Design and Evaluation of ESR Prosthetic Foot with High Energy Storage

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***Abstract*-People with lower limb amputation in Thailand increase continuously every year. To alleviate those people, energy storage and release (ESR) prosthetic foot have been developed to replace the amputated part and to imitate energy storage and release of natural human ankle-foot system during stance phase. Design the energy release during push-off of ESR prosthetic foot to approach the human ankle-foot system is very important for providing symmetrical gait. Nevertheless, commercial ESR prosthetic foot was able to generate less energy release than human ankle-foot system. The purpose of this research was to development of the unique ESR prosthetic foot with high energy storage for amputees. The proposed prosthetic was comprised of two flexible carbon fiber composite structure. Each part function separately during early and terminal stance phase. Its function and strength was implemented by Finite element analysis (FEA). The developed prototype was passed mechanical test by following ISO 10328:2006 for strength verification. Clinical gait analysis was performed from three subjects with right unilateral transtibial amputation to test the performance of the prototype. From the result, we find that our unique ESR prosthetic foot has ability to provide higher energy storage and release during push off than the commercial ESR prosthetic foot.**

I. INTRODUCTION

Historically, many kind of prosthetic feet have been developed. One of the popular type is the energy storage and release (ESR) prosthetic foot. Because its abilities to store energy in form of elastic strain energy of the structure during initial ground contact toward toe off like a leaf spring and then release the energy during toe-off to propel wearer move forward [4]. It also simulate plantarflexion and dorsiflexion angle like a human ankle-foot. It is made from high strength and elasticity material such as carbon-fiber and glass-fiber composite.

Design of ESR prosthetic foot normally use Finite element analysis (FEA) to synthesize foot function [5, 6]. The main point of prosthetic foot design is to synthesize energy release during push off to approach the human ankle-foot's energy release. This approach will provide symmetrical gait between residual and intact limb [2]. Nevertheless, the energy release of the prosthetic cannot be evaluated directly by FEA because of unknown thermal loss, material damping and other loss. Thus, the performance of the ESR prosthetic foot is limited by the fact that it is incapable to release energy than its store [4]. Since FEA cannot evaluate the energy release but the energy is able to be evaluate by experimentation. Clinical gait evaluation is the most popular experiment to evaluate not only the energy storage and release but also kinematic and other kinetic data [7]. According to research of the commercial ESR prosthetic foot, the energy storage during push off is 0.07-0.12 J/kg but energy release is only 0.03-0.07 J/kg and efficiency is about 40-60 % [8, 9]. This energy release is much lower than human ankle-foot (0.13-0.21 J/kg) [2], which is not enough for amputee locomotion. Additionally, most of the commercial ESR prosthetic foot comprises of two leaf spring; those are base and forefoot spring that are attached together [3, 10]. So each spring always receives ground reaction force (GRF) throughout stance phase that will affect to its strength. Consequently, the commercial ESR prosthetic foot was more thickness for relieving the stress but loss of ability to stored energy.

In this paper, the unique ESR prosthetic foot is developed with high energy storage during push-off by using Finite element analysis so as to obtain more energy release compare to the commercial ESR prosthetic foot. Then, the developed prototype is fabricated for mechanical test by following ISO 10328:2006. Finally, clinical gait analysis is performed to evaluate dynamic response of the prototype.

II. METHODOLOGY

A. *Biomechanical of the human ankle-foot system*

During steady-state walking on level ground, human gait cycle comprises of two phase that are stance and swing phase. The stance phase is the period when foot is in contact with the ground. These is known as a stance phase approximately 60% of the gait cycle. The remaining is in the swing phase. The stance phase can be divided into five events; those are heel strike, flat foot, mid stance, heel off and toe off. The flat foot event is the first peak of ground reaction force and moment cause the ankle rotate into maximum plantarflexion angle about 5 degree. Followed by the heel off event that is the second peak of ground reaction force cause the ankle rotate into maximum dorsiflexion angle about 10-12 degree [11]. Energetically, the energy storage during push off is absorbed by muscle

 DOI:10.14416/c.fte.2017.11.075

contraction since mid stance through heel off about 0.1-0.15 J/kg and then release the energy in order to propel the leg into swing phase during heel off through toe off about 0.13-0.21 J/kg [2].

