Novel Knee Joint Mechanism of Transfemoral Prosthesis for Stair Ascent

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***Abstract*—The stability of a transfemoral prosthesis when walking on flat ground has been established by recent advances in knee joint mechanisms and their control methods. It is, however, difficult for users of a transfemoral prosthesis to ascend stairs. This difficulty is mainly due to insufficient generation of extension moment around the knee joint of the prosthesis to lift the body to the next step on the staircase and prevent any unexpected flexion of the knee joint in the stance phase. Only a prosthesis with an actuator has facilitated stair ascent using a step-over-step gait (1 foot is placed per step). However, its use has issues associated with the durability, cost, maintenance, and usage environment. Therefore, the purpose of this research is to develop a novel knee joint mechanism for a prosthesis that generates an extension moment around the knee joint in the stance phase during stair ascent, without the use of any actuators. The proposed mechanism is based on the knowledge that the ground reaction force increases during the stance phase when the knee flexion occurs. Stair ascent experiments with the prosthesis showed that the proposed prosthesis can realize stair ascent without any undesirable knee flexion. In addition, the prosthesis is able to generate a positive knee joint moment power in the stance phase even without any power source.**

Keywords—Transfemoral prosthesis; Knee joint; Stair ascent

I. INTRODUCTION

A transfemoral prosthesis is a prosthetic leg that is used by individuals who have been amputated above the knee. In such a prosthesis, the mechanism and control method are important to guarantee safe walking. Recent advancements in the mechanism and control method for transfemoral prostheses have drastically improved the gait of amputees and realized a safer stance phase for the prostheses [1]. In particular, computer-controlled transfemoral prostheses has significantly contributed to considerably increase safety when walking with a prosthesis on level ground, as well as improve the smoothness of the swing phase [2], [3], [4]. In addition, the smoothness of prosthetic walking was quantified based on the effective use of the inertia property of the prosthesis on flat ground [5] [6], and it was applied to evaluate walking skill [7].

It is known that prosthesis users find it difficult to ascend stairs. Thus, prosthesis users tend to avoid using stairs in their daily lives. This difficulty in ascending stairs is mainly due to the generation of a insufficient extension moment around the knee joint of the prosthesis to lift the body to the next step on the staircase and prevent any unexpected flexion of the knee joint in the stance phase. Hobara et al. first demonstrated that a transfemoral prosthesis without an actuator and with very low friction allowed amputees to ascend stairs by developing a special gait through rehabilitation training [8]. Although this was an important finding, it involved users being trained to use the hip extensor and flexor for compensatory motion, which could lead to fatigue. In fact, it was shown that a knee extension moment is required in the stance phase of stair ascent by healthy individuals [9]. Thus, the knee extension moment is required for stair ascent with a natural gait using a prosthesis.

One technical solution is to use an actuator in the knee joint. Powered knee joints that use large motors to produce the knee extension moment were developed and realized a stable gait [10][11]. Their use is, however, associated with issues involving durability, cost, and maintenance. As an alternative solution, a prosthesis that does not use an actuator but just locks the knee flexion in the stance phase has been developed [12][13][14]. These prostheses can avoid the unexpected knee flexion in the stair ascent. Assistive devices such as handrails are indispensable for ascending stairs because amputees cannot generate the positive power around the knee joint that is required for knee extension during ascent.

Therefore, this paper proposes a novel passive mechanism for a prosthetic knee joint that prevents undesired knee flexion and generates knee extension moment in the stance phase, as a basic study for realizing more effective prosthesis design. First, the developed novel knee joint mechanism is introduced. Then, its fundamental characteristics are explained. Finally, the effectiveness of the developed knee joint mechanism is demonstrated based on stair ascent experiments with healthy individuals.

II. DEVELOPMENT OF NOVEL KNEE JOINT

A. Basic Functions

A novel knee joint with the following functions was designed for stair ascent to solve the problems in the stance phase.

