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Nakagawa, Akio
Furukawa, Hiroshi
Matsubara, Hiroyuki
Chin, Takaaki

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Influence of Walking Speed Change and the Swing Phase Adjustment of the Intelligent Prosthesis Users on the Intact Limb

*Akio Nakagawa, **Hiroshi Furukawa, ***Hiroyuki Matsubara, ****Takaaki Chin

The aim of this study was to investigate the function of the intact limb of the trans-femoral prosthesis users in terms of joint movements, moments and powers in the saggital plane when they change walking speeds and when they change the swing phase control of the knee joint of their prosthesis. Five trans-femoral amputees and ten able-bodied subjects walked at 40 meters per minutes(m/min), 60 m/min, 80 m/min, 90 m/min (amputees only, maximum speed for the amputees,) and 100 m/min (able-bodied subjects only) along a walkway. All the amputees used the Intelligent Knee Joint. They had finished sufficient gait training to use this type of the prosthesis. They walked with their Intelligent function active (IPOn) and inactive (IPOff). When the Intelligent Knee with IPOff was applied, it is the same as the amputees using the conventional prosthetic knee, with a pneumatic swing phase control cylinder. As a result, joint moments and powers increased according to the increase of speed which was similar to those of able-bodied subjects. Nevertheless, the knee joint moment of the intact limb was larger than the able-bodied subjects, but there was no difference in the intact limb function between IPOn and OPOff.

Key words
Intelligent Prosthesis, Speed change, Joint moment, Joint power

Introduction

The Intelligent trans-femoral prosthesis (IP) incorporates a computer controlled pneumatic swing phase control cylinder. The author and his group developed this system. The knee joint became commercially available, as the first computer controlled prosthetic knee through two companies from Japan and UK. The computer system automatically adjusts the valve opening to fit with the walking speeds[1]. The cylinder is attached behind the knee joint as shown in Fig.1. The basic mechanism of the cylinder is drawn in Fig.2. When the user walks fast, the needle valve moves rightward and makes the valve opening smaller, causing the cylinder to be highly compressed, further generating an air spring like function. This results in a fast flexion and extension of the below knee part of the prosthesis during the swing phase. When the user walks slowly, the needle valve moves leftward and makes the valve opening wider. Then, the cylinder generates low resistance to make the below knee part swing slow. With this mecha-

* Faculty of Rehabilitation, Kobe Gakuin University
** Faculty of Health Sciences, Kobe University School of Medicine
*** Hyogo Assistive Technology Research and Design Institute, Hyogo Rehabilitation Centre
**** Department of Orthopedics, Hyogo Rehabilitation Centre

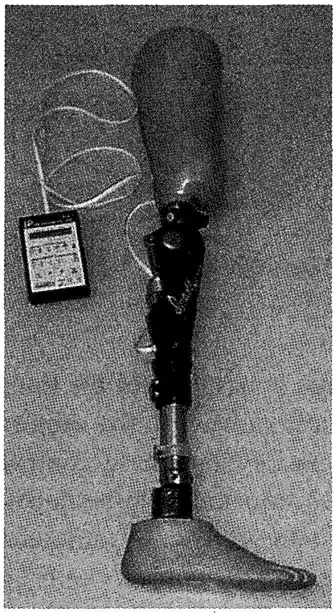

Fig.1 Intelligent Prosthesis with the Valve Opening Adjustment Unit

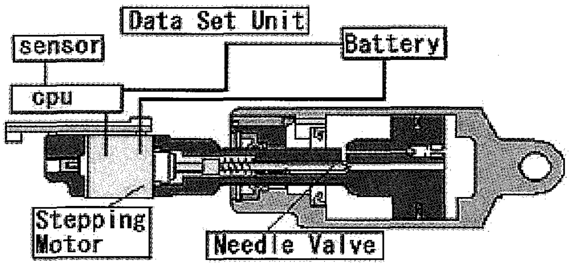

Fig.2 Construction of the Computer controlled swing phase control cylinder

nism, the user of the prosthesis can change walking speed freely. The gait pattern functions well for a wide range of walking speeds.