B. Design specification

Study of the biomechanical of the human ankle-foot system was established to formulate our design specification. There are two aspect in design specification that are ankle range of motion and energetic. The design parameter and allowable stress are tabulated as shown in Table 1. The design was based on human's foot size 25 cm and acceptable amputee weight 60-80 kg.

III. MECHANICAL DESIGN

A. Structure design

Since most of the commercial ESR prosthetic combined base and forefoot spring which were attached together. While receiving GRF, each part function all over stance phase that affected its durability and its ability to store energy. If we model each part of the prosthetic foot receive separately GRF during stance phase, the duration time of receiving GRF for each part will be lower than the commercial part. Thus, we modeled the prosthetic which comprised of heel and forefoot spring. The heel spring would receive GRF during heel strike through mid stance and the forefoot spring would receive GRF during remaining of stance phase. The prosthetic foot would respond to GRF on each parts as shown in equation (1).

$$I(t)\ddot{\theta} + \ddot{C}(t)\dot{\theta} + K(t)\theta = M(t) \tag{1}$$

where *I(t), C(t), K(t)* and *M(t)* are moment of inertia, damping coefficient, stiffness and external moment due to GRF about ankle joint.

Since the forefoot spring had to have ability to store highly energy in form of elastic strain energy. This energy depends on deformability of its structure. Moreover, the deformability of structure depends on thickness of the structure. Thick section would get more deformation but less strength than thin section. Therefore, varying thickness concept was employed to achieve more energy storage and remain strength of the structure. The forefoot spring was designed with varying thickness and long arm of moment to obtain more deformation and strength. The forefoot spring (Figure 1) would simulate dorsiflexion of ankle during mid stance through toe off. Straight tip at the top of the forefoot spring was stopper which useful to restrict deformation of below curved lever (Figure 1). The heel spring would simulate plantarflexion of ankle during heel strike through mid stance. Both heel and forefoot spring were made from carbon fiber reinforce epoxy which was very high strength per weight ratio and more flexibility. Additionally, the bumpers which were position on heel pad and stopper of forefoot are modeled so as to reduce shock between each segment of part. Those bumper were made from cast nylon-6 because of high toughness. The proposed prosthetic was established in CATIA software. Size of the prosthetic based on human's foot size 25cm and complied with the requirement of ISO 10328:2006 standard.

Table I. Design parameter of ESR prosthetic

Design parameter	Value
Foot size	25 cm
Allowable amputee's weight	60-80 kg
Max. plantarflexion angle	5°
Max. dorsiflexion angle	10°
Max. energy storage at heel off	> 0.12 J/kg

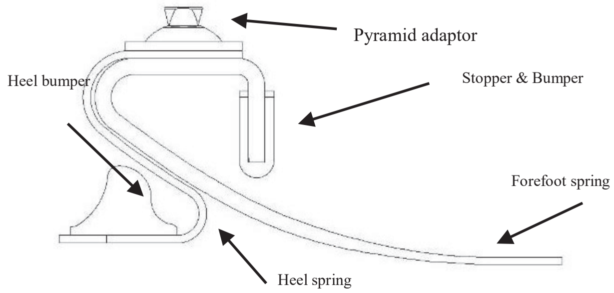


Figure 1. Side view of the proposed prosthetic foot.

B. Finite element analysis

In this section, the function of the proposed prosthetic foot was implemented by trial and error procedure that complied with the design parameter including deformation (ankle angle), stress distribution (von-mises stress) and energy storage in form of elastic strain energy. The procedure was performed by FEA via ANSYS® Workbench that was divided into two main purpose. First, to implement function of the prosthetic during normal walk on level ground. Second, to verify strength of the proposed prosthetic's structure that complied with the ISO 10328:2006 standard.

a) Boundary condition, criterion and FEA modelling

For implementation of function during normal walk, there were two boundary conditions including receiving normal walking load at flat foot and heel off event. We set boundary condition by fixed support on top of the pyramid adaptor. The GRF value of 825 N and 894 N were applied at the center of pressure of flat foot and heel off event, respectively. Each parts were connected together by bonded contact as shown in Figure 2(a) and 2(b). The criterion of the function was complied with design specification as shown in Table I. For strength verification, we complied with ISO 10328:2006 that prescribe procedure for verification of prosthetic's strength. The procedure comprises of static and cyclic test

