35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Effects of Inertial Properties of Transfemoral Prosthesis on Leg Swing Motion during Stair Ascent

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***Abstract*— Stair ascent, especially the step-over-step gait, is a difficult motor task for people with transfemoral amputation. Our previous study demonstrated the effects of foot placement on the leg swing of able-bodied subjects. The study examined stair ascent with full-foot contact (FFC) and half-foot contact (HFC) as ambulation strategies. The results suggested that HFC causes the leg swing to have a greater inertial motion than FFC, as well as the applicability of the stair ascent strategy for transfemoral amputees with transfemoral prostheses without a motorized prosthetic knee joint. The present study investigated the effects of the inertial properties of a transfemoral prosthesis on leg motion during the stair ascent swing phase in simulation trials. The joint moment at the hip became smaller than that of an able-bodied subject. The peak values of the horizontal and vertical components of the joint reaction force were approximately the same as those of an able-bodied subject. These results suggest that a transfemoral prosthesis leg swing can be achieved with similar or smaller kinetic demand at the hip joint when half-foot contact on the stair steps is used as a stair ascent strategy. The mass had the largest effect of the inertial properties on the variability of the simulated kinetic parameters. The results of the present study may enhance prosthesis design with regard to the inertial properties and usability.**

I. INTRODUCTION

One of the main problems for a gait with a transfemoral prosthesis is the leg swing motion. Recently, locomotion with transfemoral prosthesis has been improved drastically. In particular, swinging has become safe and smooth through the development of motorized and microprocessor-controlled knee joint units during level walking [1, 2]. However, stair ascent is still a demanding task for transfemoral amputees.

Stair ascent is a basic activity of daily living as stairs are often encountered in a public space and at home. The ability to ascend stairs, especially the step-over-step gait (i.e., one foot placed per stair), is important for individuals with transfemoral amputation to maintain a high quality of life. Tripping over a step should be avoided during stair ascent. Successful leg swinging is a requirement for step-over-step stair ascent, but leg swinging without colliding with the stairs is difficult for transfemoral prosthesis users owing to the lack of voluntary and actively controllable prosthetic knee joint functions [3].

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Although many prosthetic knee units have been developed, including ones that are computerized or motorized, existing prosthetic knee units do not sufficiently allow persons with transfemoral amputation to ascend stairs in a step-over-step manner [3]. The development of prosthetic knees is a reasonable solution to this problem, but human motion can also be adjusted. Because the foot position on a step determines the distance to the next step at the beginning of the leg swing phase, foot placement is a key factor in the leg swing to avoid collisions between the prosthetic foot and stairs. Young and elderly people use different foot placements to negotiate obstacles [4]. Our previous study [5] showed the effects of the foot placement on the leg swing of able-bodied subjects. We examined stair ascent with full-foot contact (FFC) and half-foot contact (HFC) as ambulation strategies (Figure 1). In the FFC condition, all the parts of the plantar surface made contact with each step during the stance phase. Meanwhile, in the HFC condition, only the front half of the plantar surface made contact with each step. To keep the cadence constant, an audible click was provided by a digital metronome signaling each step. The results suggested that HFC causes the leg swing to have a greater inertial motion than does FFC.

According to the previous study [5], thigh kinematics such as linear and angular displacements of FFC and HFC are approximately the same during the swing phase. Despite the kinetic difference, the kinematic similarity between FFC and HFC can thus be characterized as human motion. The results suggest the applicability of the stair ascent strategy for transfemoral amputees with transfemoral prostheses without a motorized prosthetic knee joint. Based on the results of the previous experiment [5], a computer simulation was conducted to investigate the effectiveness of different foot placement strategies on a transfemoral prosthetic leg being swung without any actuators. The results showed that only the inertial properties specific to the HFC condition allowed the prosthesis to avoid collision with the stairs.

