Proposal of Powered Foot Prosthesis Emulating Motion of Healthy Foot (PEHF)

Yoshitoshi Murata, Haruki Baba
Faculty of Software and Information Science
Iwate Prefectural University
Takizawa, Japan
e-mail: y-murata@iwate-pu.ac.jp, g031q310@s.iwate-pu.ac.jp

Yukihide Nishimura
Department of Rehabilitation Medicine,
Iwate Medical University
Morioka, Japan
e-mail: ynishi@iwate-med.ac.jp

***Abstract*— Several million people around the world live with limb loss. Foot prosthesis is useful to improve their quality of life, and powered foot prosthesis enables them to walk naturally. However, most are too expensive for most amputees to afford. We propose a novel powered foot prosthesis that emulates the motion of a healthy foot with a half cycle delay. When a healthy person is walking, the motions of both feet are basically the same, with a half cycle difference. On the basis of this principle, the angle velocity of the foot part of the proposed prosthesis changes in the same way as the angle velocity of the healthy foot, with a half cycle delay. After introducing the measured motion data for both feet, a prototype of the proposed foot prosthesis is presented. Since this foot prosthesis adopts an industrial cylinder motor to push and pull the foot part of prosthesis, its cost would be dramatically lower than that of existing foot prosthetics.**

Keywords-foot prosthesis; power driven; leg amputee; walking gait; Quality of Life, QOL.

I. INTRODUCTION

There are nearly 2 million people living with limb loss in the United States [1]. LeBlanc [2] estimated the number of amputees is the world was approximately 10 million in 2008, with 30% of them being arm amputees [2]. Therefore, the number of leg amputees was 7 million. Leg amputees use foot prosthesis to improve their quality of life. However, low-priced foot prostheses have rigid ankle parts and no power drive mechanism, which make it difficult to raise the heel and swing the toes up. Therefore, most users need more power to move their foot. Powered foot prostheses enable users to move their foot easily and walk more naturally. However, such prostheses are so expensive for most amputees.

One of the main reasons why the price of introducing an existing powered prosthesis is too expensive may be that the prosthetic market is not open. A few manufacturers provide them as an integrated device, and the components are not compatible between different manufacturers. Another reason is the complex structure.

Most existing powered prostheses measure the change of the ankle angle and torque using sensors mounted on the prosthesis; then, they estimate the state of gait transition and speed, and push or pull the foot part or toe part.

Therefore, we propose a foot prosthesis for the leg amputee whose foot left is healthy. The prosthesis emulates the motion of a healthy foot. When a healthy person is walking, the motions of both feet are basically the same, with a half cycle difference. The basic principle of our prosthesis is that the angle velocity of the foot part of the proposed prosthesis changes in the same as the angle velocity of the healthy foot, with a half cycle delay. The angle velocity of a healthy foot is measured with a gyro sensor mounted on the heel part of the shoe worn on the healthy foot. Since the prosthesis adopts a small industrial cylinder motor that pushes and pulls the foot part of the prosthesis, the rising and falling speed of the foot part are determined on the basis of the pulse speed inputted to a stepping motor. Since the price of small industrial stepping motors is not high and the structure of this foot prosthesis is simple, it would not be too expensive for most amputees to afford.

After introducing existing foot prosthesis in Section II, we describe basic principles of the proposed prosthesis in Section III. Changes of angle velocity and angle for both healthy feet are introduced in Sections IV. A prototype of proposed foot prosthesis is introduced in Section V. Section VI concludes with a summary of key points and future works.

II. EXISTING POWERED FOOT PROSTHETICS

In this section, we introduce existing powered foot prosthesis. Ottobock in Germany and Ösuur in Iceland provide such prosthesis to consumers and the Biomechatronics Group, a research group within MIT Media Lab., has also developed some models.

The powered foot prosthesis developed by the Biomechatronics Group, a research group within MIT Media Lab., is shown in Figure 1 [3]. Its heel part (in-series spring) is pulled up and down by the ball screw driven by the motor through the timing belt. In case of this prosthesis, the gait state transitions are determined using the ankle torque and ankle angle. These data are measured angle sensors and strain gauges mounted on the prosthesis [4]. These prostheses are not a commercial product.