IP has been widely accepted and used by many amputees. An advantage of using this prosthesis is that the users feel less tired in walking than with the conventional prostheses[2]. There have been some studies to investigate the influence of the introduction of IP [3] and tried to find out a way to measure the energy consumption[4)5)6)]. There have been some researches to confirm this by measuring the energy consumption[7)8)9)10)]. These researches concluded that the energy consumption is about 10% less than with the conventional prostheses at normal walking speed[10]. Most of them used the metabolic measurement system to gauge the energy consumption. It can measure the energy consumption of the total body, but it doesn't give any information about what and where the difference is.

Most of the users of IP could acquire the ability to change their walking speed[11], this is another advantage of this prosthesis. The users of IP can change their walking speed within a narrow range, when they could not receive sufficient gait training to use the Intelligent knee. However, they can change in a wide range of walking speeds when they receive appropriate training[12)13)].

A small number of researches referred to function of the intact limb of the trans-femoral amputees when they used prostheses, and reported that the amputees compensated the functional loss of the amputated leg, by generating more moment and power at the intact limb joints. The research found the peak dorsiflexion moment was large, whilst extension moments at the knee and hip and the peak power at the hip were also large[14]. The amputees walked at a constant speed (1.2m/sec) in the experiment. There have been no studies done to investigate the function of the intact limb, when the amputee changed walking speed. Further there have been no studies about the influence of the adjustment of the prosthetic knee on the function of the intact limb.

The purposes of this study were to analyze the gait with the Intelligent prosthesis at variety of walking speeds, plus to know the function of the intact limb by comparing with the gait of the able-bodied subjects. Another objective is to know the influence of the swing phase control of the prosthesis on the intact limb.

Materials and Methods

Subjects

The amputee subjects were 5 unilateral trans-femoral amputees (all male) aged between 21 and 54 with average age of 36.5 who had received the IP walking training. The amputees had been well trained in the use of the IP and were skilled in its use. The prostheses used for the experiment were all used by each subject in their daily life. All subjects wore their own footwear. The physical characteristics of the subjects are shown in Table 1. The able-bodied subjects (n=10), mean age 34.9 years (SD=12.3), did not report any lower limb injury or history of injury at the time of testing. All the subjects received an explanation of the objective of this research and understood clearly, with all of them agreeing to participate as the subject.

Experiment and data collection

Subjects were instructed to walk along a walkway equipped with the two Kistler Force Platforms (typeZ13216, width:600mm, length:1200mm) connected to 4 camera Elite-Plus gait analysis system. These two force platforms were placed parallel as seen in Fig.3. One is for the left leg and the other is for the right leg. Data was sampled at 50Hz from both sides simultaneously. To regulate the walking speeds, the subjects were instructed to follow the staff who used the Walking Measure (Toei Light G-1015) at 40m/min, 60m/min, 80m/min and 100m/min. In case of the amputee subjects, their maximum speed was 90m/min. The use of the Walking Measure is seen in Fig.4.

To measure the 3D coordinates of the subject body, markers were attached at the shoulders, hips, knees, ankles and 5th metatarsal joints on both sides based on DIFF recommendation[15]. This is also seen in Fig.3.

Table 1. Physical characteristics of the amputee subjects

Subject	A	B	C	D	E
Sex	M	M	M	M	M
Age(yr)	54	33	35	21	40
Body Weight(kg)	57	53	65	58	52
Height(cm)	173	163	173	173	171
Cause of amputation	Trauma	Trauma	Trauma	Trauma	Trauma

