Optimization Design of a Carbon Fibre Prosthetic Foot for Amputee

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Abstract

Compliant prosthetic foot is a popular term for all passive prosthesis that are currently preferable by amputee due to their advantage of energy saving, light weight, simple structure and easier usage over the powered counterparts. The key characteristic of this device relies on the elastic deformation of the structure to support its functions. In order to fulfil this property, design and material of the devices are the two most concerns. This paper describes an optimization design method for a prosthetic foot made from carbon fibre. Finite element analyses are carried out to evaluate the mechanical behaviours of the design obtained by the optimization procedure. Prototypes of the device are also fabricated and tested. The characteristics of the optimum prosthetic foot predicted by theory are not only quantitatively verified by experiments but also qualitatively surveyed by trials from the amputee. Results show that the fabricated prototype is capable of bending and compression to store elastic energy and release it to assist in forward propulsion of the body. These characteristics make it more versatile and reduce the impact force on residual limbs.

Keywords: Prosthetic; Compliant Mechanism; Composites; Carbon Fibre; Optimization

Abbreviations

ESAR: The Energy Storage and Return Foot; CTM: Constant-Torque Mechanism; CFP: Carbon-Fibre Prosthesis

Introduction

Function loss of amputees due to residual limbs usually induces an abnormal gait. Therefore, a prosthesis is expected to fully replicate the functions of the natural organ it replaces. While active prosthetics with an internal power sources are still in the stage of development [1-3], passive devices are widely used in the market. Normally, disable people were prescribed solid ankle cushioned heels, which mostly focused on reducing impact loading at heel strike, but stores and releases very little elastic energy [4]. The energy storage and return (ESAR) foot is an alternative which has the potential to help improve gait [5]. Despite the variety of

available commercial products in the market, it is still intricate to make them as flexible as the intact feet.

Lee., *et al.* [6] developed a multi-axis prosthetic ankle joint to mimic as many degree-of-freedom as possible of a human ankle. However, this prosthesis still can store very little elastic energy compared to the usage of large deflection compliant beams. Phan., *et al.* [7] designed a rehabilitation device for human mobility-assisting using a compliant constant-torque mechanism (CTM). Their design exploited the benefit of compliant mechanisms by simplifying the device with elastic curve beams in a monolithic form. No sensorized control systems and actuators are required in assisting the rehabilitation of human joint with this device. POM material was proposed to be used for the design and the prototypes of their devices can be fabricated using CNC milling process. The monolithic

nature of this CTM would be commercial feasibility if the manufacturing cost concern are addressed. Mass production with injection molding can resolve this issue instead of using CNC milling for every single piece.

Light and durable materials such as carbon fibre/epoxy composites are popular and ideal materials for orthopaedic prosthetics. Generally, fibre composites can be made by either long or short fibre production processes. While short fibre composites production relies on injection molding machines, composites from carbon-fibre fabric can be made manually with simple manufacturing process. The key characteristic of composite prosthetics is the fact that they can be custom-made to match the height, weight, and muscular structure of the individual. This paper proposes the design and fabrication process of a new carbon-fibre prosthesis (CFP). The shape and stiffness of the device are optimized to facilitate the mobility and functionality of lower-limb amputees. Finite element analyses are carried out to evaluate the biomechanical behaviours of the design obtained by an optimization procedure. Prototypes of the device are fabricated using carbon/polyester composites. Experiments are performed to demonstrate the effectiveness of the device.

Materials and Methods

Operational principle

Ideal prostheses are supposed to assist patients in achieving natural gait motion which includes: dorsiflexion, eversion, impact absorption, energy return and ankle torsion. The fully compliant prosthetic ankle of Pham, *et al.* [8] fulfilled above requirements. However custom-made product to meet individual needs is a matter of cost. Figure 1 illustrates the schematic of the prosthesis in this paper which consists of a base plate, a keel plate and a rubber wedge. The rubber wedge is attached to the base plate but it is separated to the keel plate. Both base plate and keel plate are designed as simple as possible to adapt to the carbon fiber composites fabrication process. The combination of these parts has the function of shock absorption, foot flexion and strain energy storage and release.