Ottobock provides a power-assist foot prosthesis called “1B1 Meridium [5].” Its mechanism is shown in Figure 2. It adopts a hydraulic pressure mechanism, in which a hydraulic pressure cylinder pushes and pulls a lever on the toe plate, causing the instep to rise and fall, respectively. Unfortunately, we could not find practical kinds of sensors and control method of the toe plate. The control board would estimate the pushing timing and speed from the angle between the shank and floor (= ankle angle) from sensors.

Össur provides a power-assist foot prosthetic called “PROPRIO FOOT® [6]” shown in Figure 3. Due to a lack of relevant material on the product's operation, we assume

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from observations that an air cylinder positioned in the area of the Achilles' tendon raises and lowers the foot part. We could not find practical kinds of sensors and control method of the foot plate. Since its appearance resembles to the appearance of the prosthesis developed by the Biomechatronics Group, MIT, its control method would be basically same as Biomechatronics Group's control method, its drive system is not different from the Biomechatronics Group's one.

Prices of the PROPRIO FOOT® and 1B1 Meridium are not available to the public, but are assumed to be more than 2 million yen ($18,000) in Japan, which is too expensive for most amputees.

The main purpose of our research is to provide a low price powered prosthetic foot based on a module structure concept.

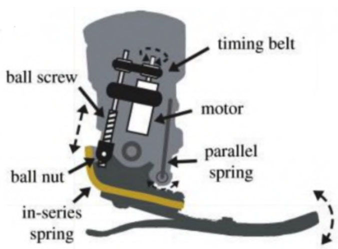

Figure 1. Power-assist foot prosthetic developed by the Biomechatronics Group of MIT Media Laboratory.

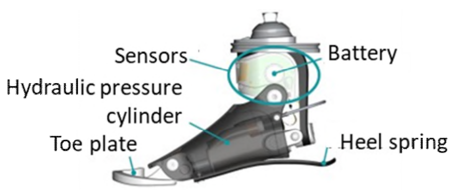

Figure 2. Mechanism of 1B1 Meridium, Ottobock.

Figure 3. PROPRIO FOOT®, Össur.

III. BASIC PRINCIPLE

As the result of measuring the differences between people with a walking disability and healthy people, it was found that the angle velocity when the heel rises and the toe angle when the foot swings for those with a walking disability are lower than healthy people [7]. We are developing walking assist devices that compensate for the shortage of heel raising power. In addition, we noticed that a foot prosthesis was similar to a person whose muscles were very weak and that the gait motion of both feet for healthy people was basically the same with a half cycle difference.

Most existing powered prostheses measure the change of the ankle angle and torque using sensors mounted on the prosthesis; then, they estimate the state of gait transition and speed, and push or pull the foot part or toe part.

However, in our proposed prosthesis, the motion of the foot part of the prosthesis is controlled on the basis of output from a gyro sensor mounted in the heel part of a shoe worn on a healthy foot, not sensors mounted in the prosthesis.

The basic structure of the proposed foot prosthesis is shown in Figure 4. The prosthesis is comprised of the following.

- Socket: a raw lower limb is inserted in a socket.
- Shin rod: a shin rod joins the socket and ankle joint.
- Ankle joint: can rotate foot up/down.
- Foot: one SACH foot is rotated up/down at the ankle joint part.
- Motor cylinder: the cylinder pushes/pulls the foot.
- Link: joins the shin rod to the motor and the motor to the foot.
- Gyro sensor: this is mounted into the heel part of the shoe worn on the healthy foot and the heel part of the foot prosthesis.
- Battery: provides electricity to the motor cylinder.
- Control board: controls the motor cylinder to emulate the gait motion of a healthy foot on the basis of output data from the gyro sensor mounted in the heel part of the shoe for the healthy foot.

The motor cylinder is a stepping motor, and the pulse stream input from the control board is used to change the pulse speed to emulate the motion of a healthy foot. The gyro sensor mounted into the heel part of the prosthesis is used to monitor the motion of the foot prosthesis. The output of both sensors is sent to the control board via Bluetooth. Bluetooth devices are mounted on the gyro sensor and control board.