1) Flexion-lock function

At the beginning of the stance phase, the ground reaction force is increased when the knee flexion exists [9]. Therefore, any unnecessary knee flexion should be avoided in the stance phase.

2) Extension function

In addition, a knee extension movement is generated in the stair ascent of healthy individuals [9]. Thus, the knee flexion moment is required in the stance phase to lift the body to the next step on the staircase.

B. Structure

A novel knee joint for a prosthesis with a link mechanism is proposed, as shown in Figs. 1, 2, and 3, for realizing the two functions mentioned above. Fig. 4 shows the link mechanism diagram. As shown in Fig. 4, knee joint J_k is represented by the relative rotation between block B and Link L_1. Block B is connected to block C through the 1-dof translational joint with a linear spring. This linear displacement/shortening of the joint from the original state without any ground reaction force is called the linear displacement *D*. Therefore, the positive direction of *D* is downward. L_1 and L_2 are connected to joint J_1 by the slide part A. The knee joint limiter can slide along L_2. The knee joint limiter is connected to block C through J_2 and J_4. The maximum knee flexion angle is determined by the collision of the sliding part A and the knee flexion limiter on L_2. The linear displacement *D* is increased by increasing the ground reaction force, leading to the sliding up of the knee flexion limiter along L_2 and the sliding down of sliding part A. This results in a decrease in the maximum knee flexion angle. The knee joint angle and linear displacement are defined as $\theta_k\,[\mathrm{deg}] \in [0,88]$ and $D[\mathrm{mm}] \in [0,20]$.

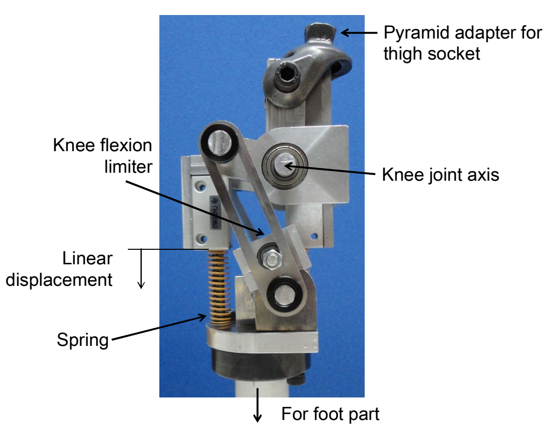


Fig. 1. Photo of developed knee joint mechanism

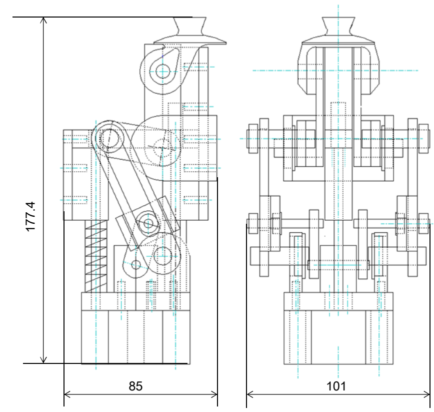


Fig. 2. Drawing of developed knee joint mechanism

Fig. 3. Photo of developed knee joint with foot part

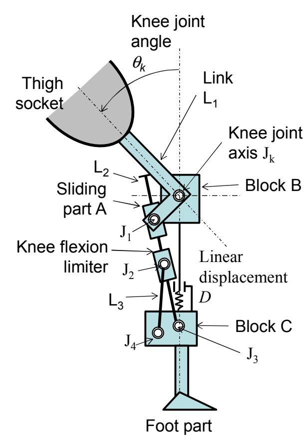
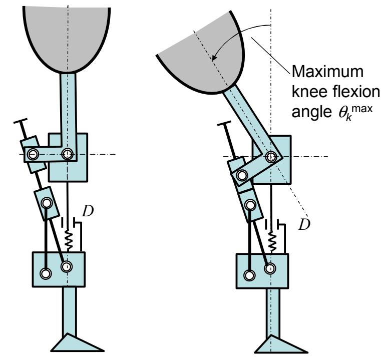


