Sleeve Muscle Actuator and Its Application in Transtibial Prostheses

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***Abstract*—This paper describes the concept of a new sleeve muscle actuator, and a transtibial prosthesis design powered by this novel actuator. Inspired by the functioning mechanism of the traditional pneumatic muscle actuator, the sleeve muscle actuator incorporates a cylindrical insert to the center of the pneumatic muscle, which eliminates the central portion of the internal volume. As a result of this change, the sleeve muscle provides multiple advantages over the traditional pneumatic muscle, including the increased force capacity over the entire range of motion, reduced energy consumption, and faster dynamic response. Furthermore, utilizing the load-bearing tube as the insert, the sleeve muscle enables an innovative "actuation-load bearing" structure, which has a potential of generating a highly compact actuation system suitable for prosthetic use. Utilizing this new actuator, the preliminary design of a transtibial prosthesis is presented, which is able to provide sufficient torque output and range of motion for a 75 Kg amputee user in level walking.**

Keywords—actuator, artificial muscle, transtibial prosthesis.

I. Introduction

For a lower-extremity prosthesis, the primary purpose is to restore the locomotive functions of the lost biological limb sections and joints. Historically, such functions have been restored by energetically passive devices, i.e., devices that only dissipate energy, or store and reuse energy within a gait cycle. The passive nature of such devices is fundamentally different from the energetic role of the corresponding biological joints, and thus poses a significant limitation to their functionality and rehabilitation effects. For example, biomechanical studies on human locomotion highlight the important energetic role of the ankle joint. In level walking, the ankle produces substantially more work than the knee and hip [1]. Unlike the knee, the energetic behavior in level walking is clearly and significantly positive (i.e, integration over a cycle of power data is clearly and significantly positive) [2]. As such, for an amputee fitted with passive transtibial prosthesis, he or she has to expend more power on the unaffected biological joints to compensate for the lack of power generation in the prosthetic ankle, resulting in an asymmetric gait and greater energy consumption [3].

To address this important issue, a considerable amount of research has been conducted on the development of energetically active transtibial prostheses with powered ankle joint. In such efforts, the primary challenge is how to generate sufficient power and torque output within a compact form factor. In the existing works, the major technical approach is the electromagnetic actuator (i.e., DC motor) in combination with electrochemical batteries, and the typical works adopting this approach include the powered ankle-foot prostheses developed by the Biomechatronics group at MIT [4,5] and the powered transfemoral prostheses developed by the Center for Intelligent Mechatronics at Vanderbilt University (which include powered ankle joints) [6,7]. In spite of the improved gait quality provided by these active devices, they also tend to suffer from multiple inherent weaknesses of the electric actuation system, primarily the heavy weight of the actuator and the short battery life that limits the duration of operation.

Unlike the aforementioned works, the research presented in this paper takes a different technical route to address this challenging issue. Instead of the electric actuation system, the transtibial prosthesis design in this paper utilizes a new sleeve muscle actuator, which is an advanced form of the traditional pneumatic muscle actuator. The traditional pneumatic muscle, also known as the McKibben muscle or fluidic muscle, simulates the functioning mechanism of biological muscles through pressurizing an air-tight elastic tube surrounded by an inextensible mesh (Fig. 1). When the muscle is inflated, the tube expands in the radial direction and shortens in the axial direction, exerting a pulling force to the external load. With this unique structure, the pneumatic muscle enjoys multiple advantages (high power density, similar elastic characteristics to biological muscles, etc). Combined with a compact pneumatic supply, such as the liquid-propellant-based supply device [8], the resultant actuation system poses a competitive choice for the actuation of prosthetic devices. Specifically for the actuation of transtibial prostheses, Klute et al. proposed the concept of pneumatic muscle-actuated transtibital prosthesis, and expected high torque output (~110Nm) and sufficient range of motion for the powered ankle joint [9-11]. Recently, Versluys et al. proposed a pneumatic muscle-actuated BK prosthesis, which utilizes three pneumatic muscle actuators (one anterior and two posterior) to drive the ankle motion [12].

In spite of these early attempts, the application of pneumatic muscle actuation in transtibial prosthesis design still faces a major challenge, which is the radial expansion of the muscle actuator. To avoid the interference with the load-bearing structure of the prosthesis, generous spacing has to be

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provided, causing the expansion of the prosthesis volumetric profile. As such, the existing designs are all too bulky for practical daily use. The new design in this paper, on the other hand, using a novel sleeve muscle actuator design, which not only generates significant performance improvement, but also provides a highly compact package with an integrated “actuation-load bearing” structure. The basic concept of the sleeve muscle actuator is presented in the subsequent section, followed by the new transtibial prosthesis design actuated by this new actuator.