 DOI:10.14416/c.fte.2017.11.075

that specified loading condition to apply on structure of the prosthetic via platform at heel and forefoot position. The magnitude of testing load depends on loading level (P) or allowable amputee's weight of the prosthetic foot. Since we prescribed about 60-80 kg that was classified into P4 loading level. Static and cyclic test were performed by applying load of 2065 N (static load) and 50-1230 N (sinusoidal load), respectively via the platform at each heel and forefoot position (separately) as shown in Figure 2(c)-2(f). It was noted that FEA based on static simulation. A nonlinear mechanical effect and large deformation solving method were determined. The model employed 418,104 tetrahedron elements by global mesh control with proximity and curvature size function. There are two assumption for solving the problem; those were negligible inertia effect & thermal strain effect and homogenized material properties of carbon fiber composite.

b) Results of FEA

The proposed prosthetic strength was evaluated by von-mises stress value. For the composite parts, the strength depends on a number of fibrous layer. To trial and error for function and strength, the thickness was increased in weakened section to limit stress concentration. In contrast, the thickness was decreased in a section that low stress to increase deformation and elastic strain energy. Finally, the design of the prosthetic foot was able to provide approximately 5 degree of plantarflexion during flat foot as shown in Figure 3(a), 10 degree of dorsiflexion during heel off and 0.145 J/kg of energy storage during heel off as shown in Figure 3(b) that achieved the value of design specification. The maximum stress (von-mises stress) value of 450 MPa that occurred at curve of the forefoot spring during heel off was underneath the ultimate and fatigue strength of the carbon fiber reinforce epoxy composite.

After obtained the specified function, the proposed prosthetic was verified by compliance with ISO 10328:2006 to ensure strength of its structure. Static and cyclic test were analyzed by applying load on heel and forefoot via platform as described above. The result of FEA reported that the maximum stress of the proposed prosthetic was lower than ultimate and fatigue strength of the material for static and cyclic test, respectively for overall loading position as shown in Figure 3. Then we found that the forefoot spring's structure was not affected by testing load at heel loading position as shown in Figure 3(d) and 3(f). Otherwise, forefoot loading not also affected the heel spring as shown in Figure 3(c) and 3(e). This result showed that each spring part of the proposed prosthetic was able to separate stress itself from the other. It was not affect the other spring part.

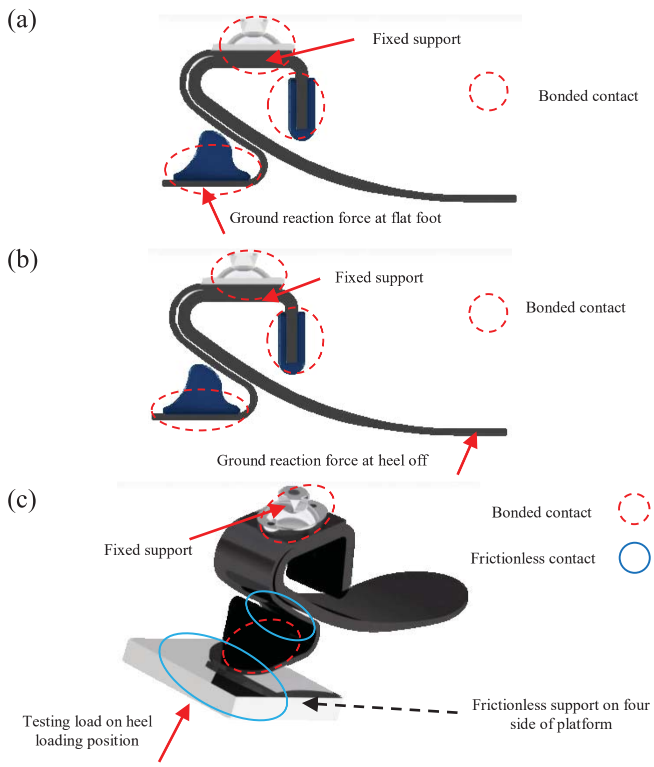


Figure 2. Boundary condition for normal walking at (a) flatfoot and (b) heel off. And boundary condition for strength validation at (c) heel loading position and (d) forefoot loading position.

