculated in each phase.

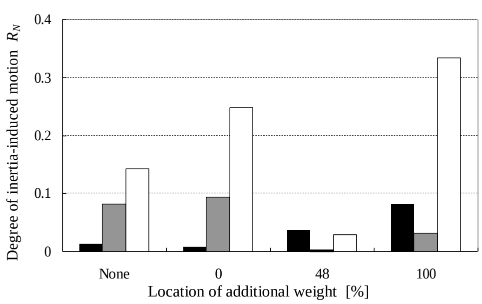
In fast velocity condition, the degree of inertia-induced motion in phase I and phase III is larger and that in phase II is smaller. It implies that at the beginning of the swing phase, active control is required and relatively less control requires and tends to use inertia-induced motion in phase II. In phase III, active control such as braking to adjust timing of heel contact is required. In normal condition, heavier condition such as 48% and 100% conditions exhibit similar results. But, in relatively lighter conditions such as none and 0% conditions, the degree of inertia-induced motion takes greater values in phase II. The degree takes larger values in slow conditions. It is understood as the effect of gravity force. With the best weight location (48%), the degrees in phase II by all velocity condition are small even though the tendency of the other weight conditions are changed according to walking velocity. It implies that the best weight location is robust to the walking velocity changes.



(a) Fast walking velocity condition



(b) Normal walking velocity condition



(c) Slow walking velocity condition

Fig. 7 Degree of Inertia-induced Motion in Each Phase

V. EXPERIMENTS WITH NOVICE USER

A. Experimental Method

Experiments with a novice user were conducted as the comparison with the expert user's results. The participant, 24 years old male has been amputated by his left thigh six months ago and uses the 3R20 (Otto Bock) as the knee joint for four months. The participant is going to train walking with the prosthesis. The prosthesis 3R20 was used in the experiments. The walking speed was only self-select speed of the participant because it is difficult for him to change his walking speed. Other experimental setups are same as the previous experiments with the expert user but location of weight is changed by participant's subjective evaluation. The inertia-property of the participant and the prosthesis were also estimated by the same way in the experiments of the exert user.

There are four conditions in inertia property by adding weight to the lower leg of the prosthesis. After several walking, a prosthetist and orthotist of one of the authors added a 0.14kgf weight at the lower leg of the prosthesis. But, the participant answered no weight condition was the best inertia property. Then, additional weight locations were selected as 0%, 21%, and 100% of the lower leg length from the knee joint. As the subjective evaluation, the participant was asked to evaluate the easiness of walk in five levels as 1 (difficult to walk), 2 (slightly difficult to walk), 3 (fair), 4 (slightly easy to walk), and 5 (easy to walk). Table III shows experimental conditions of the inertia property and the subjective evaluations. From the subjective evaluations, it is found the novice user prefers the light weight prosthesis.

TABLE III
EXPERIMENTAL CONDITIONS OF NOVICE USER AND SUBJECTIVE EVALUATION

		Subjective evaluation
Location of additional weight from knee joint [%]	None	5
	0	3
	21	4
	100	1

B. Experimental Results

Fig.8 shows the degree of inertia-induced motion with the participant as an example. It is found that the degree of the inertia-induced motion takes very large values than that of the expert prosthesis user. Especially, the degrees in phase II take very larger values. Relatively smaller values are found in the heaviest condition of 100%. In addition, there are some variation in three trials even with the same weight location and velocity conditions.

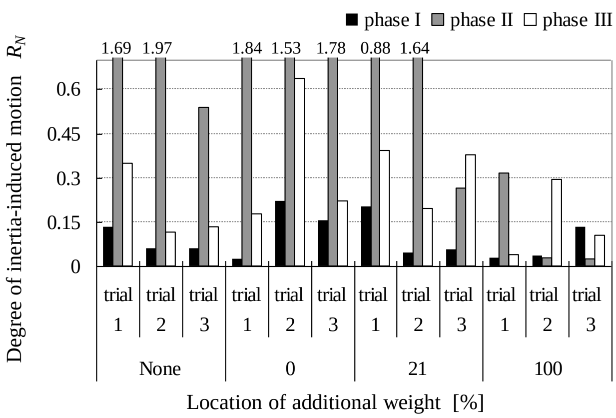


Fig. 8 Degree of Inertia-induced Motion of Novice User

VI. Conclusion

Relationship between the subjective easiness of walking with the trans-femoral prosthesis and the degree of inertia-induced motion has been investigated in order to evaluate matching of inertia property from the view point of effective use of its inertia. From the analysis of the gait of the expert user, inertia-induced measure in three phases of the swing phase showed the measure value becomes large at the initial swing and terminal swing phase of the swing phase while the measure becomes small in the midswing phase. It means the active operations are observed at the beginning and the end of swing phase while passive free motion based on inertia-induced motion can be seen in the middle phase of the swing phase. From the analysis of gait of the novice user, it is found that inertia-induced measure takes larger values than that of the exert user. In addition, there is no clear tendency in phases but the values in phase II become very large. This implies that skill of the prosthesis walk can be represented appropriate appearance of inertia induced motion in the swing phase. This can be utilized for evaluation of inertia matching of the prosthesis to the given user.

In this paper, the degree of inertia-induced motion was calculated for divided three phases. As a future study, a method to increase time resolution of the inertia-induced measure will be developed. In addition, a design method of the prosthesis inertia property will be developed as an important future studies.

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