This dual-link connection was adopted so as to rotate the foot smoothly.

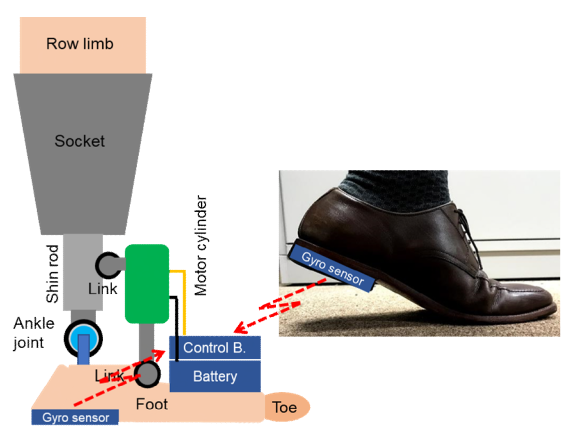

Figure 4. Basic structure of PEHF.

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IV. DIFFERENCES BETWEEN THE RIGHT FOOT AND LEFT FOOT WHILE WALKING

Our research is based on the principle that the motion of both feet for healthy people is basically the same and has a half cycle difference. A change of angle velocity and angle of both feet were measured. For this measurement, we used STEVAL-WESU1 by STMicro-electronics (See Figure. 5) [8]. This wearable unit includes four sensors:

- 3D-accelerometer,
- 3D-gyroscope,
- 3D-magnetometer,
- MEMS pressure.

This device is 37 x 40 x 8 mm and weighs 9.6 g.

We inserted STEVAL-WESU1 into the heel of a shoe as shown in Figure. 6.

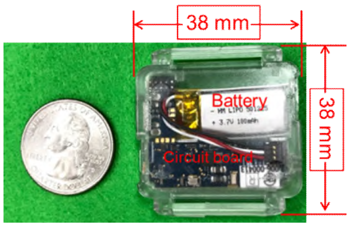


Figure 5. STEVAL-WESU1 by STMicroelectronics.

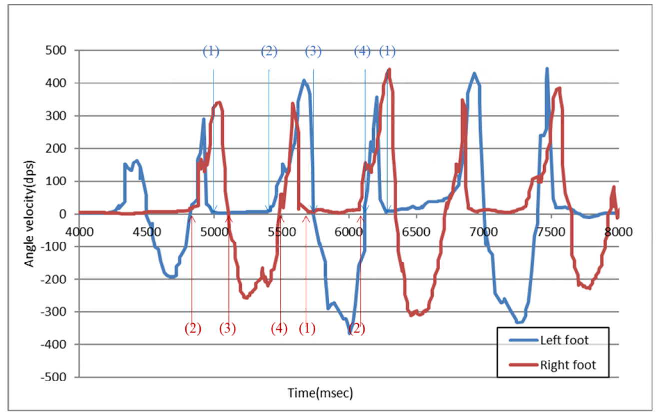
Figure 6. Sensor devices embedded into shoes.

A healthy participant walked on a straight line, 2-m-radius quarter-circle clockwise direction and 2-m-radius quarter-circle counter-clockwise direction.

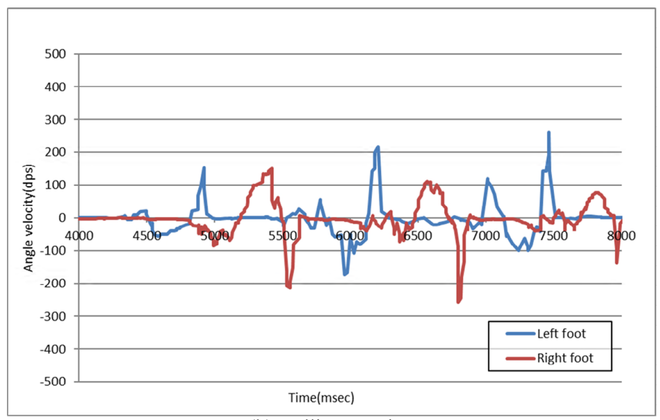
The measured angle velocity data are shown in Figures 7, 8, and 9. Figure 7 shows the data for walking on a straight line, Figure 8 shows that for walking on the 2-m-radius quarter-circle clockwise direction, and Figure 9 shows that for the 2-m radius quarter-circle counter-clockwise direction. (a) in each figure shows the change in pitch rotation, (b) shows the change in roll rotation, and (c) shows the change in yaw rotation.