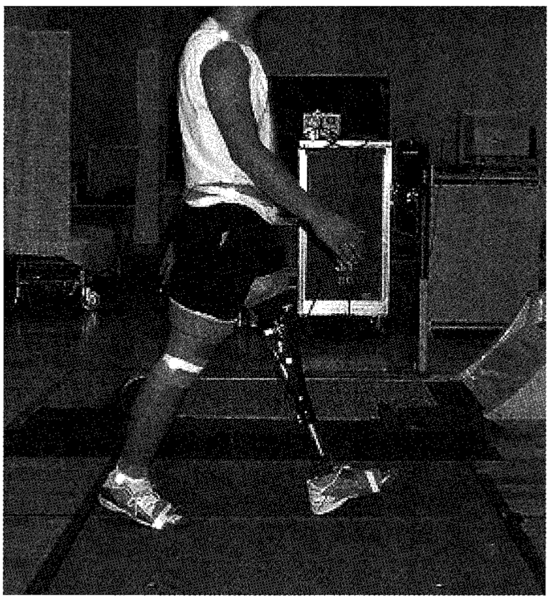

Fig.3 Gait Analysis, Force plates and Marker set

Fig.4 Walking Measure (Speed measuring system)

To minimize the variations of the data, each subject walked along the walkway more than 4 times at the same walking speed to collect a complete set of data from a heel strike to the next heel strike of the same leg. Especially for amputees, data of interest is of the intact leg, they had to undergo extra walking until appropriate data could be accumulated.

The amputee subjects walked with the Intelligent function active (IPOn) and inactive (IPOff). To change from IPOn to IPOff, it was easily made by changing the control data to set to the constant values, for all walking speeds. At this time, the subjects tried to walk until they became accustomed to the new settings.

DATA ANALYSIS

DIFFGait and WAVE_EYES programs were used for the kinematic and kinetic data analysis. The Clinical Gait Analysis Forum of Japan developed these programs to calculate the floor reaction force, joint angular movement, joint moment and joint power in a sagittal plane. Joint power is defined as a product of joint moment and angular velocity. This expresses the work done in a unit time. The gait analysis data was accumulated by Elite-Plus system, then the data was converted to fit with the DIFF format. It was later processed by the DIFFGait program. This program deals with the low-pass filtering, joint angle, joint center, center of gravity, joint moment and joint power. The joint angles measured when the subject stood still were calculated as the zero angles of each joint. WAVE_EYES gives the graphical expression of these records as well as the normalization to time. As a result, we can get a one-cycle data in the percentage expression. This data was used to calculate the