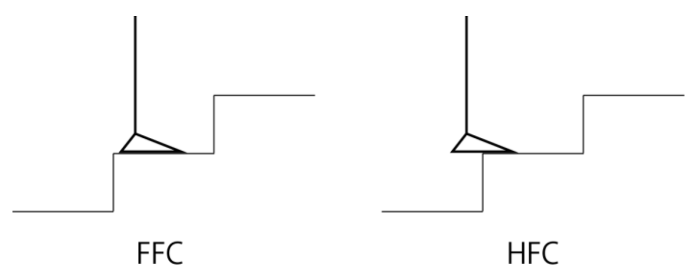

Figure 1. Foot placement strategies for stair ascent: full-foot contact (FFC) and half-foot contact (HFC).

In such prostheses, the inertial properties determine the swing motion of the prosthetic leg because the thigh kinematics is the same. The present study therefore investigated the effects of the inertial properties of a transfemoral prosthesis on the leg motion during the stair ascent swing phase through simulation trials.

II. SIMULATION METHODS

Numerical simulations were carried out to investigate the inertial properties of a transfemoral prosthesis that allows users to ascend stairs with the same thigh motion as that of able-bodied subjects (five males: age of 22.2±1.3 years; body height of 169.6 ± 6.3 cm; and body mass of 59.0 ± 8.4 kg [5]) without colliding with the steps. The transfemoral prosthesis was modeled as a rigid body pendulum below the knee, which was a passive joint. Based on the body parameters of the subjects in the experiment [5], the length of the prosthesis below the knee joint was 0.42 m, and the toe was set 0.16 m forward (Figure 2). The stairs used in the present and previous studies [5] had five steps (Figure 3). The height, depth, and width of the steps were 0.17, 0.30, and 0.90 m, respectively. The global coordinate system is shown in Figure 3. The origin was at the bottom of the first step, the x-axis was horizontal toward the direction of movement, and the direction of the y-axis was vertically upward. The simulated motion of the prosthetic leg (right leg) swing started at the toe off from the 2nd step and ended at the landing to the 4th step.

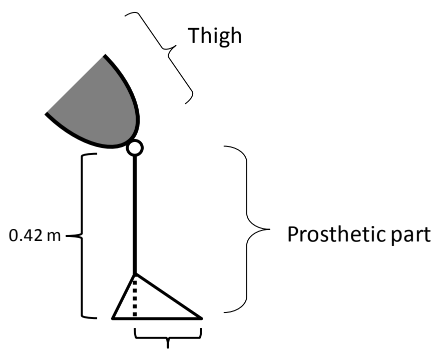

Figure 2. Model for transfemoral prosthetic leg.

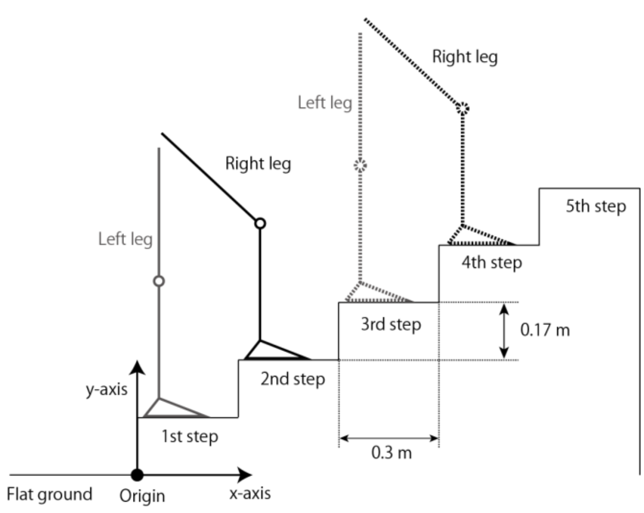

Figure 3. Stairs and global coordinate system used in study.