Fig. 4. Link mechanism diagram

C. Flexion-Lock Function

The linear displacement *D* is increased by increasing the ground reaction force. Then, the sliding part A is moved relatively downward along link L_2, and the resultant distance between the sliding part A and the knee flexion limiter is decreased, as shown in Fig. 5-(a). The knee joint angle θ_k can take a positive value, and it reaches its maximum value θ_k^{max} when the sliding part A contacts the knee flexion limiter, as shown in Fig. 5-(b). The maximum knee flexion angle θ_k^{max} is decreased when the linear displacement *D* is increased (see section III for the details).

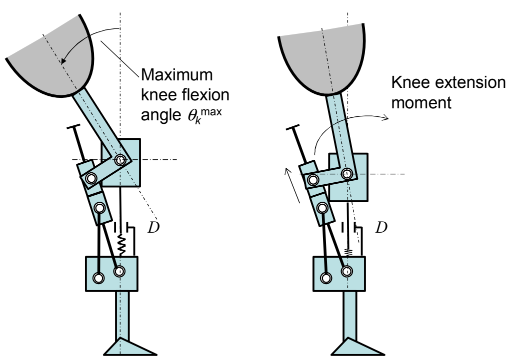


(a) free state (b) maximum knee flexion state

Fig. 5. Flexion-lock function

D. Extension Function

Suppose that the knee joint angle reaches its maximum value θ_k^{max} with a given linear displacement D as shown in Fig. 6-(a). In this situation, the sliding part A moves relatively upward along L_2 when the ground reaction force is increased or the linear displacement D is increased, as shown in Fig. 6-(b). This causes a decrease in the knee joint angle θ_k ($=\theta_k^{max}$). This means the knee extension motion is produced by increasing the ground reaction force, which helps in stair ascent.



(a) smaller D (b) larger D

Fig. 6. Extension function

III. KINEMATICS AND STATICS OF KNEE JOINT

A. Kinematics

A kinematics analysis shows that the maximum knee flexion angle θ_k^{max} is uniquely determined by the linear displacement D. Therefore, a description of θ_k^{max} (D) is used to represent this relationship. Because of space constraints, the details are omitted. However, the relationship between the linear displacement D and the maximum knee flexion angle θ_k^{max} is given by Fig. 7, which shows that θ_k^{max} decreases with increasing D. Note that this knee joint mechanism was designed so that θ_k^{max} = 88deg at D = 0 mm and θ_k^{max} = 30deg at D = 20 mm.

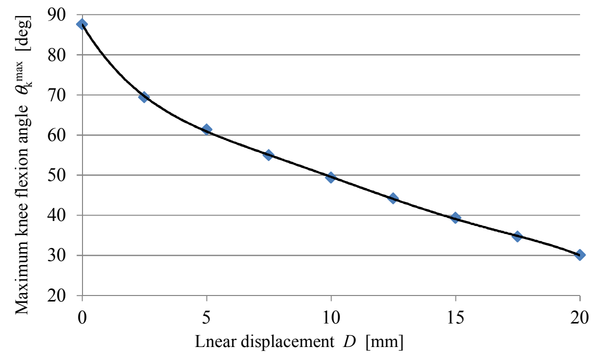


Fig. 7. Kinematic relationship between linear displacement D and maximum knee flexion angle θ_k^{max} The diamonds represent numerical solutions of the kinematics, and the curve denotes the fitted results by a fifth-order function

B. Statics

Suppose that the mechanism shown in Fig. 8 is in equilibrium. According to the principle of virtual work, eq.(1) is obtained (see Fig. 8).

$$\boldsymbol{f}_{th}^T \delta \boldsymbol{x}_{th} - f_k^y \delta D = 0 \tag{1}$$

where $\boldsymbol{f}_{th}$ denotes the force exerted on $\boldsymbol{x}_{th}$, and f_k^y denotes the vertical force exerted on the knee joint. δD denotes the virtual displacement of D. $\delta \boldsymbol{x}_{th}$ denotes the virtual displacement of the vector $\boldsymbol{x}_{th}$, which is given by eq.(2).

