# Effects of Functional Range of Knee Extension for Transfemoral Prosthesis on Stair Ascent Motion

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*Abstract*— **We previously proposed a passive mechanism as the link knee joint unit (LKJ) for a transfemoral prosthesis for stair ascent. The prototype allowed the experimental subjects to ascend stairs without the use of a handrail. In the present study, we modified the LKJ unit and developed further two designs of the LKJ unit. One has full knee extension function during the prosthetic stance phase (condition 1). The other design mechanically trades off the functional range of knee extension against stability of the LKJ unit (condition 3). In the stair ascent experiment with six able-bodied subjects, all subjects succeeded in ascending stairs with the three LKJ conditions without the use of a handrail. No difference was found in joint angels and joint moments of the intact and prosthetic legs among all LKJ conditions. However, subjective assessment for ease of LKJ extension during stair ascent showed that the participants felt easier to extend the LKJ unit in the condition 1 and 2 than the condition 3. It is suggested that the condition 1 or 2 is appropriate for prosthesis users who can ascend stairs with the LKJ unit. For prosthesis users who are not familiar with the LKJ unit, the condition 3 would be useful to learn how to use it.**

## I. INTRODUCTION

Function of the knee joint unit is important for transfemoral prostheses because it mainly determines the activity level of prosthesis users. The advancements of the mechanism and control for prosthetic knees 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]. Nevertheless, it is known that prosthesis users find it difficult to ascend stairs. Therefore, 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.

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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 [5][6]. 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 [7][8][9]. 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. Against this problem, we proposed a passive knee joint mechanism of transfemoral prosthesis for stair ascent in the previous study [10]. The link knee joint (LKJ) unit can extend its knee by weight shit to the LKJ unit. The LKJ unit demonstrated that it realized stair ascent without handrail for able-bodied subjects who used a simulated socket.

However, we have found necessities to improve the LKJ unit from the stair ascent experiment of the literature [10]. The range of the LKJ extension function was designed based on the motion of able-bodied persons, who do not fully extend their knees at stance phase. Nevertheless, the LKJ was fully extended at the end of the prosthetic stance phase in the stair ascent experiment. Furthermore, it seemed difficult to control the LKJ extension function in the first half of the prosthetic stance phase. In this period, the load on the LKJ unit which caused by weight shift might not be able to extend the knee joint and even might induce unintended knee flexion.

The present study addressed these problems modifying the LKJ design for each problem. For the specific motion such as full knee extension of the transfemoral prosthesis, we expanded the knee extension range (condition 1). Meanwhile, for the lack of stability in first half of the prosthetic stance phase, we contracted the knee extension range because trade-off relationship mechanically exists between knee extension range and stability of the LKJ unit (condition 3). In the present study, first, the mechanics of the LKJ unit and its modified designs were introduced. Then, the effects of prototypes of the LKJ unit on stair ascent motion were demonstrated with able-bodied subjects in the experiments. We hypothesized that each design of the LKJ unit differently affects stair ascending motion and subjective assessment. The condition 1 (full knee extension) would reduce joint moments and make joint angles similar to intact limb motion than the existing model (condition 2). The condition 3 would be stable but require more joint moment to ascend stairs.

## II. Mechanism of the LKJ Unit

### A. Structure and Kinematics

The LKJ unit for a prosthesis with a link mechanism was proposed, as shown in Fig. 1 and 2. As shown in Fig. 3, knee joint $J_k$ is represented by the relative rotation between part A and L-shaped link $L_1$. Part A is connected to part B through the 1-dof translational joint with a linear spring. This linear displacement/shortening of the joint from the original state without any load on the pyramid adapter is called the linear displacement $D$. Therefore, the positive direction of $D$ is downward. $L_1$ and $L_2$ are connected at $P_1$, but $P_1$ can move in the slit of $L_2$. The maximum knee flexion angle $\theta_k^{max}$ is determined by the collision of $P_1$ and the bottom of the slit of $L_2$ (Fig. 4). The linear displacement $D$ is increased by increasing the force on the pyramid adapter caused by weight shift to the LKJ unit, leading to the sliding up of the bottom of $L_2$ slit and the sliding down of $P_1$. This results in a decrease in the maximum knee flexion angle $\theta_k^{max}$, which functions as knee extension (Fig. 4). Kinematic analysis shows the unique relationship between the maximum knee flexion angle $\theta_k^{max}$ [deg] and linear displacement $D$ as following.

$$\theta_k^{\max} = 100 - \cos^{-1}\left(\frac{D^2 - 2bD + 2ab\cos(90-\alpha)}{2a(b-D)}\right) \tag{1}$$

where $a$ [m], $b$ [m] and $\alpha$ [deg] are the geometric parameters of the LKJ unit defined in Fig. 3.