The normal walking gait cycle is divided into four phases: (1) entire sole touching, (2) swinging backward (raising heel), (3) swinging forward, and (4) contacting floor. The acceleration is zero for entire sole touching, for the end of swing backward, and for the end of swing forward. The initial entire contact timing, end of the entire contact timing (initial raising heel), end of swing backward, and end of swing forward are shown in Figure 10.

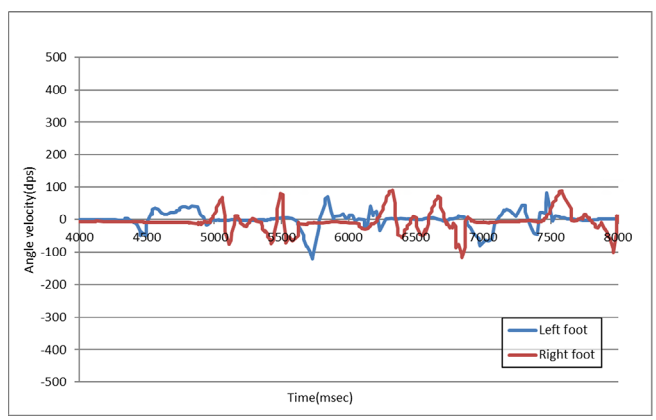
Figure 7 shows that the gait motions of both feet of a healthy person are basically the same and have a half cycle difference. The shape of the angle velocity curve for one foot was very similar to that for the other foot. A foot started to swing forward just after the other foot entirely touched the floor as shown in Figure 7 (a). In addition, the cycle periods of both feet were the same. The roll rotation changed synchronously with the pitch rotation as shown in Figure 7 (a) and (b). However, the ankle joint moved for the pitch rotation, and the change in roll rotation was caused by the motion of the knee. There was little change in yaw rotation because the participant was walking straight.



(a) Pitching rotation



(b) Rolling rotation

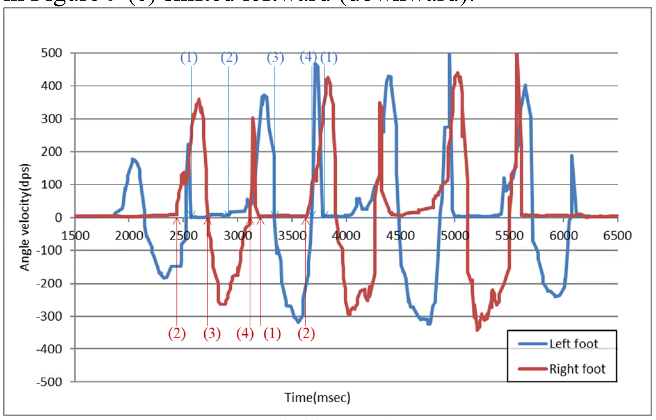


(c) Yawing rotation

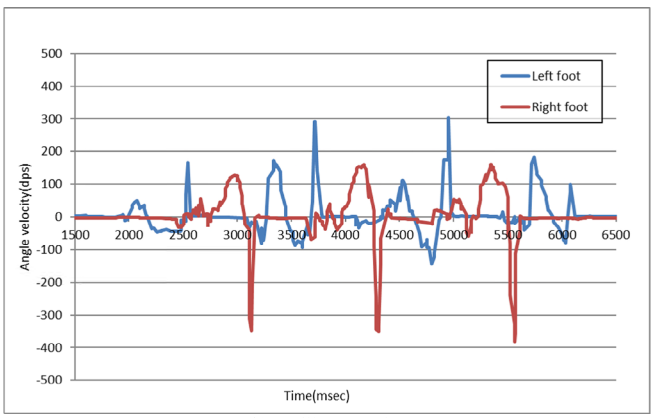
Figure 7. Walking on a straight line.
(1) initial entirely contact timing, (2) terminal entirely contact timing, (3) terminal swing backward, and (4) terminal swing forward

The timings of the gait cycle for walking on the 2-m-radius quarter-circle were the same as for walking on the straight line as shown in Figure 7 (a), Figure 8 (a), and Figure 9 (a).