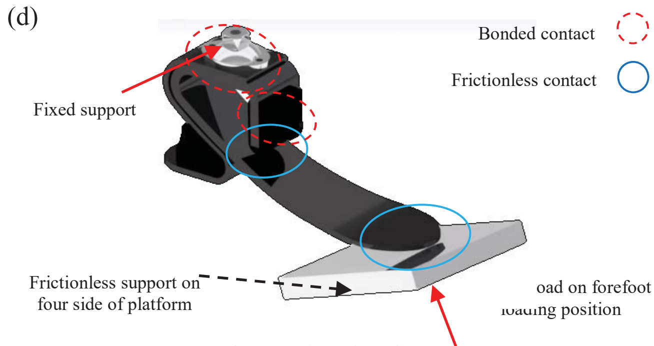
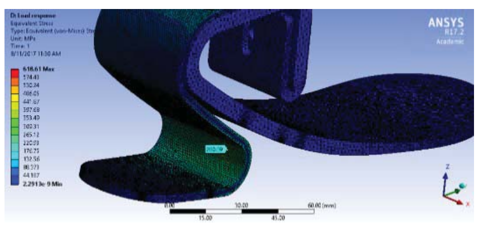
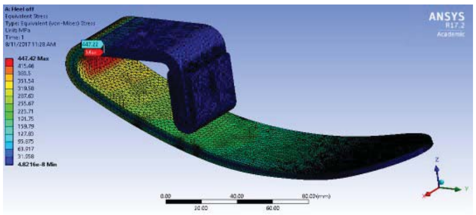


Figure 2. (Continued)



a)

DOI:10.14416/c.fte.2017.11.075



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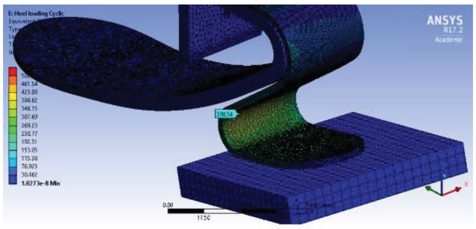

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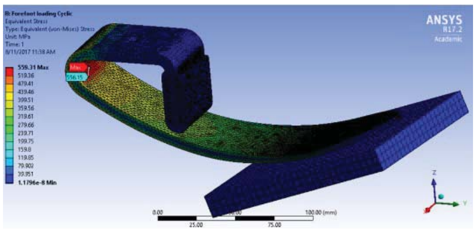

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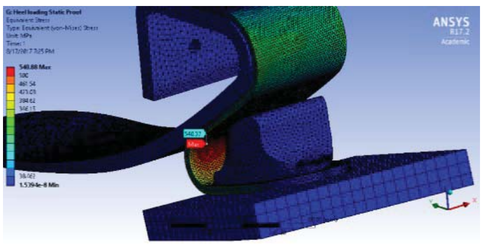
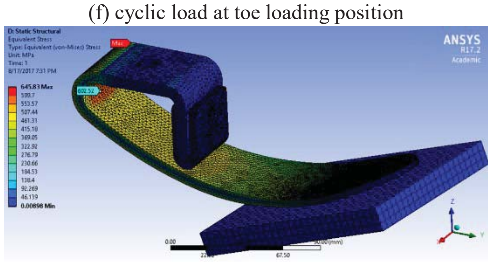


(e)

Figure 3. Results of stress distribution by FEM in case of (a) flat foot event (b) heel off event (c) static load at heel loading position (d) static load at toe loading position (e) cyclic load at heel loading position
(f) cyclic load at toe loading position



(f)

Figure 3. (Continued)

Figure 4. The prosthetic foot prototype.
Fabrication by vacuum bag curing and CNC machining.

C. Developed prosthetic foot prototype

The heel and forefoot spring which were carbon fiber structure was fabricated by laying up on aluminium alloy mold and vacuum bag curing process. The process was performed by autoclave that controlled pressure at 6 bar, temperature at 110 ◦C for 90 minutes of curing time. The connector and stopper part which was make by stainless steel 304 and cast nylon-6 were manufactured by CNC machine. The assembly all of the parts as shown in Figure 4.