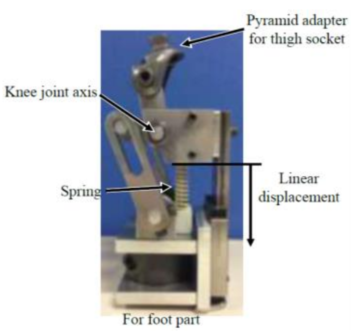


Figure 1. Photo of the proposed knee joint unit mechanism.

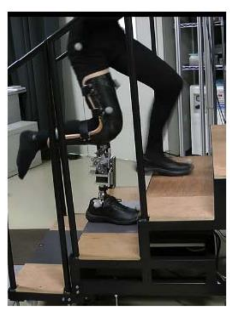

Figure 2. Photo of developed knee joint with foot part.

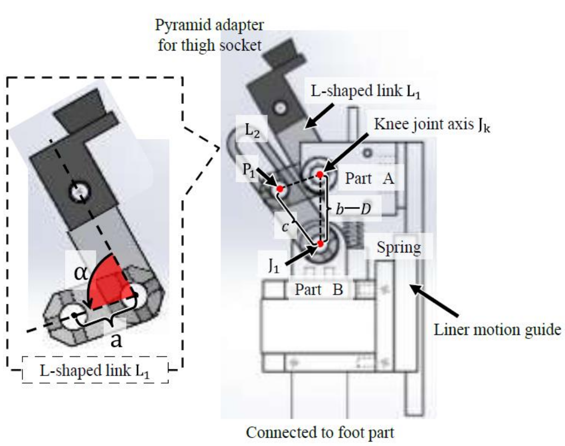


Figure 3. The proposed link mechanism diagram.
Part A slides down with the force on the pyramid adapter caused by weight shit.

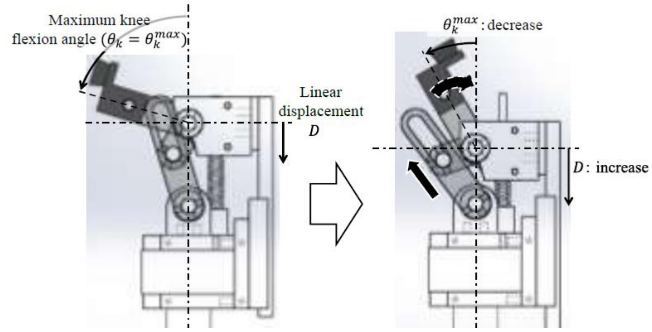


Figure 4. Extension function of the proposed knee joint unit. $\theta_k^{max}$ decreases when $D$ increases.

### B. Statics for Knee Extension Function

It is assumed that the prosthesis stands upright, that the mass of each part of the LKJ unit is zero, and that $\theta_k^{max}$ is being limited by $L_2$. The forces and moment acting on the LKJ unit and the prosthesis are shown in Fig. 5. The origin is set at $J_k$ , the horizontal direction is x-axis, the vertical direction is y-axis. The force action on the top of $L_1$ from the thigh socket is defined as $\boldsymbol{F}_1 = [F_{1x} \;\; F_{1y}]^T$. Then, force $\boldsymbol{F}_2 = [F_{2x} \;\; F_{2y}]^T$ is from $L_2$, and force $\boldsymbol{F}_3 = [F_{3x} \;\; F_{3y}]^T$ is from the spring and the linear joint to $L_1$. In the aspect of whole prosthesis, the ground reaction force $\boldsymbol{F}_4 = [F_{4x} \;\; F_{4y}]^T$ is generated against $\boldsymbol{F}_1$. $\boldsymbol{r}_1 = [r_{1x} \;\; r_{1y}]^T$, $\boldsymbol{r}_2 = [r_{2x} \;\; r_{2y}]^T$, $\boldsymbol{r}_4 = [r_{4x} \;\; r_{4y}]^T$ indicate the position vectors to the point of application of $\boldsymbol{F}_1$, $\boldsymbol{F}_2$, $\boldsymbol{F}_4$, respectively. The vector from the intersection of the planter surface of the prosthetic foot and y-axis to the heel is $\boldsymbol{r}_R = [r_{Rx} \;\; 0]^T$, and to the toe is $\boldsymbol{r}_F = [r_{Fx} \;\; 0]^T$, respectively. $M_1$ is moment of force acting on the top of $L_1$ from the thigh socket, indicating the hip joint moment.