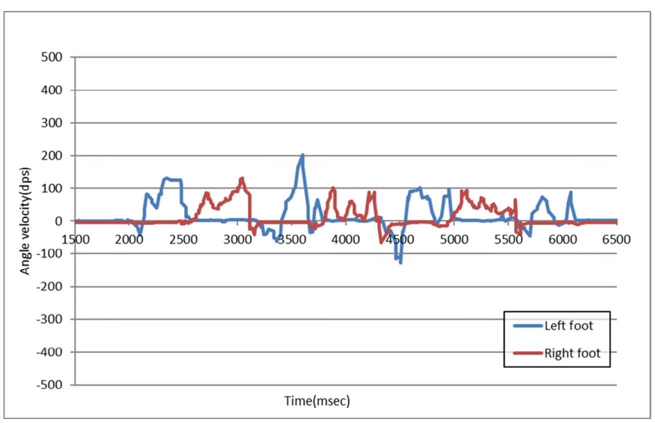
The cycle period of the left foot was the same as that of the right foot in Figure 8 (a) and Figure 9 (a). This characteristic is the same as Figure 7 (a). The yaw rotation in Figure 8 (c) shifted rightward (upward). The yaw rotation in Figure 9 (c) shifted leftward (downward).



(a) Pitching rotation



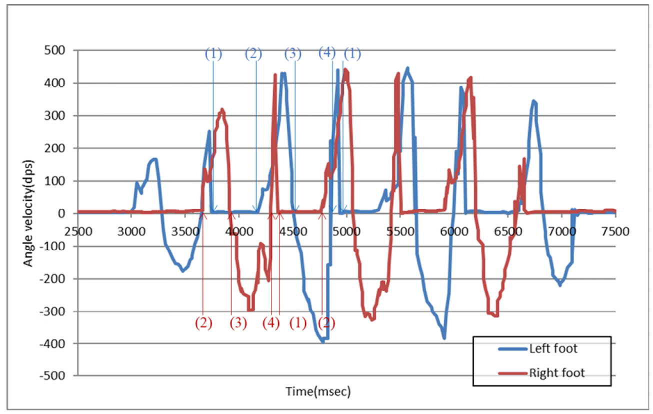
(b) Rolling rotation



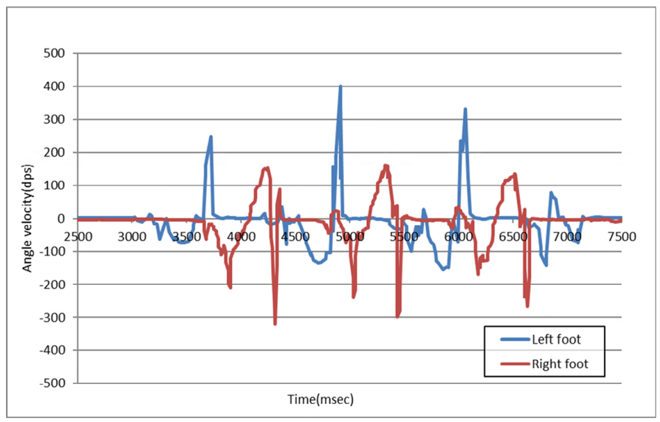
(c) Yawing rotation

Figure 8. Walking on 2 m radius quarter circle clockwise direction. (1) initial entirely contact timing, (2) terminal entirely contact timing, (3) terminal swing backward, and (4) terminal swing forward