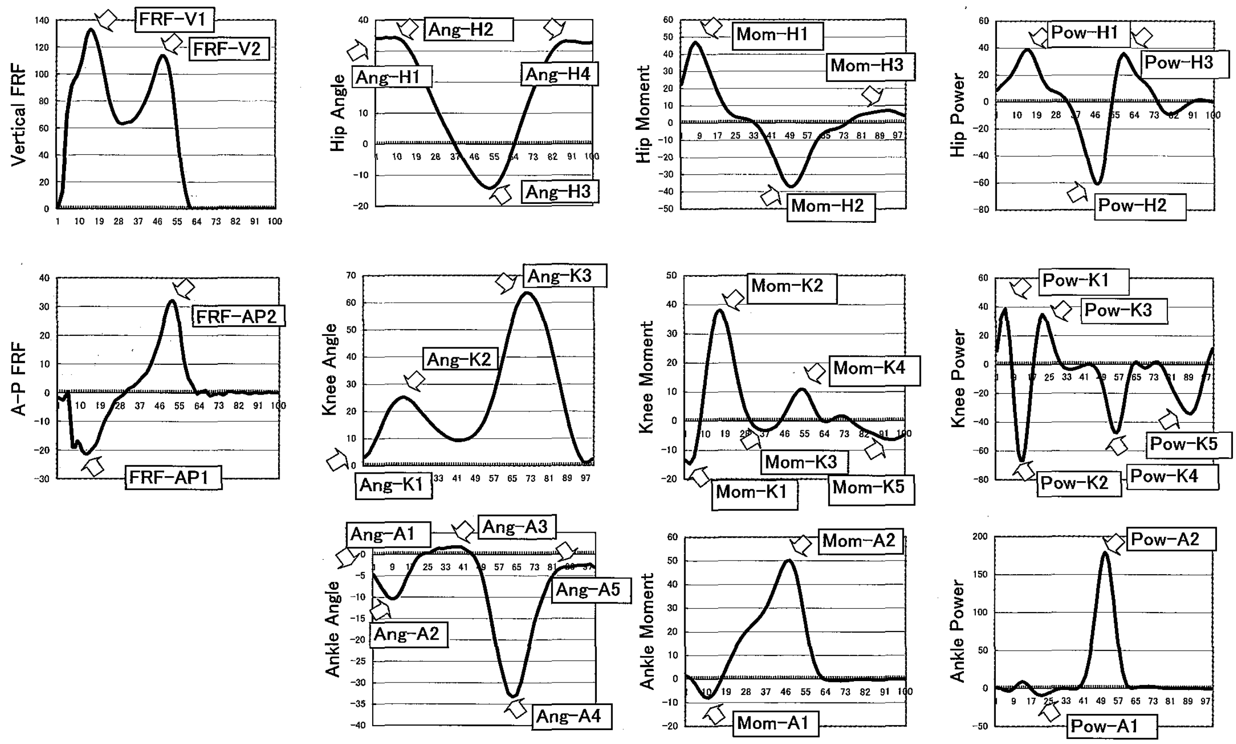

Fig.5 Parameter positions and abbreviations
FRF: Floor Reaction Force
V: vertical AP: Antero-Posterior
H,K,A: Hip, Knee, Ankle
Ang,Mom,Pow: angle, moment, power

average values and to compare with the other conditions, or other subjects at the same walking speeds. The average values were calculated from 4 gait data. In case when the sampling was incomplete, three or two gait trials were used for the calculation. Fig.5 shows the 36 parameters (specific values) for the comparison.

To test whether there is a statistical difference between corresponding parameters at the same walking speed, within able-bodied and IPOn subjects, F-test was applied to evaluate the equality of variances in two data sets, then appropriate t-tests were applied. For the comparison of IPOn and IPOff parameters, the paired t-test was applied. The adopted significance level was 5%.

Results

Gait cycle time and Stance Phase Percentage in the gait cycle

In Table 2, it shows the results of the gait cycle time, stance percentage and swing percentage. There were no differences between the gait cycle time, within the able-bod-