IV. MECHANICAL TEST

In order to verify the strength of the developed prototype's and security of user for ethic statement. The developed prototype must comply with the requirement of the ISO 10328:2006 standard. Thus the developed prototype must be subjected to test loading condition according to the standard. The standard testing procedure was divided into two loading condition that are static and cycle test. The test were performed by applying P4 loading level according to the allowable amputee's weight 60-80 kg of the developed prototype.

A. Static test

The static test is divided into two loading position that are heel and forefoot loading position. We performed the test by using universal test machine (Instron, ElectroPuls

DOI:10.14416/c.fte.2017.11.075

model E10000) to apply testing load. The testing load was measured by Dynamic load cell (10 kN, Dynacell 2527-202 series) which was located under platform. The developed prototype was prepared and aligned to platform according to the ISO specification as shown in Figure 5(a) and 5(b). Both heel and forefoot loading were applied with a force of 2065 N at a ramping rate 100N/s from zero. After reaching 2065 N, the force is maintained for 30±3 seconds and then decreased to zero.

After tested, we recognize that the prototype was able to receive testing load without failure on both heel and forefoot loading position. The testing load-displacement versus time graph as shown in Figure 6(a).

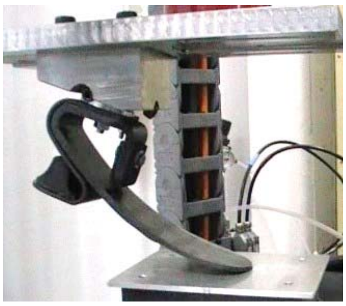
B. Cyclic test

In this test, the developed prototype was applied sinusoidal replicated load on heel and forefoot. This test method was performed by Artificial Limb Testing Machines (Si-Phan Electronics Research Ltd.) at Sirindhorn National Medical Rehabilitation Institute. The prototype was prepared and aligned to platform according to the ISO specification as shown in Figure 5(c). The test load was applied to the prototype with a sinusoidal wave force range 50-1230 N, 1 Hz for two million cycles.

After two million cycles passed, we summarized that the prototype was able to pass the test because its structure had no failure. The cyclic test data include testing load and displacement profile and number of cycle passed as shown in Figure 6(b).

(a)

(b)



(c)

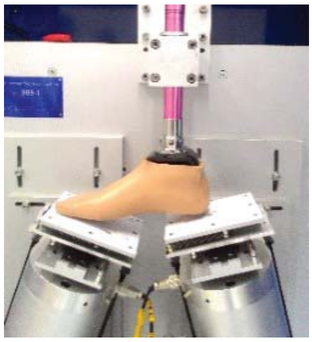
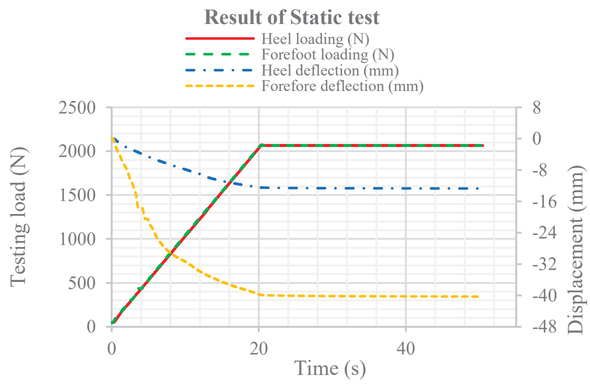
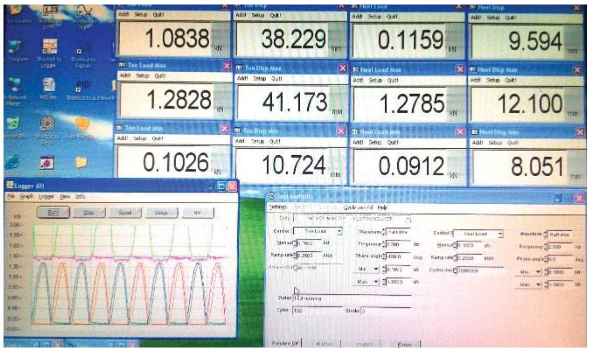


Figure 5. Static test on (a) heel loading position and (b) forefoot loading position were performed by universal test machine. (c) Cyclic test was performed by Artificial Limb Testing Machines.