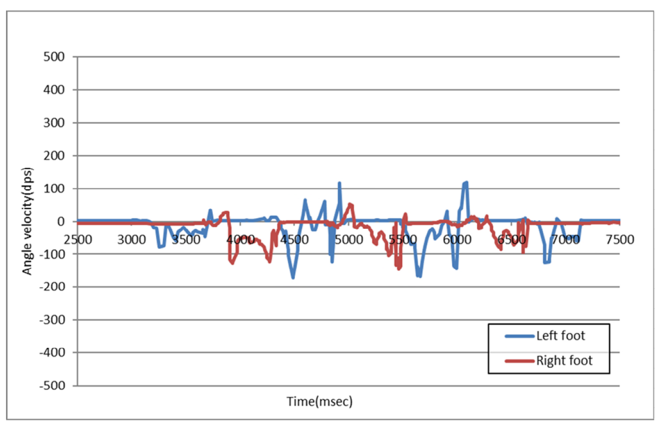
Before measuring the differences between both feet for gait motion, we assumed that the cycle period for the outside of the foot would be longer than that of the inside. However, the measured data shows there was no difference in cycle period between both feet. The participant in this case walked on a polygon, not a circle.



(a) Pitching rotation



(b) Rolling rotation



(c) Yawing rotation

Figure 9. walking on 2 m radius quarter circle counter-clockwise direction. (1) initial entirely contact timing, (2) terminal entirely contact timing, (3) terminal swing backward, and (4) terminal swing forward

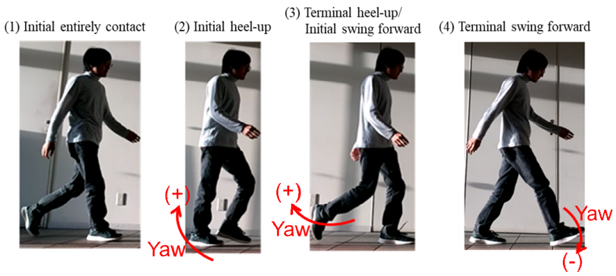


Figure 10. Normal walking gait cycle (See a right foot).

We would like the proposed prosthesis to be used not only for walking but also for jogging. Therefore, jogging motions must be measured to confirm whether there are differences between both feet.

V. PROTOTYPE OF PEHF

We developed a prototype of the foot prosthesis as shown in Figure 11. In case of the powered prosthesis designed by the Biomechatronics group of MIT Media Laboratory and PROPRIO by Össur, the heel part is pushed/pulled in order to rotate the pitch direction of the foot. However, the front part of the foot is pushed/pulled to rotate the foot in the prototype prosthesis because an existing SACH foot is used. A single motor cylinder is adopted to push/pull the foot. The motor is connected to the front of the shin and the foot via two links as shown in Figure 11.

For the single motor cylinder, we used an Oriental Motor DR series with a 30-mm stroke, 4-kg carrying force, 4-Newton thrust, and a 100-mm/sec maximum stroke speed [9]. It is a stepping motor that moves 0.001 mm per pulse. A stream of pulses is input to the motor. The pulse speed and timing are changed according to the output of the gyro sensor mounted on the heel part of the shoe worn on the healthy foot to emulate the motion of the healthy foot.

We examined the characteristics of the prototype when inputting a pulse stream to the motor. As shown in Figure 5, the sensor was attached on the heel part of the foot. The angle velocity and angle data were measured with 30-KHz and 60-KHz pulses. The measured data are shown in Figures 12 and 13. The maximum range of the angle of the prototype was 18 degrees, and the foot rotated 0.0008 degrees per pulse (dpp).