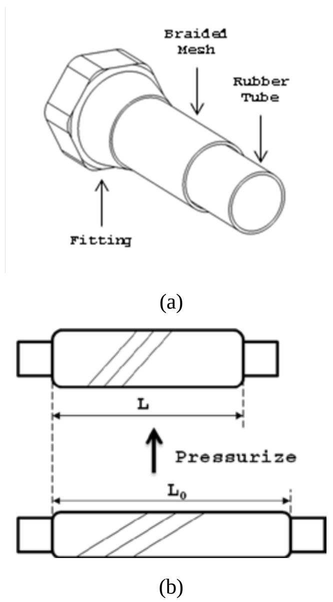


Fig. 1. Structure (a) and functioning mechanism (b) of pneumatic artificial muscle.

II. SLEEVE MUSCLE ACTUATOR

The basic idea of sleeve muscle actuator is inspired by the working principle of the traditional pneumatic muscle. Specifically, pneumatic muscle generates power output through contraction under internal gas pressure, and in this process, the internal volume also increases as a result of the radial expansion of the muscle structure. Applying the principle of virtual work, the following equation for the actuation force can be obtained [13]:

$$F = \left(-dV/dL\right)\cdot\left(P - P_{atm}\right) \tag{1}$$

where F is the actuator contraction force, V is the actuator internal volume, L the length of the elastic part of the muscle, P is the actuator internal pressure, and P_{atm} is the atmosphere pressure. As indicated by this equation, the contraction force F is proportional to the gauge pressure $\left(P - P_{atm}\right)$, with the coefficient of proportionality as $\left(-dV/dL\right)$. Therefore, in order for the pneumatic muscle to generate a contraction force output, the internal volume V needs to expand when the muscle length L shortens, resulting in a positive value of $\left(-dV/dL\right)$. To gain further insight into the actuation process, the internal volume V can be divided into two parts, including V_1, a cylindrical volume at the center of the muscle, whose diameter is equal to that of the muscle’ end connectors; and V_2, a ring-shaped volume surrounding V_1 (Fig. 2a).

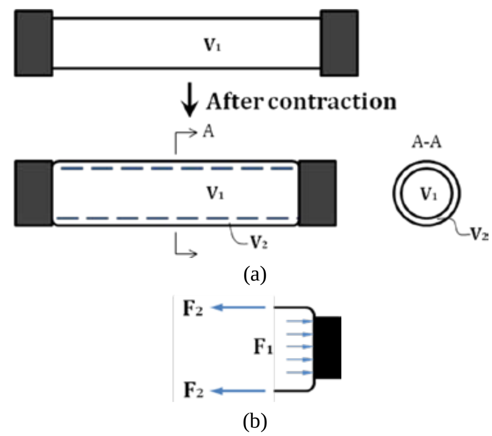


Fig. 2. Analysis of the pneumatic muscle functioning mechanism: (a) dividing the internal volume into two parts; (b) the corresponding contributions to the contraction force (F_1: pushing force due to the internal pressure; F_2: pulling force by the membrane).

Note that V_2 is negligible at the equilibrium state. Both V_1 and V_2 changes with the length of the actuator, and thus the total contraction force can be expressed as the sum of F_1 and F_2:

$$F = F_1 + F_2 \tag{2}$$

where F_1 and F_2 are the contributions of the changes of V_1 and V_2, respectively:

$$F_1 = \left(-\frac{dV_1}{dL}\right)\cdot\left(P - P_{atm}\right) \tag{3}$$

$$F_2 = \left(-\frac{dV_2}{dL}\right)\cdot\left(P - P_{atm}\right) \tag{4}$$

To further analyze the contributions of these volume changes, V_1 can be expressed as the product of the fixed cross-sectional area $D^2/4$ (D is the diameter of the muscle end connector) and the muscle length L, and thus decreases with the shortening of the muscle. Consequently, the contribution of V_1 to the contraction force is always negative, as indicated by the following equation:

$$\frac{dV_1}{dL} = \frac{1}{4}D^2 > 0 \Rightarrow F_1 = \left(-\frac{dV_1}{dL}\right)\cdot\left(P - P_{atm}\right) = -A_C\left(P - P_{atm}\right) < 0 \tag{5}$$

On the other hand, V_2 expands with the shortening of the pneumatic muscle, and thus contributes positively to the generation of the output force:

$$\frac{dV_2}{dL} < 0 \Rightarrow F_2 = \left(-\frac{dV_2}{dL}\right)\cdot\left(P - P_{atm}\right) > 0 \tag{6}$$

As such, it can be concluded that the total contraction force F is less than F_2, due to the fact that the negative F_1 actually reduces the force output. For a better understanding of the physical nature, F_2 is essentially the pulling force generated by the membrane, while F_1 is the pushing force applied to the end connector due to the internal air pressure (Fig. 2b). As such, to generate a pulling force to the external load, the membrane has to overcome the pushing force first, resulting in a loss of the actuator force capacity. This conclusion suggests that the total contraction force output F can be increased by eliminating the volume V_1 (and the corresponding negative force contribution F_1). Inspired by this conclusion, a new sleeve muscle actuator is developed, in which a rigid cylinder or tube is inserted into the center of the muscle and rigidly connected to the fixed end, while the moving end of the muscle slides on the outer surface of the tube (Fig. 3).