A typical gait cycle of a sound foot is composed of two phases: stance phase and swing phase [8]. A proper prosthesis should have the capacity to store energy in order to replicate the motion of an intact foot. The design of a passive prosthetic foot mostly focused on the stance phase. When the foot hits the ground, owing to the flexibility of the heel or the rubber wedge in this design it absorbs

the shock induced by the terrain (heel-strike). This passive device continues to deflect and stores elastic energy when the patient stands completely on one leg (foot-flat). A dorsiflexion in the subdivision of stance phase releases part of the energy in the heel and transfers partly to the keel. The foot accumulates mechanical energy in elastic structures in early to mid-stance and reverts it to the amputees to propel their body forward in late stance (toe-off).

Figure 1: Schematics of the CFP.

Design formulation

An appropriate prosthetic design should ensure for a natural gait. It is expected to have the capability of reducing the impact due to the reaction force, enabling ankle rotation flexibility during movement and allowing walking on undulating and rough terrains. The CFP in this research includes two flexural beams and an elastic rubber wedge. The key component of the design is a cubic Bezier curved beam functioned as the keel whereas its curvature and thickness will define the global stiffness of the prosthesis. This stiffness could be modulated to accommodate individual needs by varying the shape of the curve beam or the corresponding Bezier control points in an optimization procedure. Generally, an n^{th} order Bézier curve is determined by $(n+1)$ control points K_i as shown in figure 2:

$$P(v) = \sum_{i=0}^n K_i B_{i,n}(v) \quad (1)$$

$$B_{i,n}(v) = C_n^i v^i (1-v)^{n-i} \quad (0 \leq v \leq 1) \quad (2)$$

$$C_n^i = \frac{n!}{i!(n-i)!} \quad i = 0, \dots, n \quad (3)$$

where $B_{i,n}(v)$ is Bernstein polynomial, v is independent parameter, $P(v)$ and K_i denotes position vectors of points on the curve and control points, respectively.

Figure 2: Design parameters.

The fixed positions of two points K_0 and K_3 are constrained in a design space (100 x 200) mm. Coordinates of the remaining two control points $K_1(x, y)$ and $K_2(x, y)$ are design variables. The thickness t_1 and t_2 of the keel beam and the base beam are also design variables. During the calculation process, three loading cases in figure 3 are applied to the CFP which resemble the three phases in a walking cycle: the heel strike, the foot flat and the toe-off, respectively.

Figure 3: Three loading phases during the walking cycle: (a) Heel strike, (b) Foot flat, (c) Toe-off.

In this research, dry carbon-fiber fabric and polyester resin are used to make the composite material by using hand layup method. Firstly, the resin is mixed with catalyst and hardener additive and stirred thoroughly by hand. This compound is then vacuumed to eliminate completely all remaining bubbles. Then dry fabrics are placed into the mold layer by layer with resin mixture addition following each layer. The expected thickness of the product will define the number of fabric layers required. The final curing step with pressure applied via a vacuum bagging is maintained at room temperature for 45 minutes (Figure 4a). A fabricated carbon/polyester composite sample is shown in the inset of figure 4a.

To determine the flexural properties of the fiber-reinforced plastic composite, test specimens are prepared following the British Standard BS EN ISO 14125:1998. The dimensions of specimens for three-point bending test are (100 x 15 x 2) mm. A tensile testing machine Instron 3367 (USA) is used to implement the experiment as shown in figure 4b. The Young's modulus, Poisson's ratio and tensile strength of the specimens are found as 16.0GPa, 0.39 and 243MPa, respectively.

Figure 4: Fabricated sample preparation (a) and the three-point bending test experiment (b).

Optimization

The deformation analysis of the CFP is a large deflection problem. Appropriate stiffness of the structure according to each phase of the gait is searched by using an optimization scheme. An effective shape optimization procedure using genetic algorithm for the design of compliant prostheses was developed in our previous researches [9]. In the optimization process, the structure, constraints on the design variables, number of generations, and size of populations are specified.

Due to the large deflection and nonlinearity problem, the relation between applied force and displacement is calculated by using finite element analysis. A commercial software ANSYS [10] is used to calculate for the deflection of the structure. In order to make the model simple and easy for the optimization, beam element is used.