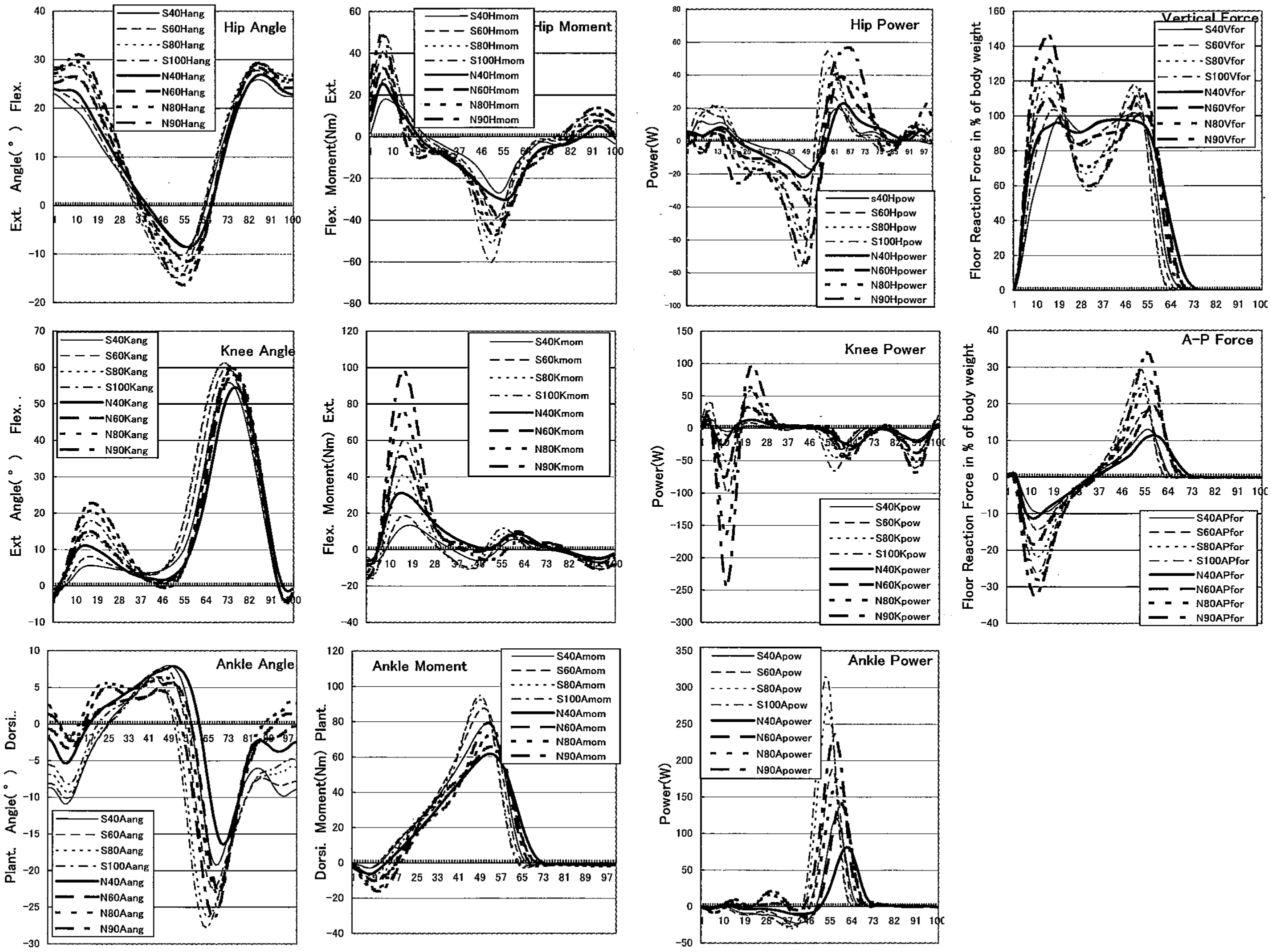

Fig.6 Comparisons between Able-bodied and Amputees(IPOn)
S: Able-bodied N: IPOn
40,60,80,90,100: Speed(m/min)
H, A, K: Hip, Ankle, Knee
ang, mom, pow: angle, moment, power
Vfor, APfor: vertical force and A-P force
of floor reaction force

Table 2 Comparison of One cycle time and Stance, Swing Phase rate

	One cycle time (sec)	Stance Phase (%)	Swing Phase (%)
Able-bodied 40m/s	1.22	67.8	32.2
IPOn 40m/s	1.2	72.1	27.9
IPOff 40m/s	1.21	71.9	28.1
Able-bodied 60m/s	1.02	65.9	34.1
IPOn 60m/s	1.01	69.9	30.1
IPOff 60m/s	0.97	70.0	30.0
Able-bodied 80m/s	0.88	64.4	35.6
IPOn 80m	0.89	68.8	31.2
IPOff 80m	0.86	67.3	32.7
Able-bodied 100m/s	0.84	63.9	36.1
IPOn 90m/s	0.83	67.6	32.4
IPOff 90m/s	0.82	66.7	33.3

ied, IPOn and IPOff subjects. The gait cycle times were about 1.2 seconds (40m/min), 1.0 second (60m/min), 0.88 seconds (80m/min), and 0.82 seconds (90-100m/min). Stance phase percentage of the intact leg of amputees in the gait cycle was 4% longer than able-bodied subjects. Stance phase percentages were 68% : 72% (able-bodied : amputees, 40m/min), 66% : 70% (60m/min), 64% : 68% (80m/min) and 64% : 67% (100-90m/min).