Suppose that the mechanism shown in Fig. 5 is in equilibrium. According to the equilibrium of force and moment of force acting on the whole prosthesis, the following (2) and (3) are obtained.

$$\boldsymbol{F}_1 + \boldsymbol{F}_4 = 0 \tag{2}$$

$$M_1 + \begin{vmatrix} r_{1x} & r_{1y} \\ F_{1x} & F_{1y} \end{vmatrix} + \begin{vmatrix} r_{4x} & r_{4y} \\ F_{4x} & F_{4y} \end{vmatrix} = 0 \tag{3}$$

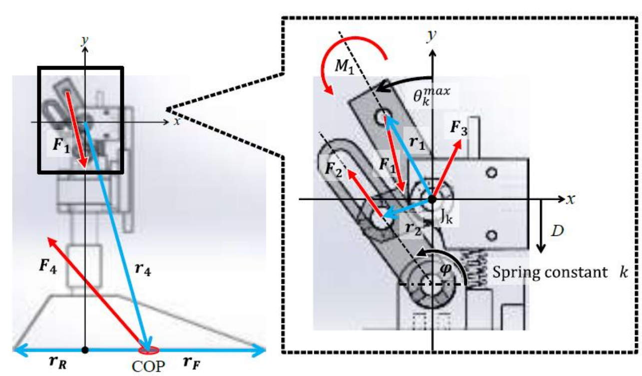


Figure 5. The forces and moment acting on the LKJ unit and the prosthesis.

Then, substituting the range of plantar surface size ($\boldsymbol{r}_R$ and $\boldsymbol{r}_F$) into the above two equations yields the following (4) and (5).

$$F_{1y}(r_{Rx} - r_{1x}) + F_{1x}(r_{4y} - r_{1y}) + M_1 \geq 0 \tag{4}$$

$$F_{1y}(r_{1x} - r_{Fx}) + F_{1x}(r_{4y} - r_{1y}) + M_1 \leq 0 \tag{5}$$

According to the equilibrium of force acting on $L_1$ and moment of force around $J_k$, the following (6) and (7) are obtained.

$$\boldsymbol{F}_1 + \boldsymbol{F}_2 + \boldsymbol{F}_3 = 0 \tag{6}$$

$$M_1 + \begin{vmatrix} r_{1x} & r_{1y} \\ F_{1x} & F_{1y} \end{vmatrix} + \begin{vmatrix} r_{2x} & r_{2y} \\ F_{2x} & F_{2y} \end{vmatrix} = 0 \tag{7}$$

Let us consider the case where the sum of the right hand members of (7) is negative, which indicates that knee extension moment is generated around $L_1$ or $J_k$. Here, substituting the geometric parameters of the prosthesis, such as the prosthetic foot size and the shank length, into this case yields the following (8).

$$F_{1y}(r_{1x} - r_{2x} + \frac{r_{2y}}{\tan\varphi(D)}) - F_{1x}r_{1y} - kD(r_{2x} - \frac{r_{2y}}{\tan\varphi(D)}) + M_1 \leq 0 \tag{8}$$

where $F_{1y} = -kD$, $k$ is spring constant, and $\varphi$ is the angle of $L_2$ which is uniquely dependent on $D$ like $\theta_k^{max}$ (see Fig. 5). When $\boldsymbol{F}_1$ meets (4), (5) and (8) simultaneously, the prosthetic knee extends or does not rotate. Figure 6 is a schematic diagram of the requirement area of $\boldsymbol{F}_1$ for knee extension.

### *C. Modification of the LKJ Mechanism*

Changing the geometric parameters of the LKJ unit (see Fig. 3 and (1)) provides two different functional range of knee extension and requirement area of $\boldsymbol{F}_1$ for knee extension. In addition to the existing model (condition 2) in which the LKJ extension function works by 30 degrees of $\theta_k^{max}$, we proposed two different functional range of knee extension of the LKJ unit. One is full knee extension (0 degrees, condition 1 in Fig. 7) based on the transfemoral prosthetic motion. However, it narrows the requirement area of $\boldsymbol{F}_1$ for knee extension as shown in Fig. 8. The other has wider requirement area of $\boldsymbol{F}_1$ for knee extension (condition 3 in Fig. 8), indicating more stable, instead of smaller range of knee extension function. The LKJ extends by 60 degrees of $\theta_k^{max}$ with its mechanical function in condition 3 (Fig. 7).