Boundary conditions are set for a person weight from 75 kg to 80 kg; 1.8m in height and walking with a speed of 1.25 m/s [11]. Optimization problem for the CFP shape design is formulated in table 1. In this optimization process, the number of generations and the size of populations are set to 50 and 20, respectively. Optimum numerical results of the design are described in table 2 and the CFP's shape is plotted in figure 5.

Table 1: Optimization formulation of the CFP.

1. Objective:	
• $Min K_{fem} - K_e $	(4)
• $Min(\sigma_{max})$	(5)
2. Design variables:	
• Control points: $K_1(x, y), K_2(x, y)$	
• Beam thickness: t_1, t_2	
3. Constraints:	
• Design space: (200 x 100) mm	
• Beam thickness: $2.0 \leq t_i \leq 8.0\text{mm}$	
• The maximum stress: $\sigma_{max} < \sigma_y / SF$	

Table 2: Value of optimum design variables.

Variables	Value (mm)	Variables	Value (mm)
$K_0(x, y)$	(0, 0)	t_1	5.0
$K_1(x, y)$	(-160, 0)	t_2	2.0
$K_2(x, y)$	(-173, 42)	Width	42.0
$K_3(x, y)$	(-195.1, 95.6)

Figure 5: Optimum shape of the CFP.

Fabrication and testing

In order to realize the optimum design of the CFP for practical investigation, prototypes of the device are fabricated. Firstly, a mold whose cavities encapsulate the shape of the beams is prepared as shown in figure 6a. The top surface is the cavity of the keel

and the bottom surface is the cavity for the base plate. Following the hand layup fabrication process described in the section design formulation, a prototype is fabricated as shown in figure 6b. Figure 6c is the CFP when it is assembled with a socket for a lower-limb amputee.

Figure 6: Fabricated mold and product.

The CFP was mechanically tested using a uniaxial tensile testing machine (Instron 3367, USA) (Figure 7). It was clamped to the base of the machine and loaded by a flat plate attached to a load cell follow the ISO 10328 standard for structural testing of lower-limb prostheses. It was axially loaded in three different configurations: heel-only contact (b), foot-flat (c) and toe-only contact (d).

Figure 7: Mechanical testing (a) of the CFP during (b) heel-only, (c) foot-flat and (d) toe-only.

Results and Discussion

Using the experimental setup in section fabrication and testing, the stiffness of the CFP at three different loading configurations are demonstrated. Their corresponding experimental force - displacement (f - d) curves are also obtained. Figure 8-10 show the measurement results for each loading configuration. The calculated stiffness is also presented in these figures for comparison. It can

be seen that only the foot-flat loading in figure 8 shows a slightly proportionality between force and displacement with the regression stiffness of simulation and experiment are 34.1 (N/mm) and 41.1 (N/mm), respectively. The difference between calculated and practical results can be attributed to the uncertainties in material, geometry and loading conditions of the experiments. Any slightly change in the resin mixture as well as curing treatment during the fibre composite fabrication will affect its mechanical properties. The heel-strike and toe-off loading curves in figure 9 and 10 are completely nonlinear but they comparatively follow the simulated results. The contact of the flat plate of the testing machine with the surfaces of the prosthesis in all loading cases are not fixed, where sliding may occur, and the alignment of the load cell with the symmetry plane of the CFP may not be perfect during the experiments. In a nutshell, the results show rather good agreement between simulation and measurement.

Figure 8: Force-displacement curve for foot-flat loading.

Figure 9: Force-displacement curve for heel-strike loading.

Figure 10: Force-displacement curve for toe-off loading.

Conclusion

In this research, the design methodology of a new carbon/polyester composite prosthesis has been proposed. Taking advantage of fully compliant mechanisms, without any movable joints, the use of the designed device would result in reduced wear; no need for maintenance and enhanced performance by increasing precision. In order to evaluate the effectiveness of the device, prototypes of the device are fabricated using a simple hand layup composite production process. They are also validated by experiments. The combination of choosing appropriate material and using compliant mechanisms further enhances the capability of storing elastic energy in early to mid-stance of the prosthesis and releasing it to assist the amputees in forward propulsion in late stance. The optimization methodology using in this research could be incorporated to forward dynamics simulations [12] of amputee walking and replace fixed setting component stiffness by a proper metabolic cost and joint loading objective functions [13] in order to account for practical feelings of amputees.

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Conflict of Interest

None.

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