$$\boldsymbol{x}_{th} = \begin{bmatrix} -l_{th}\sin\theta_k \\ l_{th}\cos\theta_k + D_0 - D \end{bmatrix} = \boldsymbol{r}_{th} - \begin{bmatrix} 0 \\ D - D_0 \end{bmatrix} \tag{2}$$

where $\boldsymbol{r}_{th}$ denotes vector from the knee joint to the point $\boldsymbol{x}_{th}$ as shown in Fig.8, l_{th} denotes the length of $\boldsymbol{r}_{th}$, and D_0 denotes the natural length of the spring. Please note that eqs. (3) and (4) are obtained from an instantaneous kinematics analysis.

$$\delta \boldsymbol{x}_{th} = J_{th}(\theta_k)\delta\theta_k \tag{3}$$

$$\delta D = J_D(\theta_k)\delta\theta_k \tag{4}$$

where $J_{th}(\theta_k)$ and $J_D(\theta_k)$ denote Jacobian matrices obtained by differentiating the kinematic relationships. Substituting eqs. (3) and (4) into eq.(1) yields eq.(5).

$$J^T{}_{th}(\theta_k)\boldsymbol{f}_{th} - J_D(\theta_k) f_k^y(D) = 0 \tag{5}$$

where Jacobian matrix J_{th} is obtained as eq.(6) by differentiating eq.(2) with time:

$$J_{th}(\theta_k) = \begin{bmatrix} -l_{th}\cos\theta_k \\ -l_{th}\sin\theta_k - J_D(\theta_k) \end{bmatrix} \tag{6}$$

Substituting eq.(6) into eq.(5) yields eq.(7).

$$[0,0,1](\boldsymbol{r}_{th} \times \boldsymbol{f}_{th}) + J_D(\theta_k)\{-f_k^y - f_{th}{}^y\} = 0 \quad (7)$$

According to eq.(7), total knee flexion moment τ_k is calculated using eq.(8):

$$\tau_k = M_{th} + [0,0,1](\boldsymbol{r}_{th} \times \boldsymbol{f}_{th}) + J_D(\theta_k)\{-f_{sp}^y(D) - f_{th}{}^y\} \quad (8)$$

where M_{th} and $f_{sp}(D)$ denote the torque exerted on the thigh socket by the hip moment and the spring force, respectively.

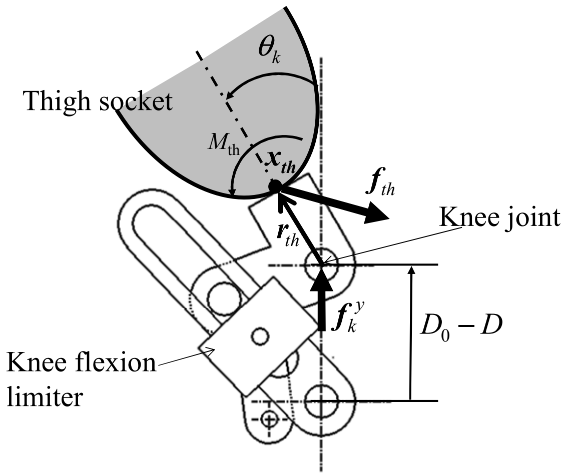


Fig. 8. Detailed mechanism around knee joint and knee flexion limiter. Force exerted on the key segment is shown in detail.