The angular and linear motion equations to simulate the prosthetic leg motion of the shank part are given by (1) and (2), respectively:

$$I_{OS}\ddot{\theta}_S = (-\boldsymbol{r_S}) \times \boldsymbol{F_K} \tag{1}$$

$$\boldsymbol{F_K} + m_S\boldsymbol{g} = m_S\boldsymbol{a_S} \tag{2}$$

where I_{OS} is the moment of inertia of the prosthesis (shank part) about the center of mass, $\ddot{\theta}_S$ is the angular acceleration of the shank part, $\boldsymbol{r_S} = r_S[\sin\theta_S, -\cos\theta_S]^{\mathrm{T}}$ is the relative position from the knee joint to the center of mass of the shank part, $\boldsymbol{F_K} = [F_{Kx}, F_{Ky}]^{\mathrm{T}}$ is the force acting on the shank part from the knee joint, m_S is the mass of the shank part, $\boldsymbol{g} = [0, g]^{\mathrm{T}}$ is the acceleration of gravity, and $\boldsymbol{a_S} = [a_{Sx}, a_{Sy}]^{\mathrm{T}}$ is the linear acceleration of the center of mass of the shank part. $\boldsymbol{a_S}$ can be written as shown in (3):

$$\boldsymbol{a_S} = \boldsymbol{a_K} + \dot{\boldsymbol{\omega}}_{\boldsymbol{S}} \times \boldsymbol{r_S} + \boldsymbol{\omega_S} \times (\boldsymbol{\omega_S} \times \boldsymbol{r_S}) \tag{3}$$

where $\boldsymbol{a_K} = [a_{Kx}, a_{Ky}]^{\mathrm{T}}$ is the linear acceleration of the knee, $\dot{\boldsymbol{\omega}}_{\boldsymbol{S}}$ is the angular acceleration vector of the shank part, and $\boldsymbol{\omega_S}$ is the angular velocity vector of the shank part. Consequently, (1) can be written as (4) by substituting (2) and (3).

$$(I_{OS} + m_S r_S^2)\ddot{\theta}_S = m_S r_S(g\sin\theta_S - a_{Kx}\cos\theta_S - a_{Ky}\sin\theta_S) \tag{4}$$

The translational motion parameters of the knee joint, including the position and acceleration ($\boldsymbol{a_K}$), were obtained from the experimental data of the able-bodied subjects. Sixth-order polynomials were fitted to the acceleration data and used to solve the differential equation of the prosthesis motion (4), which was modeled as a rigid body pendulum whose rotation axis was the translationally moving knee joint. Other parameters that determined the initial state were also obtained from the experimental data. The variables for the prosthesis properties were the (i) mass (m_S), (ii) moment of inertia around the axis passing through the center of mass (I_{OS}), and (iii) distance from the knee joint to the center of mass (r_S). The ranges of these parameters were as follows: (i) 0.5–3.0 kg in 0.1-kg increments, (ii) 0.01–0.30 kg m^2 in 0.01 kg m^2 increments, and (iii) 0.05–0.40 m in 0.05-m increments. If the toe did not contact any part of the stairs after taking off from the second step until it passed over the edge of the fourth step, the trial was regarded as a success. All combinations of inertial properties were tested under the FFC and HFC conditions.

After success or failure was determined, the joint reaction force $\boldsymbol{F_H} = [F_{Hx}, F_{Hy}]^{\mathrm{T}}$, which is simply related to the segment motion, and joint moment M_H acting on the hip joint were calculated for the successful trials. The values were calculated using the equations of thigh motion given below. The angular and linear motion equations of the thigh are given by (5) and (6), respectively (Figure 4):

$$I_{OT}\ddot{\theta}_T = \boldsymbol{r_{T2}} \times (-\boldsymbol{F_K}) + (-\boldsymbol{r_{T1}}) \times \boldsymbol{F_H} + M_H \tag{5}$$

$$\boldsymbol{F_H} - \boldsymbol{F_K} = m_T(\boldsymbol{a_T} - \boldsymbol{g}) \tag{6}$$

where I_{OT} is the moment of inertia of the thigh about the center of mass, $\ddot{\theta}_T$ is the angular acceleration of the thigh, $\boldsymbol{r_{T1}} = r_{T1}[\sin\theta_T, -\cos\theta_T]$ is the relative position from the hip joint to the center of mass of the thigh, $\boldsymbol{r_{T2}} =$