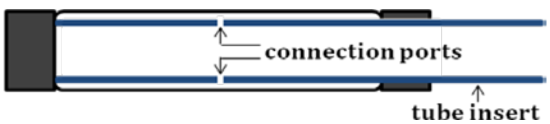


Fig. 3. Schematic of the sleeve muscle actuator.

With this new structure, the volume V_1 is largely eliminated, generating an increase in the output force over the entire range of motion:

$$\Delta F = -F_1 = \frac{1}{4}\pi D_I{}^2\left(P - P_{atm}\right) \tag{7}$$

where D_I is the outer diameter of the insert, which needs to be slightly less than the muscle end-connector diameter D to ensure the structural strength of the moving end. From Eq. (7), it can be deduced that, under a certain internal pressure, the magnitude of force increase is a constant, i.e., not affected by the shortening of the muscle. As such, this effect is especially important in the large-contraction region, where the contraction force is much less than that at the equilibrium state.

In addition to the greater force capacity, the sleeve muscle displays two significant advantages over the traditional pneumatic muscle, including the faster dynamic response and lower energy consumption in operation. These can be largely attributed to the elimination of the space at the center of the muscle, and the resulting reduction of the total internal volume in the muscle. As demonstrated in the experiments, a sleeve muscle prototype displayed a faster dynamic response, and significantly lower energy consumption (20% to 37%), in comparison with the identical traditional pneumatic muscle (without the insert) [14]. As indicated by these results, the sleeve muscle provides a substantially improved performance, and thus represents a significant step forward from the traditional pneumatic muscle.

III. Preliminary Design of the Transtibial Prosthesis

In addition to the performance improvement, the sleeve muscle also provides a potential for the integration of the actuator and the load-bearing structure. Specifically, the cylindrical insert in the sleeve muscle can be used to serve the dual roles of the structural insert of the sleeve muscle as well as the load-bearing tube (i.e., the robotic counterpart of the bone). As such, the muscle actuator completely surrounds the load-bearing tube (i.e., the insert), with the moving end of the actuator sliding on the surface of the tube. This unique integrated “actuation-load bearing” structure forms a highly compact robotic limb design, which conceptually mimics the biological muscle-bone structure while avoiding the excessive complexity of utilizing multiple muscle actuators in the design.

Note that a sleeve muscle is still a single-acting actuator, which only generates a pulling force to the external load. To provide bi-directional actuation for a robotic joint, two possible solutions can be adopted: (1) Utilizing two sleeve muscle actuators for a single joint, with each occupying the opposing limb sections with respect to the joint, or (2) Utilizing a torsional spring to provide the torque in one direction and the sleeve muscle to provide the torque in the opposite direction. Here the second solution is favorable because of its simplicity in the configuration and implementation. Furthermore, for the transtibial prosthesis design, the torque requirement for the locomotion is highly asymmetric, with the plantar-flexive torque dominating the dorsi-flexive torque [2]. As such, if the torsional spring is used in the direction with low torque requirement, its weight and dimensions can be minimized, and thus has a potential to be integrated into the rotary joint itself. This will result in a significant reduction in the volumetric profile of the system.

Based on the analysis above, the schematic of a sleeve muscle-actuated transtibial prosthesis is determined, as shown in Fig. 4. In this design, the insert of sleeve muscle actuator also functions as the load-bearing structure between the standard prosthetic connector and the ankle joint. As such, the expansion of the muscle actuator is unobstructed, completely eliminating the interference issue while maintaining the small diameter of the device. As mentioned above, the plantarflexion requires a significantly higher torque compared with the dorsiflexion, and thus is driven by the sleeve actuator through a cable-pulley mechanism. Also, a torsional spring is incorporated into the joint design, providing the required torque for the dorsiflexion of the ankle.