Let us consider the case where $M_{th} = 0$ and $\boldsymbol{r}_{th} \times \boldsymbol{f}_{th} = \boldsymbol{0}$. In such a case, the extension moment $\tau_k < 0$ can be written as eq.(9).

$$J_D(\theta_k)\{f_{sp}(D) + f_{th}{}^y\} > 0 \quad (9)$$

It is clear that $J_D(\theta_k) < 0$ is satisfied, as seen in Fig. 7. This yields eq.(10).

$$f_{sp}(D) + f_{th}{}^y < 0 \quad (10)$$

This inequality shows that if the vertical force $f_{th}{}^y$ is greater than the reaction force generated by the spring, then the extension moment is generated around the knee joint axis. This means that the knee joint extension moment can be generated by adding vertical force $f_{th}{}^y$, even when $M_{th} = 0$ and $\boldsymbol{r}_{th} \times \boldsymbol{f}_{th} = \boldsymbol{0}$, which are usually needed to generate the extension moment.

IV. Stair Ascent Experiments

A. Method

Two healthy males (21 and 22 years old) who gave informed consent participated in the experiments. A simulated socket was used to allow the individuals with intact limbs to participate in the experiments, as shown in Fig. 3.

Each participant was fitted with the following three knee prostheses to compare their gaits during stair ascent.

a) 3R95: The single-axis modular knee joint 3R95 (Otto Bock), which uses a simple damper mechanism. Its weight was 0.335 kg.
b) NFJ (no friction joint): a very low friction knee joint that was produced by eliminating the springs for knee extension from the single-axis modular knee joint 3R15 (Otto Bock). Its weight was 0.475 kg.
c) LKJ (link knee joint): Our developed novel knee joint. Its weight was 0.985 kg.

The foot part IS49 (Otto Bock) was used under all the prosthesis conditions. The simulated socket and prosthesis were used on each participant's right leg. The knee joint condition mentioned above was treated as the within-subject factor. The order of the conditions varied between the two participants. After experiencing the three knee joint levels, stair ascent with intact limbs without any prosthesis was also studied for comparison.

For LKJ, the laboratory staircase consisted of five steps (rise height: 17 cm, tread length: 30 cm, and width: 90 cm). For the experiments with the other prostheses and the intact limbs, the staircase consisted of three steps (rise height: 16 cm for the first step and 17 cm for the other steps, tread length: 30 cm, and step width: 90 cm).

Force plates were installed on the flat ground and on the first step. A motion capture system, Move-tr/3DS (Library Ltd.), which used six near-infrared high-speed cameras, was used to measure the three-dimensional positions of reflection markers attached to the left toe, left lateral malleolus, left knee joint, both greater trochanters, the prosthetic knee joint, and the toe of its foot part. The sampling rates of the motion capture system and the force plates were 150 Hz and 3000 Hz, respectively, and they were synchronized. The thigh angle and knee joint angle, the definitions for which are given in Fig. 9, were calculated from the measured marker positions.

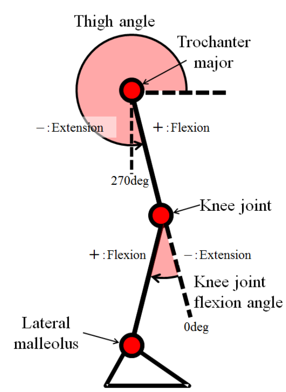


Fig. 9. Definitions of angles

Trial walking and stair ascent were performed under each knee joint condition before each measurement. Then, the participants were asked to maintain their stance posture at the flat ground and then start to ascend using the right leg or the leg with the prosthesis, as shown in Fig. 10. No instructions were given about the velocity of the ascent, which implies that the participants determined the velocity. Three trials were conducted under each knee joint condition. After the experiments with the three knee joints, three stair ascent trials with intact legs without any prosthesis were conducted for each participant. The data between the first toe off of the right limb (RTO) to the next RTO are shown in Fig. 10. The period between LTO and LHC represents the single stance phase of the prosthesis. The knee joint moment and knee joint power were calculated using a quasistatic analysis.