Observation of the averaged wave patterns of the able-bodied and IPOn subjects

In Fig. 6, it shows all the wave patterns for the joint angles, joint moments, joint powers and floor reaction forces for the able-bodied and IPOn subjects at 40, 60, 80 and 90-100 m/min. When the able-bodied subjects walked faster, the joint angle values did not change significantly but the peak moment and power values became larger accordingly. In the graphs of the Knee Angle, the timings of the peak flexion in the swing phase became earlier when the subject walked faster. The Ankle Angles showed a similar tendency, in that the timings of the peak planter-flexion became earlier. In case of the IPOn subjects, the peak moment and power values became larger ac-

Table 3 Parameter comparison between Able-bodied and IPOn subjects

Velocity	FRF-V1	FRF-V2	FRF-AP1	FRF-AP2	Ang-H1	Ang-H2	Ang-H3	Ang-H4	Ang-K1	Ang-K2	Ang-K3	Ang-A1
40			SD									
60												
80	SD		SD									
90(100)	SD		SD									

Velocity	Ang-A2	Ang-A3	Ang-A4	Ang-A5	Mom-H1	Mom-H2	Mom-H3	Mom-K1	Mom-K2	Mom-K3	Mom-K4	Mom-K5
40							SD					
60	SD								SD			
80	SD							SD	SD			
90(100)	SD								SD			

Velocity	Mom-A1	Mom-A2	Pow-H1	Pow-H2	Pow-H3	Pow-K1	Pow-K2	Pow-K3	Pow-K4	Pow-K5	Pow-A1	Pow-A2
40		SD					SD	SD				
60		SD	SD				SD	SD				
80							SD	SD				
90(100)							SD	SD				

