VALIDATION OF PNEUMATIC ARTIFICIAL MUSCLE FOR POWERED TRANSFEMORAL PROSTHESES

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Abstract—**Despite progression in technology and medicine, lower limb prostheses users still endure many challenges that prohibit them from regaining their original movement abilities. Currently developed lower limb prostheses have been drastically improved in the last two decades; however, they still lack the actuation element which corresponds to the skeletal muscle in a biological limb. The main obstacle to this achievement has been the lack of a high power, small and light actuator. Electrical motors, hydraulic and pneumatic cylinders are very effective for most industrial applications; however, they are not suitable for prostheses-type applications. The implemented actuator must be light, powerful, energy efficient as well as safe to interact with user. In this paper, the validation of the Pneumatic Artificial Muscle for powered transfemoral prostheses is investigated. This is achieved by determining the kinetic and kinematic requirements of a human knee throughout walking, stair ascending/descending and sits to stand movements.**

***Index Terms*—Pneumatic Artificial Muscle, powered lower limb prostheses**

I. INTRODUCTION

For industrial applications, many forms of actuators have been developed such as electrical motors, hydraulic and pneumatic cylinders and piezoelectric actuators. However, none of these actuators have been shown to be feasible for medical assistive devices such as powered lower limb prostheses. The Pneumatic Artificial Muscle (PAM) is a pneumatically-powered actuator whose structure and behavior differ significantly from other type of actuators. In its most common configuration, PAM consists of a tubular braided mesh that wraps around an elastic bladder and both are strapped to end fixtures as shown in Fig.1. PAM operation is simple, as the PAM is inflated through the end fixtures, radial expansion of the muscle diameter occurs and causes the braid angle to steepen which in turn forces the tubular braid to foreshorten and thus produces a muscle contraction distance (see Fig.1). If the muscle contraction is resisted, PAM produces a substantial pulling force.

Fig.1. PAM prototype.(a) Pressurized state (b) Deflated state.

PAM offers a combination of properties that makes it very appealing for powered transfemoral prostheses compared to other forms of actuators. These properties have been characterized in [1] as follows:

*PAM mass:*PAM is distinctively light; its mass can be less than 50 grams. This is attributed to the thin braided mesh and an elastic bladder that constitute the muscle structure.

*PAM force:*In spite of its light mass, PAM produces a substantial force that is approximately six times greater than the force produced by a pneumatic cylinder of the same diameter.

*PAM stiffness:*PAM offers non-linear stiffness behaviour which is essential for legged locomotion and beneficial for passive-type actuation.

*PAM compliancy:*PAM offers a safe interaction with the user. This is attributed to the inherent pneumatic compliant behavior and to the flexibility of its outer shell.

*PAM mechanical connection:*Whereas typical actuators require a transmission mechanism and a precise alignment with the apparatus, PAM can be directly connected to deliver muscle contraction distance and muscle force.

*PAM speed and bandwidth:*Attributed to its unidirectional contraction and no transmission mechanism, the PAM is a relatively fast linear actuator.

*BPM Contraction distance:*A typical PAM achieves a contraction distance ratio with respect to the muscle unstressed length of 30%. This is comparable to biological skeletal muscles[2]**.**

Whereas there have been many claims [3] [4] [5] that the PAM is an ideal actuator for biomedical applications, to the author's knowledge, there is currently no study that quantitatively confirms the validation of the PAM for powering transfemoral prostheses. This paper will first present a comprehensive study of knee biomechanics to characterize its actuation requirement. Subsequently, based on existing PAM design capabilities, the validation of PAM powered transfemoral prostheses is determined.

II. PNEUMATIC ARTIFICIAL MUSCLE

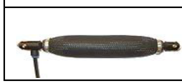
Throughout the last decades, various types of PAM have been developed; they can be mainly distinguished by the design of their external membrane and end fixtures. Currently, PAMs are manufactured only by two companies, namely Shadow Robot and Festo Corporation.

As shown in Fig.2, similar to the majority of PAM, the Shadow Air Muscle by Shadow Robot is made of an elastic bladder protected by a separate plastic braided sleeve. Both materials constitute the wall structure of the muscle. To transfer muscle force and retain muscle pressure, gear clamps and crimp rings are used to hold the bladder and the sleeve together with plastic end-fixtures. In this case, the muscle expansion is governed by the braided sleeve geometrical properties; the sleeve contracts axially when the tube inflates and expands radially. The use of gear clamp and crimp rings in these muscle designs limits the mechanical load that can be transferred by the muscle and also provides a weak gas seal at the muscle end-fixture. As a consequence, Shadow Robot [6] recommends that the Shadow Air Muscle should not be inflated to a gage pressure surpassing 206.8 KPa (30 psi) without load and should never surpass an operating pressure of 413.6 KPa (60 psi).

Fig.2. The Shadow Air Muscle by Shadow Robot [6].

Shadow Robot does not offer detailed technical information about their PAM; however, primary muscle performance for the three available muscle sizes is presented on their website as shown in Table I. According to the provided information, the maximum muscle force generated by the largest muscle size (30 mm in Diameter) is 686.5 N.

Table I: Shadow Air Muscle Range [6]

Products	Diameter	Length	Maximum Pull
	6mm	150 mm	69 N
	20mm	210 mm	196 N
	30mm	290 mm	686.5 N

Whereas the muscle produced by Shadow robot consists of a bladder that is wrapped by a braided sleeve, Festo Corporation offers a muscle design that consists of an elastic bladder embedded with aramid fibers that creates a trapezoidal pattern with a three-dimensional braid structure. Both muscles operate similarly, as the muscle is inflated, the muscle expands in diameter causing the braid or fiber angle to steepen and forcing the muscle to shorten. Distinctively, as shown in Fig.3, the Fluidic Muscle DMSP/MAS [7] incorporates a press fit design for the end-fixture which permits the muscle to withstand a large gage pressure and mechanical loading. The service life of this PAM is estimated between 100,000 and 10 million switching cycles for typical applications. This performance is a function of the relative contraction, operating pressure, loading behaviour and operating temperature of the muscle [7].