$r_{T2}[\sin\theta_T, -\cos\theta_T]$ is relative position from the center of mass of the thigh to the knee joint, m_T is the mass of the thigh, and $\boldsymbol{a_T} = [a_{Tx}, a_{Ty}]^{\mathrm{T}}$ is the linear acceleration of the center of mass of the thigh. $\boldsymbol{a_T}$ can be written as given below in (7):

$$\boldsymbol{a_T} = \boldsymbol{a_H} + \dot{\boldsymbol{\omega}}_{\boldsymbol{T}} \times \boldsymbol{r_{T1}} + \boldsymbol{\omega_T} \times (\boldsymbol{\omega_T} \times \boldsymbol{r_{T1}}) \qquad (7)$$

where $\boldsymbol{a_H} = [a_{Hx}, a_{Hy}]^{\mathrm{T}}$ is the linear acceleration of the knee, $\dot{\boldsymbol{\omega}}_{\boldsymbol{T}}$ is the angular acceleration vector of the thigh, and $\boldsymbol{\omega_T}$ is the angular velocity vector of the thigh. Consequently, (5) is written as (8) by substituting (6) and (7):

$$\begin{aligned}&(I_{OT} + m_T r_{T1}^2)\ddot{\theta}_T \\ &= m_S r_T\{g\sin\theta_T - a_{Kx}\cos\theta_T - a_{Ky}\sin\theta_T - \ddot{\theta}_S r_S \\ &\quad + \dot{\theta}_S^2 r_S(\sin\theta_S\cos\theta_T - \cos\theta_S\sin\theta_T)\} \\ &\quad + m_T r_{T1}\{g\sin\theta_T - a_{Hx}\cos\theta_T - a_{Hy}\sin\theta_T\} + M_H\end{aligned} \qquad (8)$$

where $r_T = |\boldsymbol{r_{T1}}| + |\boldsymbol{r_{T2}}|$ is the length of the thigh. To evaluate the effects of the inertial properties on the thigh kinetics, the joint moment at the hip joint (M_H) and joint reaction force at the hip joint ($\boldsymbol{F_H}$) were calculated.

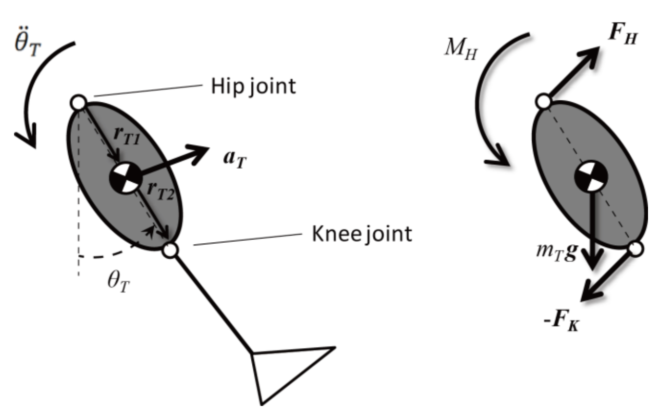

Figure 4. Free body diagram of the thigh of the prosthetic side.

III. SIMULATION RESULTS AND DISCUSSION

All of the trials under the FFC condition failed; none cleared the third step. On the other hand, some inertial properties under the HFC condition allowed the prosthesis to avoid collision with the stairs. The dots depicted in Figure 5 indicate the inertial properties of the prosthesis that resulted in a successful leg swing under the HFC condition in the computer simulation; the dots are spread out in a strip in each graph.