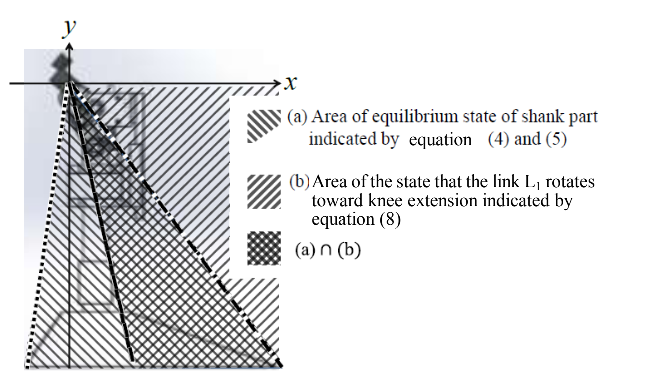


Figure 6. Schematic diagram of the requirement area of $\boldsymbol{F}_1$ for knee extension. The requirement area of $\boldsymbol{F}_1$ is the overlapped area for the shank part stable (a) and for $L_1$ rotation toward knee extension (b).

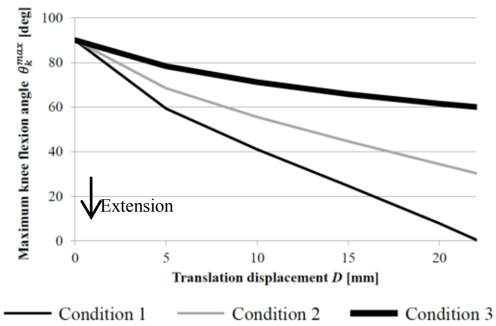


Figure 7. Kinematic relationship between the linear displacement $D$ and maximum knee flexion angle $\theta_k^{max}$.

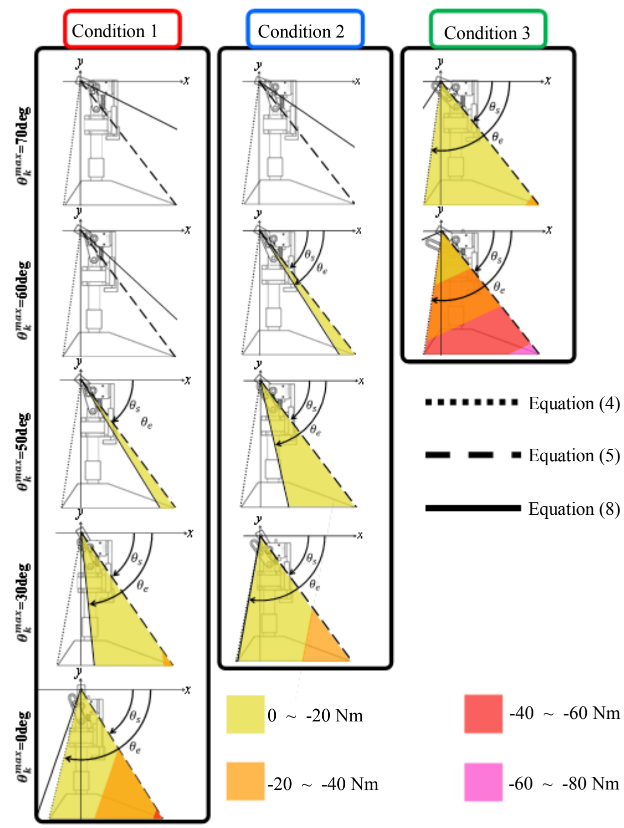


Figure 8. The requirement area of $\boldsymbol{F}_1$ for knee extension. The filled areas indicate the direction of the force $\boldsymbol{F}_1$ for knee extension obtained from (8). No filled area shown in the condition 1 and 2 indicates that the LKJ unit does not extend its knee with its function and that the hip extension moment $M_1$ is necessary for knee extension.