SD: Significantly Different

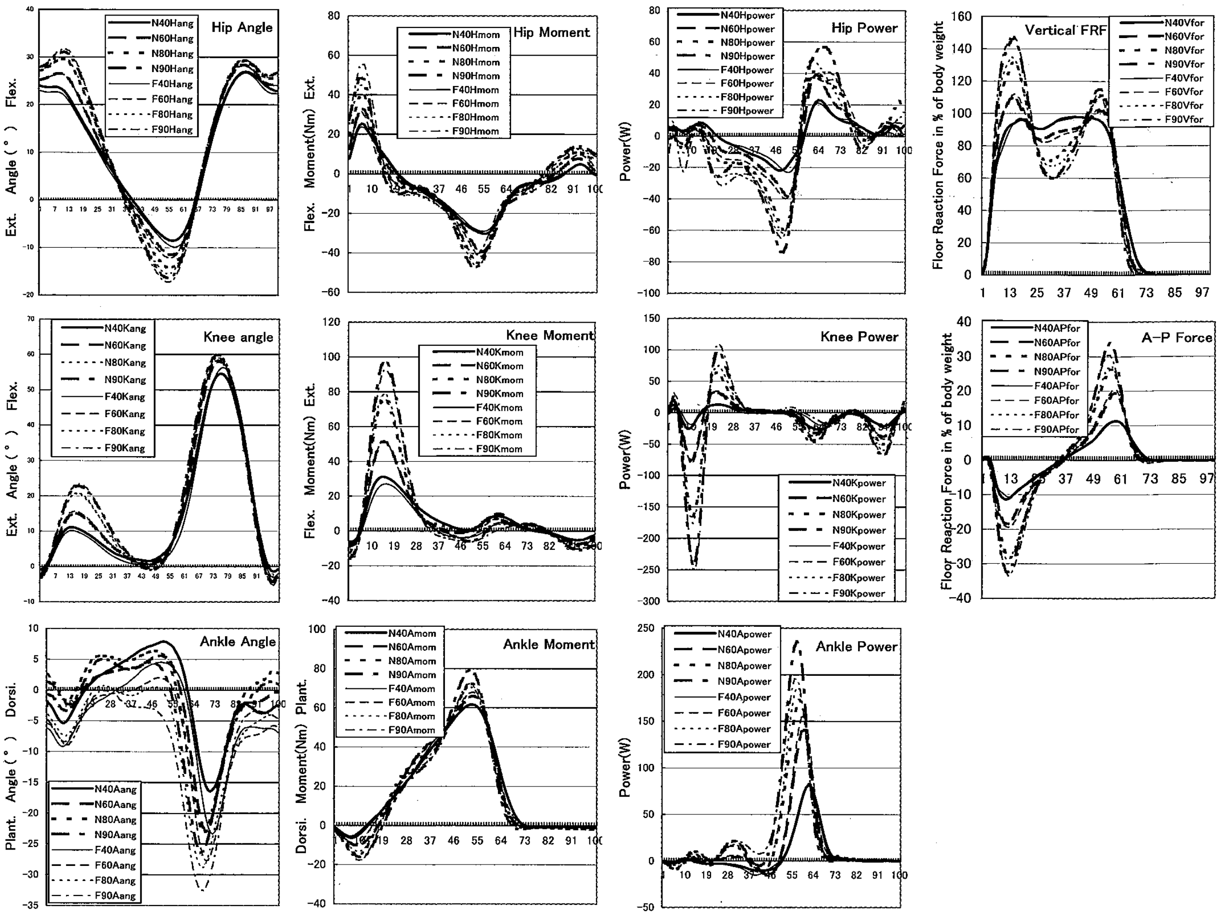

Fig.7 Comparisons between IPOn and IPOff
N: IPOn F: IPOff 40,60,80,90 : Speed(m/min) H, A, K: Hip, Ankle, Knee
ang, mom, pow: angle, moment, power Vfor, APfor: vertical force and A-P force

Table 4 Parameter comparison between IPOn and IPOff subjects

Velocity	FRF-V1	FRF-V2	FRF-AP1	FRF-AP2	Ang-H1	Ang-H2	Ang-H3	Ang-H4	Ang-K1	Ang-K2	Ang-K3	Ang-A1
40												
60												
80												
90												
Velocity	Ang-A2	Ang-A3	Ang-A4	Ang-A5	Mom-H1	Mom-H2	Mom-H3	Mom-K1	Mom-K2	Mom-K3	Mom-K4	Mom-K5
40												
60												
80												
90												
Velocity	Mom-A1	Mom-A2	Pow-H1	Pow-H2	Pow-H3	Pow-K1	Pow-K2	Pow-K3	Pow-K4	Pow-K5	Pow-A1	Pow-A2
40												
60			SD									SD
80										SD		
90		SD										

SD: Significantly Different

cording to the walking speeds, but the timings of the peak values did not change so much as those in able-bodied cases.

In Table 3, it shows the comparisons of the average values of the parameters between able-bodied and IPOn subjects. SD means that the values have a significant difference. In the data of the Floor Reaction Forces, IPOn subjects generated a significantly greater ($p<0.05$) value at the first vertical force peak and the first A-P force peak. They also generated a significantly greater ($p<0.05$) value at the knee extension moment at the early stage of the

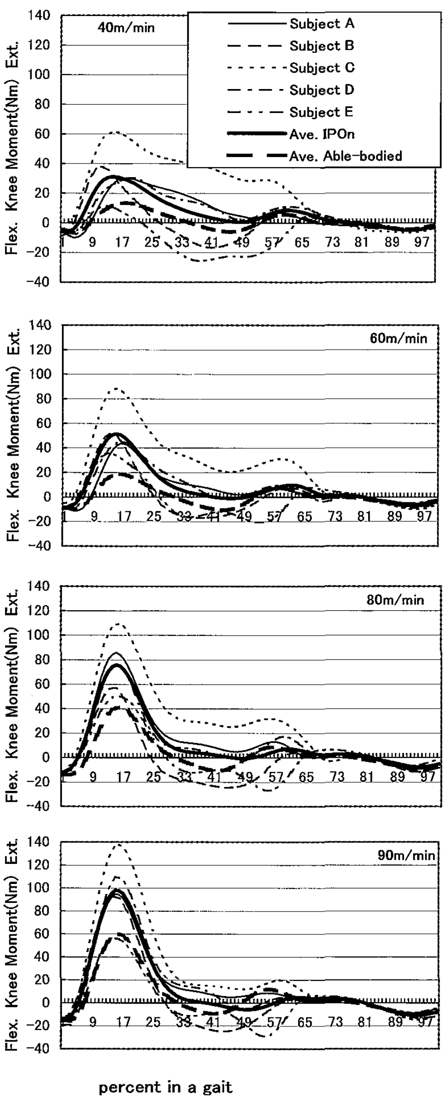

Fig.8 Knee Moment of Amputees and Averaged Able-bodied subjects

stance phase, and ankle planter flexion moment was small at the end of the stance phase. In the graphs of the joint power, IPOn subjects generated a significantly greater ($p<0.05$) value at the first minus knee power and the following plus knee power.