The present simulation revealed the specific inertial properties in the successful computer simulation trials, as shown in Figure 5. These results indicate that a transfemoral prosthesis with such inertial properties should be able to swing a leg with a thigh motion similar to that of able-bodied subjects under the HFC condition. In addition, if the prosthesis has inertial properties around the dots in the graphs, slight adjustments to the thigh motion could make the leg swing successful.

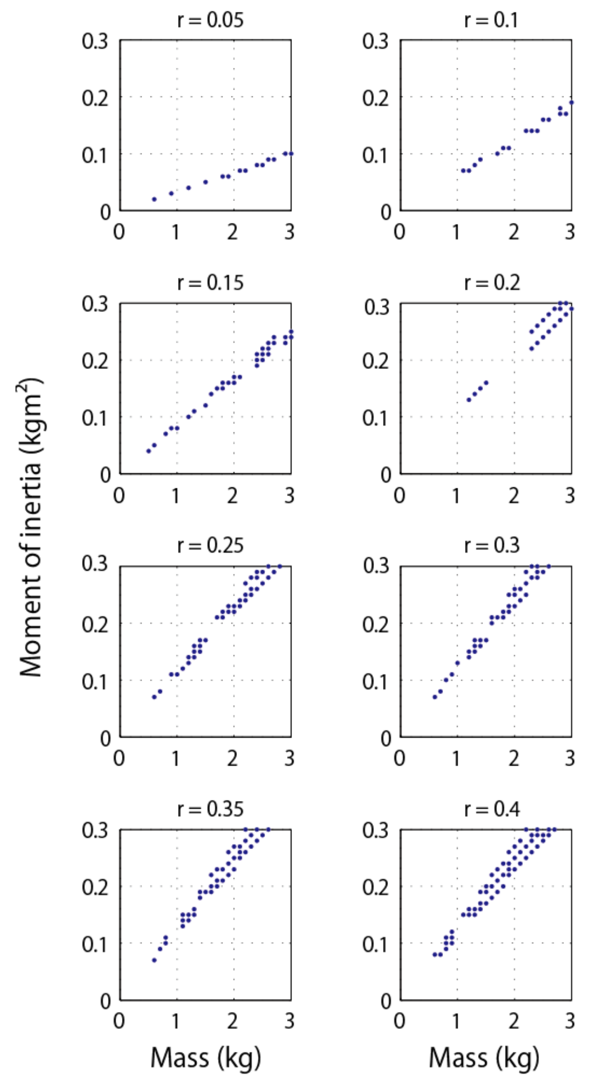

Figure 5. Inertial properties of the prosthesis. *r* is the distance from the knee joint to the center of mass of the prosthesis. The dots indicate the inertial properties that allowed a successful leg swing under the HFC condition in the computer simulation (data derived from Inoue et al., 2012).

Figure 6 shows the joint moment at the hip joint for simulated and able-bodied leg motions. All simulated trials with a successful leg swing that did not collide with the steps are superimposed. The peak values of flexion and extension moments in the simulation trials were lower than those for able-bodied subjects. The time to the peak extension moment was earlier in the simulation trials. The joint reaction force is shown in Figure 7. The time to the peak value of the horizontal (x-axis) component was earlier in the simulation trials than for the able-bodied subjects. The peak horizontal and vertical (y-axis) components did not differ greatly between the simulated trials and the able-bodied subjects, even though the simulated trials had very light mass properties. The kinetics of the simulated trials suggest that a transfemoral prosthesis leg swing can be achieved with a similar or smaller kinetic demand at the hip joint when a half-foot contact strategy is used for foot placement on stair steps.