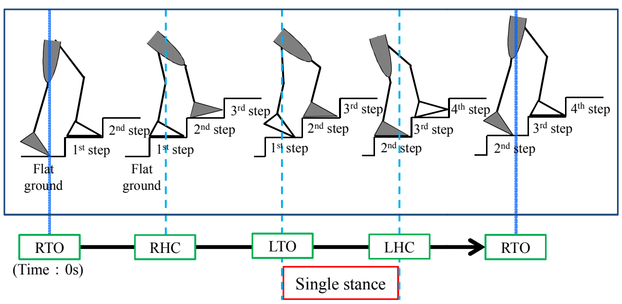


Fig. 10. Analysis range of stair ascent. RTO: right toe off, RHC: right heel contact, LTO: left toe off, LHC: left heel contact

B. Results

1) LKJ realized stair ascent without handrail

The experimental results revealed that both participants succeeded in ascending stairs with LKJ without the use of a handrail (Fig. 11). It should be noted that it was impossible to realize stair ascent without a handrail using the other two prostheses.

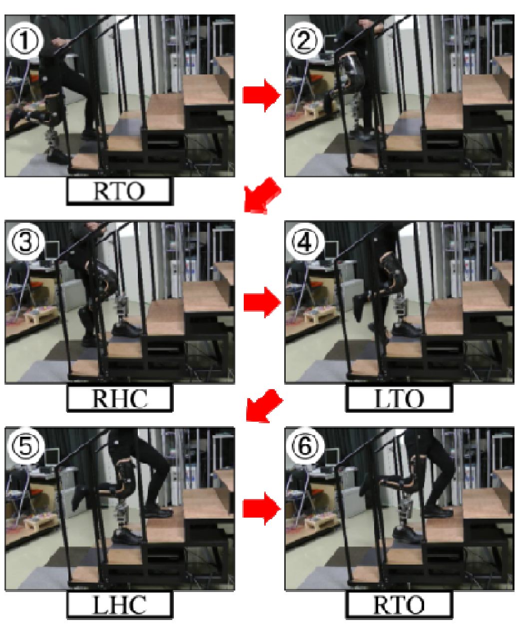


Fig. 11. Stair ascending with LKJ (participant O)

2) Gait analysis

Typical examples of the thigh angle, knee joint flexion angle, joint moment, and joint power of the intact and prosthetic legs (participant O: body weight is 51kg) are shown in Fig. 12. The NFJ results were similar to those for 3R95. Therefore, they were omitted from Fig. 12.

The thigh angle and knee flexion angle were smaller with LKJ than with an intact limb at LTO, where the leg with LKJ was aligned more straight. This difference would have been caused by the length of the shank. The knee joint of the LKJ unit was positioned lower than that of the intact limb because the transfemoral prosthesis simulator was used. Although the time-series curves of the angles did not show the same tendencies between LKJ and the intact limb, the knee joint of the LKJ unit extended through the prosthetic single stance phase with no unintended knee flexion. Meanwhile, the thigh angle became approximately 270°, and the knee joint fully extended in the early part of the prosthetic single stance phase under the LKJ and 3R95 conditions. The straight alignment of the prosthetic leg would work to prevent unintended knee flexion during the prosthetic stance phase without particular functions.