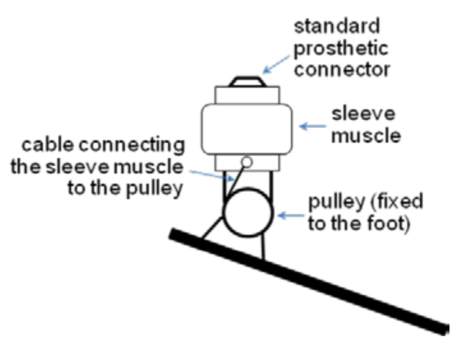
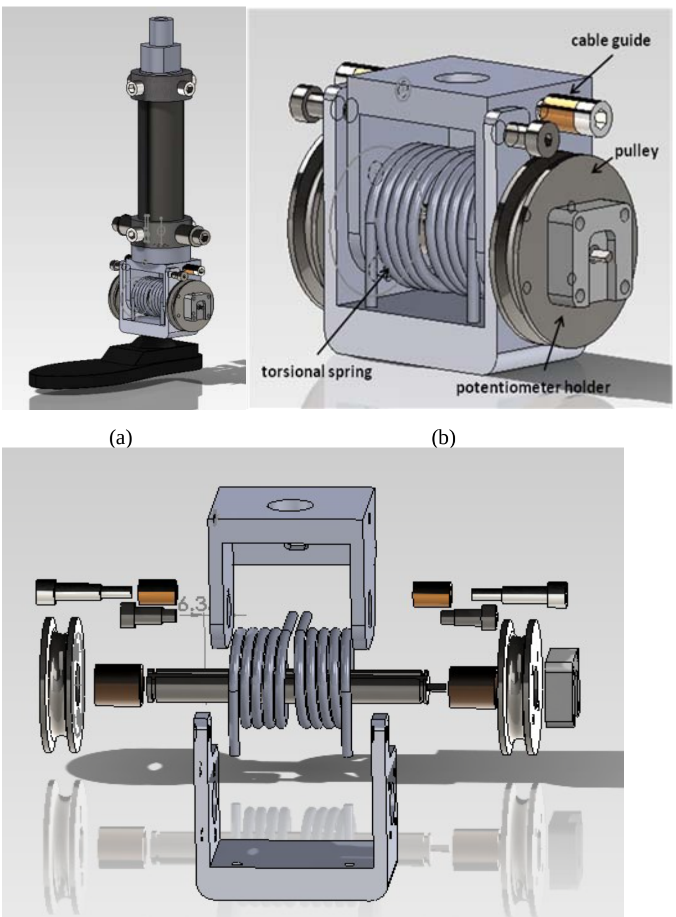


Fig. 4. Schematic of the sleeve muscle-actuated transtibial prosthesis.

A solid model of the preliminary design is shown in Fig. 5. To reduce the risk, a commercial pneumatic muscle from FESTO (Fluidic Muscle DMSP-40-140N-RM-CM) will be modified to incorporate the cylindrical insert at the center and function as the sleeve muscle actuator. In the design, the primary goal is to provide sufficient torque and joint range of motion (~45˚) for a 75 Kg amputee to support his/her level walking. According to the biomechanical data by Winter [2], the peak torque is approximately 115 Nm at the joint angle of 8˚. Based on the design objective, a standard 1-1/4-inch pulley is selected to meet this torque requirement, and also provide the desired range of motion. In the detailed joint design (Fig. 5b), two symmetric torsional springs (model 9271K598 from McMaster-Carr) are incorporated to provide the dorsiflexive torque, and the exploded view is shown in Fig. 5c. Finally, a cable guide is utilized to ensure the cable force to be collinear with the sleeve muscle, and a rotary potentiometer is attached to the joint for the position feedback to facilitate the motion control of the prosthesis.



(a) (b)

(c)

Fig. 5. Solid model of the transtibial prosthesis (a) and the details of the ankle joint design (b), exploded view (c).

IV. Conclusions and Future Work

In this paper the authors presented a new transtibial prosthesis design, which utilizes a unique sleeve muscle actuator to drive the ankle joint motion. As an advanced form of the pneumatic muscle actuator, the sleeve muscle incorporates a rigid cylindrical insert at the center. Multiple advantages are obtained as the result of this structural change, including greater force output, lower energy consumption, and faster dynamic response. More importantly, this new actuator also enables the integration of the actuator and the load-bearing structure, and thus can generate a highly compact system. Utilizing the sleeve muscle actuator, the transtibial prosthesis presented in this paper is able to provide the desired torque capacity and range of motion without significant enlargement of the radial dimension.

Currently, a prototype of the transtibial prosthesis described above is being fabricated, with the purpose of demonstrating its basic concept, as well as the desired kinetic performance and rehabilitation effects. For the future improvement of the basic design, the research will be focused on the development of a new sleeve muscle actuator with greater force capacity. Note that the current design utilizes a commercial muscle actuator, and its limited force output necessitates the selection of a pulley large enough to generate the desired torque output. The large diameter of the pulley, in turn, requires a longer muscle actuator to obtain the desired range of motion. Motivated by this issue, the authors intend to adopt a custom-made muscle actuator with greater force capacity (possibly via enlarging the actuator radially). As such, the length of the actuator, along with the overall length of the prosthesis, will be reduced to make the prosthesis design more practical, capable of benefiting a larger amputee population in the future.

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