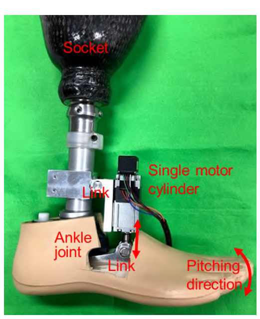
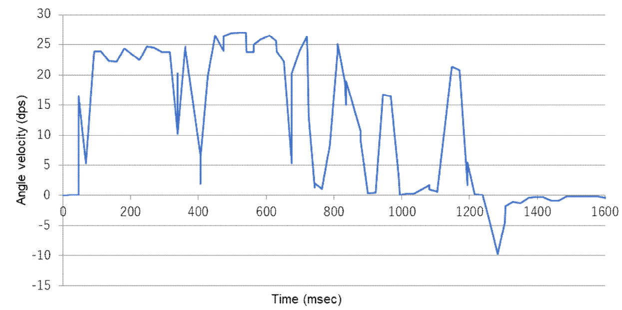
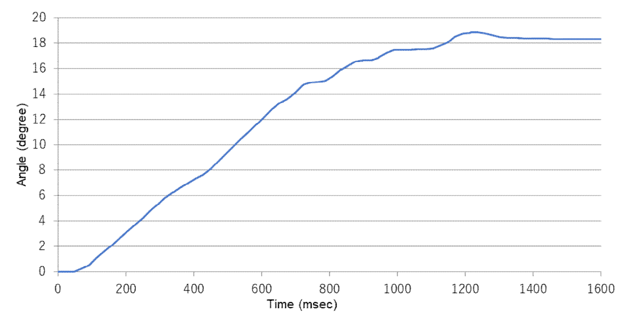


Figure 11. Prototype of PEHF.

The speed of the angle velocity did not stay the same for 30 and 60 Kpps, and the motor stopped at more than 60 Kpps. The reason is that its thrust was not enough. We have to solve this problem, which may involve changing the motor cylinder.

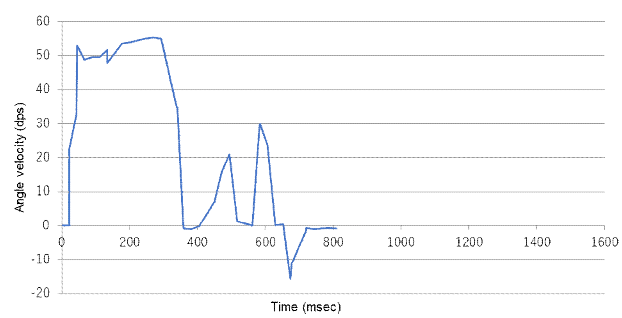


(a) Angle velocity

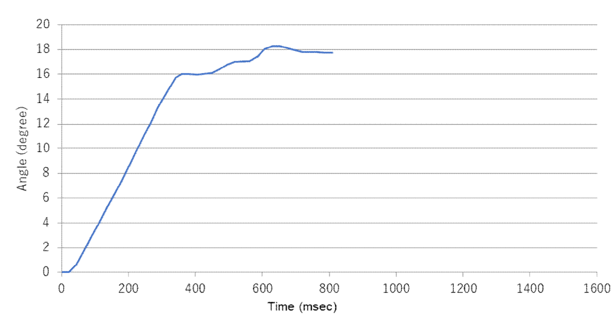


(b) Angle

Figure 12. Measured angle velocity and angle in 30 KHz pulse.



(a) Angle velocity



(b) Angle

Figure 13. Measured angle velocity and angle in 60 KHz pulse.

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VI. CONCLUSION AND FUTURE WORK

A foot prosthesis that emulates the gait motion of a healthy foot was proposed. After introducing the basic idea, we described the differences in motion between the right and left feet on the basis of the output of gyro sensors mounted in the heel part of the shoes. Basically, there were no differences between both feet except for the half cycle delay. The shape of the curve of angle velocity and the cycle period of the right foot was very similar to those of the left foot. This was the case not only when a participant walking on a straight line but also on a circle line except for the yaw direction. It was possible to recognize whether a person walked straight or turned to the right or left from the yaw data. Thus, it is possible to emulate the motion of a powered foot prosthesis with the gait motion of a healthy foot. Limb amputees could walk smoothly like a healthy person.

However, the research we introduced in this paper is the first step to develop the proposed prosthesis; at least we should work on the followings in the future;

(1) The introduced single motor cylinder could not drive a foot part well. We have to look for a suitable cylinder motor.
(2) Establishing a method for controlling cylinder motors on the basis of the angle velocity of the gyro sensor mounted into the heel part of the shoe worn on the healthy foot.
(3) Examining a prototype with a real limb amputee.
(4) Since the proposed foot prosthesis emulates motion of a healthy foot, leg amputees who use this prosthesis and have healthy foot; must start a walk with their healthy foot. We must investigate whether they could accept this limitation. If not, we have to introduce a new method to solve.

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