(a)



(b)

Figure 6. Results of Mechanical test include (a) Testing load-Displacement versus Time graph for Static test and (b) Cyclic test data from Artificial Limb Testing Machines at Sirindhorn National Medical Rehabilitation Institute.

V. CLINICAL GAIT EVALUATION

Since energy release of ESR prosthetic foot could not be evaluated directly by FEA. However, experimentation was able to evaluate these energy. The clinical gait analysis was useful experiment to investigate dynamic motion of body segment i.e. angle, force, moment and power of ankle, knee and hip joint. The analysis of human gait were solved by using inverse dynamic concept. The

 DOI:10.14416/c.fte.2017.11.075

obtained results could be useful to research and develop of the prosthetic foot.

A. Subject

Three healthy subjects (male, ages 18-60 year old, weight 60-80 kg, foot length 25 cm) with right unilateral transtibial amputation participanted to study. The subjects had experienced wearing the prosthetic foot more than 2 years and could ambulate without gait aid.

B. Instrument

The dynamic motion of subjects were captured by Motion Analysis system at Siriraj hostipal including eight motion capture camera (200 Hz; Motion Analysis Corp, Santa Rosa, CA, USA) as shown in Figure 7(a), two force platform (2000 Hz; AMTI, Watertown, MA, USA) which located centrally in a walk way as shown in Figure 7(b) and twenty nine retro reflection markers (12 mm diameter) were positioned on segments of the upper and lower extremities as shown in Figure 7(c).

(a)

(b)

(c)

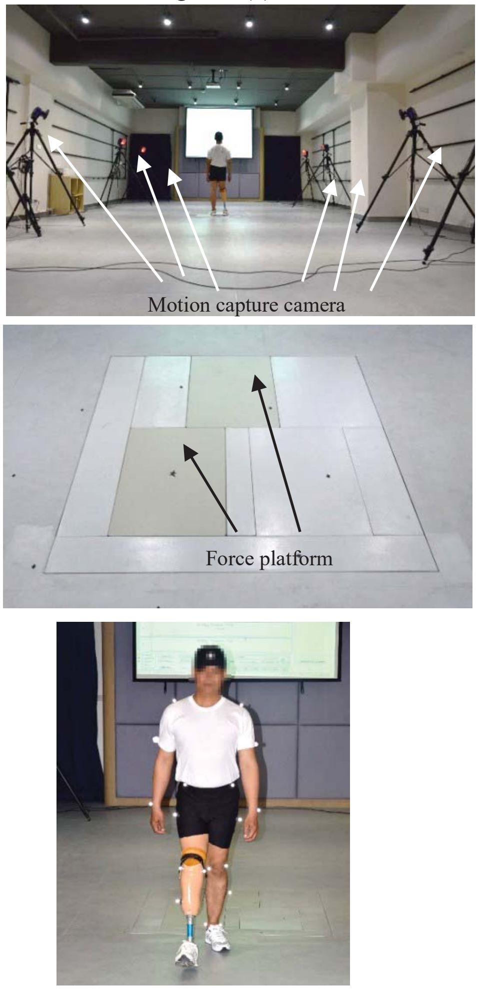


Figure 7. Clinical gait analysis at gait motion analysis in Siriraj hostipal. In the laboratory include (a) motion capture camera, (b) force platform and (c) the retro-reflective marker which were positon on the subjects.

C. Experiment protocol

After providing the subjects, they were verified alignment to fit the prototype by prosthetist and orthoptist. Then the subjects have training period for adapting on each prosthetic model for 15 minutes. The experiment was started by asking each subjects to walk with self-selected speed along a walkway for 30 minutes. The data including kinematic of the reflective marker and GRF were collected when the subjects stepped on the force platforms. Finally, the data was computed using inverse dynamic technique.