The variability in the simulated joint reaction force and joint moment at the hip joint was mainly caused by the mass size of the prosthetic part. A possible explanation is given below. Equation (4), which determines the prosthetic leg motion, can be written as (9).

$$\ddot{\theta}_S = \frac{m_S r_S}{I_{OS} + m_S r_S^2}\left(g \sin\theta_S - a_{Kx}\cos\theta_S - a_{Ky}\sin\theta_S\right) \tag{9}$$

For the inertial parameters simulated in the present study, the mass (m_S) was found to have the greatest effect on the prosthetic leg angular acceleration among all the inertial properties. For (8), which is the motion equation of the thigh, the following terms were included: mass of the prosthetic part (m_S), angular acceleration ($\ddot{\theta}_S$), and angular velocity ($\dot{\theta}_S$). These parameters were determined by the inertial properties of the prosthesis. It means that the hip kinetics was also affected by the prosthesis inertial properties. However, few previous studies referred to inertial properties [6, 7]. Therefore, these results can enhance transfemoral prosthesis design.

IV. Conclusion

The effects of the inertial properties of a transfemoral prosthesis on thigh kinetics were evaluated through stair ascent simulations of certain motion strategies. The joint moment at the hip became smaller than that of able-bodied subjects. The peak values of the horizontal and vertical components of the joint reaction force were approximately the same as those of able-bodied subjects. These results suggest that a transfemoral prosthesis leg swing can be achieved with similar or smaller kinetic demand at the hip joint relative to able-bodied subjects when half-foot contact is used as a stair ascent strategy. The mass was found to have the largest effect among all the inertial properties on the variability of the simulated kinetic parameters. The results of the present study may enhance prosthesis design with regard to inertial properties and usability.

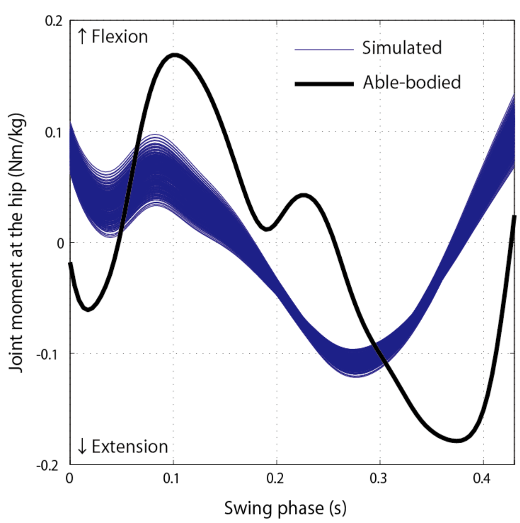

Figure 6. Joint moment at the hip joint during the swing phase. All successful simulation trials are superimposed.

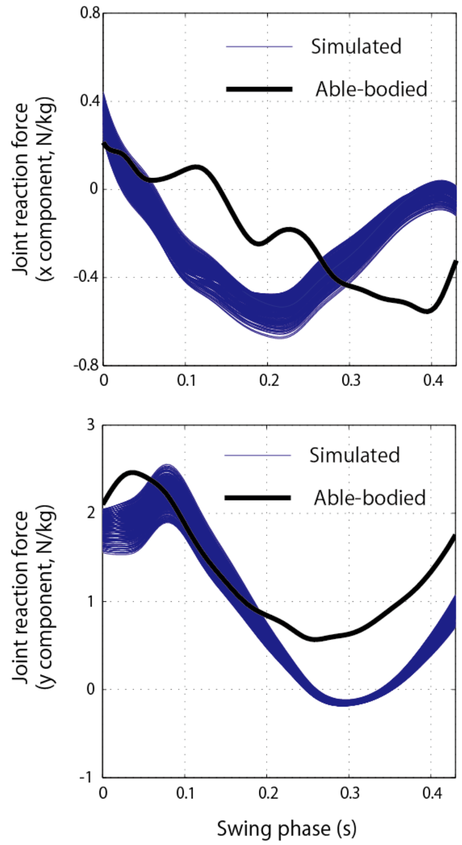

Figure 7. Joint reaction force at the hip. All successful simulation trials are superimposed.

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