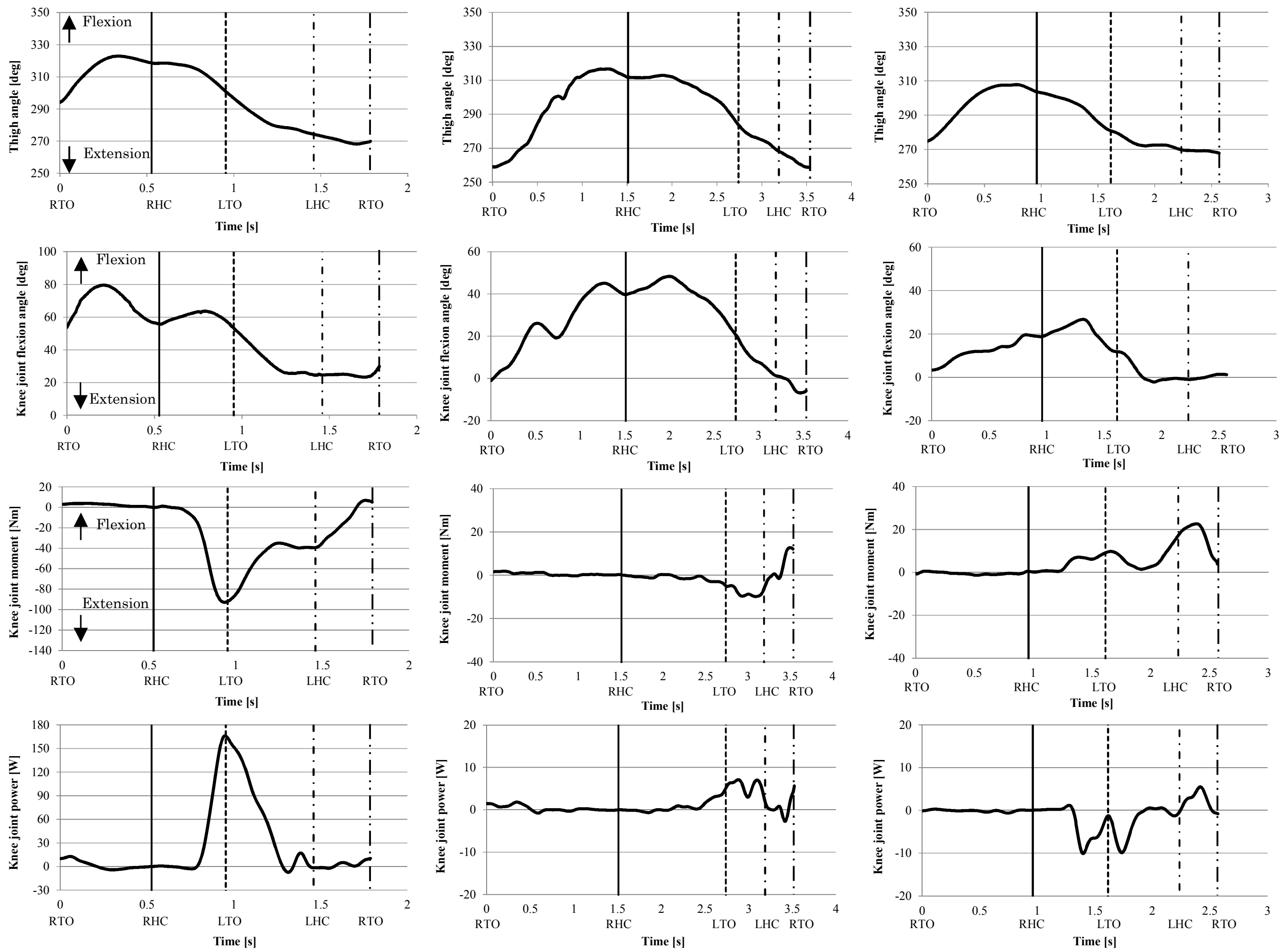
Positive joint moment power generation was observed with the knee extension moment and movement under the LKJ condition. In contrast, negative power was observed for the knee flexion moment and extension movement with 3R95 because an oil damper was used in the knee mechanism. Thus, the observation of positive power generation with LKJ indicates that the functions designed for LKJ worked appropriately. The positive joint moment power of LKJ was derived by transducing the negative power at the linear joint embedded in the LKJ unit. However, the amount of joint moment and power generated by LKJ were not as large as those for the intact limb. One of joints contributing the most to compensate the lack of power for the prosthetic stance phase would be the hip joint of the prosthetic side.

Further, the results for the other participant showed the same trends.

V. Conclusion

A knee joint mechanism that generates the knee extension moment to prevent unintended knee flexion and realize knee extension was proposed for ascending stairs when using a prosthesis. The stair ascent experiments demonstrated that the proposed knee joint LKJ allowed the participants to ascend stairs without any assistive device such as a handrail. In addition, it was shown that LKJ was able to generate knee extension movement that realized stair ascent with positive power without any power source. This positive power arose from an increase in the linear displacement D, which indicates a loss of potential energy.