Observation of the averaged wave patterns of IPOn and IPOff subjects

In Fig. 7, it shows the averages of the amputees with IPOn and IPOff at a variety of walking speeds. The wave patterns are similar in most cases. The moment, power and floor reaction force values became larger according to the change of walking speed. In most cases, it looks that the wave forms of IPOn and IPOff are almost the same.

In Table 4, it shows the comparisons of the average values of the parameters between IPOn and IPOff subjects. There were only 4 cases of which reported differences, besides that no significant differences were evident. Even when there was a significant difference, they did not have any effect on the change of the walking speed. They might be caused by the variability of the subjects and the number of samples were small.

Inter-subject data variability

In Fig. 8, it shows examples of variability between subjects. The graphs show the knee moment at each walking speed as an example. At 40m/min, inter-subject variability was substantial. When walking speed went up, wave patterns became similar though the values were different.

Discussion

In the case of IPOn subjects, the first peak value of vertical floor reaction force was larger than that of the able-bodied subjects. It is the timing of the knee flexion at the end of the stance phase of the opposite prosthetic knee. The IP prosthesis does not have any mechanism to resist the knee flexion when the knee flexes large in angle. It is suggested that the intact leg supports the body weight and controls the prosthetic knee flexion speed. This results in the A-P component of the floor reaction force to be larger.

The knee extension moment at the beginning of the stance phase was larger than that of the able-bodied subjects. This is related to the larger knee flexion during the stance phase. Larger knee minus power (absorption) followed by larger knee plus power (generation) could explain this. The knee flexion just after the start of the stance phase resulted in the flexion of the hip. This could be proved because the hip moment at this period was the only extension moment. The hip flexion after the start of the stance phase was not intentional. This knee flexion is used for the shock absorbing function also observed in gait of the able-bodied subjects. In the IP prosthesis, it does not have this function. This is the method of compensation used by the amputees[13]. This may results in keeping the vertical movement of the center of the gravity less.

The timing of the maximum knee flexion during the swing phase of the amputee subjects did not change according to the walking speeds. It was late comparing to those of the able-bodied subjects. This means that the amputees kept the stance phase of the intact limb as long as possible. This is also proved by the long duration of the stance phase period by 4% longer for the amputee subjects.

At the ankle, the angles of the IPOn were different from those of the able-bodied subjects but the ranges of motion of the ankle were almost the same in these two groups[14]. The angles at the beginning of the stance phase became dorsi-flexed according to an increase of the walking speeds. Ankle moment was larger at 40 and 60 m/min for the amputee subjects. This is to generate the propellant force by the intact limb. Nevertheless, the ankle moment was not larger than the able-bodied gait at 80 and 90-100 m/min. This does not mean that the ankle of the intact limb did not generate the large moment to generate the propellant force, but the able-bodied subjects also generated the ankle moment to walk fast.

More parameters were expected to have statistically different but small number of them showed the differences. To think about the intact leg of the amputee subjects, the prosthesis users required small assistance or compensation to keep walking even at the fast walking speeds.

From the comparison of IPOn and IPOff data, there were no differences from the statistic calculations, or that they were similar in almost all of the data. Though the swing phase control affects much of the outlook of the gait, the intact leg keeps the same way to walk. It is true that the swing phase control affects the step length of the prosthetic limb and the timing of the prosthetic swing. It may affect the components in the other direction of the intact limb, for instance, in the frontal plane or rotation of the limb. This experiment was done for IPOn and IPOff in a short time, the data may differ if they walk with IPOn or IPOff for a long time.

Conclusion

The comparison of gait analysis data of the able-bodied and trans-femoral prosthesis users at the variety of walking speeds showed little difference in the saggital plane function of the intact limb. Those of the IPOn and IPOff showed almost no difference. The swing phase control did not affect the saggital plane function of the intact limb. Further work is needed to investigate what is the cause of difference of energy consumption between IPOn and IPOff gait.

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