D. Result and discussion

The result showed that the average self-selected gait speed of subjects was 1.12 m/s. The ankle rotation of each subject with self-selected gait speed for each subject as shown in Figure 8. The data reported that the developed prototype could provide maximum plantarflexion angle during flat foot about 4.8-6 degree (average of 5.4 degree) and maximum dorsiflexion angle during heel off about 11-12.8 degree (average value of 11.9 degree) that achieve the design specification.

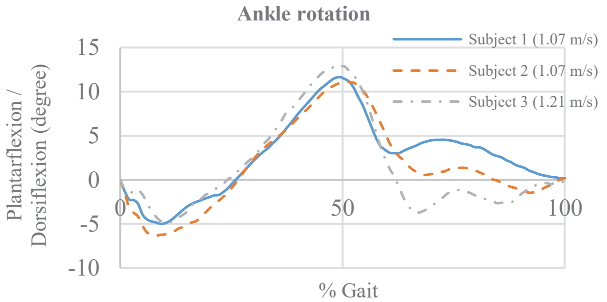
The result of ankle power for each subject with self-selected gait speed, as shown in Figure 8, reported that the developed prototype could generate peak power about 0.85-1.15 W/kg during heel off. From the power's result, the energy storage and release of the prosthetics could be calculated from area under the power and time graph the power. The calculated data showed that the average energy storage was 0.1504 J/kg during mid stance through heel off and the average energy release of the prototype was 0.0877 J/kg during heel off through toe off. The calculated data of each subject was tabulated in Table II.

From the results, both energy storage and release of the developed prototype were significant higher than the commercial ESR prosthetic foot (0.07-0.12 J/kg and 0.03-0.07 J/kg) respectively [8, 9]. But its energy release was also lower than human ankle-foot of 0.13-0.21 J/kg [2]. We only considered subject 1 and 3 because of abnormal gait of subject 2. Each subject had difference self-selected gait speed. We found that high energy storage and release of the developed prototype significantly depend on how fast of the subject walk. Since the receiving GRF of the structure increased as gait speed increased. Consequently, the structure of the developed prototype obtained more deformation as same as its energy storage and release and ankle rotation. However, the maximum plantarflexion and dorsiflexion during flat foot and heel off respectively were slightly higher than the design specified value because the implementation of the developed prototype by FEA based on negligible inertia effect assumption. Thus, the deformation that was evaluated by FEA was slightly lower than the experiment.

 DOI:10.14416/c.fte.2017.11.075

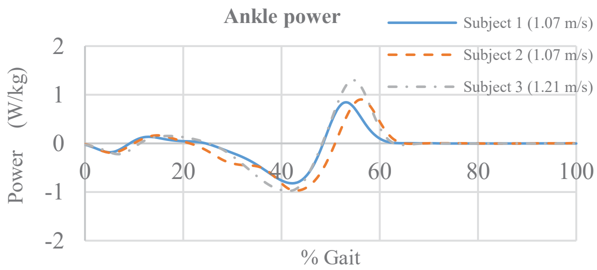
VI. CONCLUSION AND FUTURE WORK

The unique ESR prosthetic foot has been successfully developed which comprises of two leaf springs; those are heel and forefoot spring. Each part would function separately in early and terminal stance phase. Thus, the duration time of each part was decreased and relieving stress. The developed prototype has ability to store energy of 0.15 J/kg and release energy of 0.088 J/kg during push off. This energy release of the developed prototype is higher than the commercial ESR prosthetic foot about 25 % that achieved the objective of this research.



(a)

Figure 8. Result of clinical gait evaluation include (a) ankle rotation and (b) ankle power of each subject (self-selected gait speed)



(b)

Figure 8. (Continued)

Table II. The energy storage & release and efficiency of the prosthetic feet.

	Energy storage (J/kg)	Energy release (J/kg)	Efficiency (%)
Subject 1	0.1265	0.0731	57.8%
Subject 2	0.1790	0.0758	42.3%
Subject 3	0.1457	0.1142	78.4%
Average	0.1504	0.0877	59.5%

VII. ACKNOWLEDGEMENT

This research received financial supports from Chulalongkorn University through Chulalongkorn Academic Advancement into Its 2nd Century Project (Smart Medical Device).

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DOI:10.14416/c.fte.2017.11.075