## III. Stair Ascent Experiment

### *A. Methods*

Six able-bodied males participated in the experiments. A simulated socket was used to allow the individuals with intact limbs (the right leg) to participate in the experiments, as shown in Fig. 2. The laboratory staircase consisted of four steps (rise height: 0.17 m, tread length: 0.30 m, and width: 0.90 m), and force plates (Library Ltd., Japan) were embedded in the first and second steps. A motion capture system, Move-tr/3DS (Library Ltd., Japan), which used six near-infrared high-speed cameras, was used to measure the three-dimensional positions of reflection markers attached to the legs. The sampling rates of the motion capture system and the force plates were 100 Hz and 1000 Hz, respectively, and they were synchronized. The cadence of stair ascending was controlled at 40 bpm for each step with a digital metronome. Three trials were conducted under each LKJ condition. After the experiments with the LKJ conditions, three stair ascent trials with intact legs without any prosthesis were conducted for each participant. Subjective assessment was conducted with visual analog scale to evaluate ease of LKJ extension. The gait data between the first toe off of the right limb (RTO) to the next RTO (fig. 9) were analyzed. The joint angles were obtained from the measured marker positions, and the joint moments were calculated using inverse dynamics.

## *B. Results and Discussion*

As the result of the experiment, all participants succeeded in ascending stairs with the three LKJ conditions without the use of a handrail. We first hypothesized that each LKJ condition differently affects gait, but marked difference was not found in the joint angles and moments of both of prosthetic and intact sides among the three conditions (Fig. 10). This result suggests that once prosthesis users learn how to ascend stairs with the LKJ mechanism, the motor skills can be applied to other LKJ conditions.

Compared to gait analysis, subjective assessment showed that the participants felt easier to extend the LKJ unit in the condition 1 and 2 than the condition 3. Therefore, it is suggested that the condition 1 or 2 are appropriate for prosthesis users who can ascend stairs with the LKJ unit. For prosthesis users who are not familiar with the LKJ mechanism, the condition 3 which is stable modification at the first half of the prosthetic stance phase would be useful to learn how to use the LKJ unit.

Thus, the modified designs developed in the present study provides the different functions, but they did not affect the stair ascent motion. The different modifications could provide adjustment of the LKJ function for users who have different motor skills. The simulated socket may differently effect the motion because it changes the knee position compared to amputees, so further study is needed.

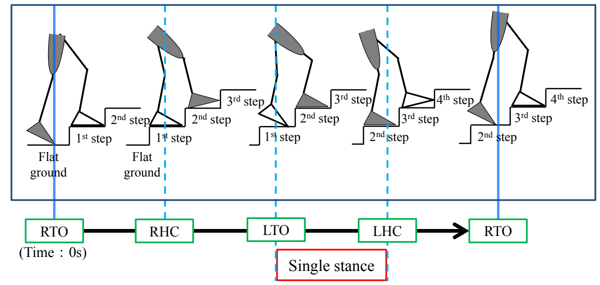


Figure 9. Analysis range of stair ascent. RTO: right toe off, RHC: right heel contact, LTO: left toe off, LHC: left heel contact. The leg which filled by gray indicates the prosthetic (right) side.

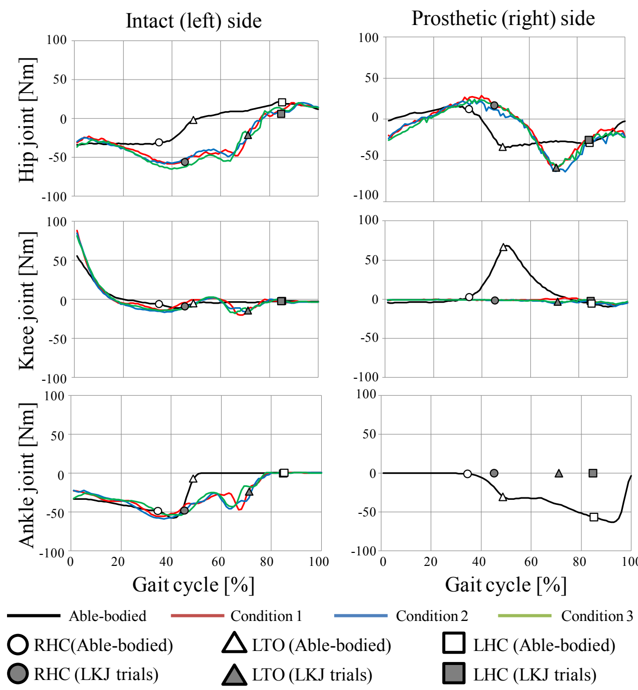


Figure 10. Joint moment of the lower extremities during stair ascending. The three LKJ conditions did not show marked difference each other.

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