In a future study, the proposed basic mechanism for the stair ascent should be applied to a conventional knee joint to realize level walking.



(a) Intact limb (b) LKJ (c) 3R95

Fig. 12. Knee joint angle, thigh angle, knee joint moment, and knee joint power of prosthesis (participant O)

REFERENCES

[1] M. Kutz, Standard Handbook of Biomedical Engineering & Design Handbook, Chapter 3, McGraw-Hill Professional, 2002

[2] A. Nakagawa, "A Swing Phase Control of Above Knee Prosthesis – The Development of The Intelligent A/K Prosthesis", Journal of the Society of Biomechanisms, Vol.14 supplement, pp.101-111, 1990.

[3] K. James, et al. "Active Suspension Above Knee Prosthesis", Proceedings of Sixth International Conference on Biomedical Engineering, pp.317-320, 1990.

[4] T. Chin, S. Sawamura, R. Shiba, H. Oyabu, Y. Nagakura, I. Takase, K. Machida, and A. Nakagawa, "Effect of an Intelligent Prosthesis (IP) on the Walking Ability of Young Transfemoral Amputees: Comparison of IP Users with Able-Bodied People", American Journal of Physical Medicine and Rehabilitation, Vol.82, pp. 447-451, 2003.

[5] T. Wada, T. Takeuchi, M. Sekimoto, Y. Shiba, and Arimoto, S., "Evaluation of Gait with Trans-Femoral Prosthesis based on Riemannian Distance", Proceedings of IEEE International Conference on Mechatronics and Automation, pp.1021-1026, 2009.

[6] T. Wada, T. Takeuchi, M. Sekimoto, Y. Shiba, and S. Arimoto, "Evaluation of Inertia Matching of Trans-Femoral Prosthesis based on Riemannian Distance", Proceedings of International Conference on Complex Medical Engineering, pp.244-249, 2010

[7] T. Takeuchi, T. Wada, K. Awakihara, and M. Sekimoto, "Analysis of Walking Skill with Trans-Femoral Prosthesis based on Inertia-Induced Measure", Proceedings of IEEE/ICME International Conference on Complex Medical Engineering, pp.641-646, 2011

[8] H. Hobara, Y. Kobayashi, T. Nakamura, N. Yamasaki, K. Nakazawa, M. Akai, and T. Ogata, "Lower Extremity Joint Kinematics of Stair Ascent in Transfemoral Amputees", Prosthetics and Orthotics International, Vol.35, No.4, pp.467-72, 2011

[9] A. Protopapadaki., W. I. Drechsler, M. C. Cramp, F. Coutts, and C> M. Scott, "Hip, Knee, Ankle Kinematics and Kinetics during Stair Ascent and Descent in Healthy Young Individuals", Clinical Biomechanics, Vol.22, pp.203-210, 2007

[10] F. Sup, A. Bohara, and M. Goldfarb, "Design and Control of A Powered Transfemoral Prosthesis", The International Journal of Robotics Research, Vol.27, No.2, 2008

[11] Power Knee Webpage, www.ossur.com/owerknee/

[12] M. Bellmann, T. Schmalz, E. Ludwigs, and S. Blumentritt, "Immediate Effects of a New Microprocessor-Controlled Prosthetic Knee Joint: A Comparative Biomechanical Evaluation", Archives of Physical Medicine and Rehabilitation, Vol.93, No.3, pp.541-549, 2012

[13] M. Ninomiya, "Development of a Knee Joint for a Trans-femoral Prosthesis that can Ascend and Descend Slopes and Stairs", Proceedings of International Society for Prosthetics and Orthotics 2004 World Congress, 2004

[14] T. Wada, K. Inoue, and M. Ozaki., "A Knee Joint Control Method of Transfemoral Prosthesis for Stair Ascent", Asian Prosthetic and Orthotic Scientific Meeting, 2012