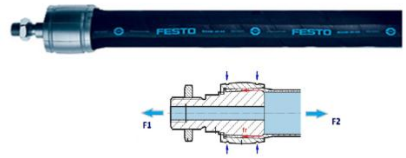
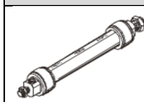


Fig.3. The Fluidic Muscle (DMSP) by Festo Corporation [7]

Similarly to a typical PAM, the Fluidic Muscle by Festo generates a maximum pulling force at full extension and decreases as the muscle contracts. Festo states that the effective operating range of the Fluidic Muscle is up to 15% of contraction length. The maximum pulling force by the Fluidic Muscle for the available diameter sizes is shown in Table II.

Table II: FESTO Fluidic Muscle Mechanical Properties

Products	DMSP Diameter	DMSP Length	Max. Pressure	Max. force
	10 mm	40 - 9000 mm	116 psi	630 N
	20 mm	60 – 9000 mm	90 psi	1500 N
	40 mm	120 - 9000 mm	90 psi	6000 N

III. Biomechanics of the Knee Joint

The knee is the largest and most complex joint of the human body. It has an indispensable role in our daily activities; it supports the body weight as an individual travels horizontally and vertically such as walking and ascending stairs. While transferring a substantial load from the upper body, the knee joint allows an impressive relative motion in three anatomical planes, namely sagittal, coronal and transverse plane, (Fig.4).

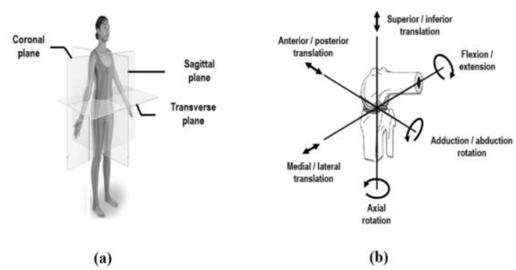


Fig.4. (a) Anatomical planes [8] (b) Degrees of motion of the knee [9]

The relative motion of the knee joint is a combination of rolling and sliding that is achieved in the three anatomical planes as shown in Fig.4. The knee joint has six degrees of freedom; however, the most significant amount of motion occurs in the sagittal plane. Angular displacement range in this plane ranges from 0^{o} to 140^{o}or from maximum extension to maximum flexion, respectively. The range of motion of the knee joint during level walking for all three anatomical planes is shown in Fig.5.

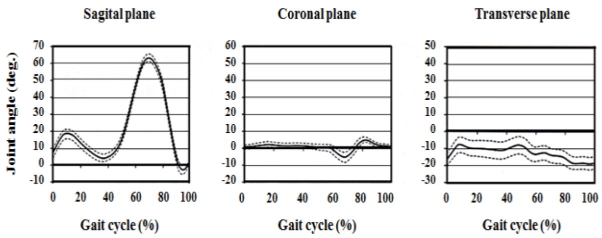


Fig.5. : Knee range of motion during level walking. Sagital plane: extension > 0^o, flexion < 0^o. Coronal plane: adduction > 0^o, abduction < 0^o. Transverse plane: internal rotation > 0^o, external rotation < 0^o.[10]

According to [10]**,** the angular displacement of a knee joint for typical daily activities is 117^o.Thus, the contraction distance of the quadriceps muscle can be determined as a function of the distance between the center of rotation of the knee and the patella-femoral contact. Measurements of this distance are obtained from radiological imaging. Fig.6 shows the relationship between the moment arm and flexion angle of the knee joint.

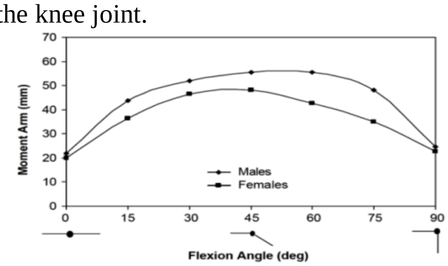


Fig.6. Knee moment arm versus knee flexion angle from male and female adults **[11]**

Assuming a hinge type of joint for the knee prosthesis and average moment of arm of 4.1 cm, to achieve an angular displacement of 117^o about the knee, this would require a PAM contraction distance of:

$$\Delta L = \theta * r = \frac{117^o}{180}\pi * 4.1 = 8.37\ cm$$

$$or\ contraction\ radio\ C_r = \frac{8.37}{30} = 27.9\%$$

Kinetics is the portion of biomechanics that studies the motion of a body when considering the mass and forces. Static force analysis during gait was achieved by developing a Solid Works three-dimension model of a human subject using anthropometric data from [12][13]. Thus, the dimensions of the model shown in Fig.7 are proportional to a standard human body height.

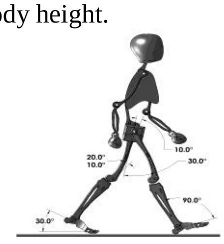


Fig.7. three-dimension model of a human subject

Whereas a static analysis throughout the gait for each study case is possible, static force analysis is only performed at peak knee moment that coincides with the highest demand of muscle forces during the stance phase.

Static analysis during walking

With reference to [10], the peak knee flexion angle through the stance phase occurs at approximately 16.5 % of the gait cycle which corresponds to the loading response or flat foot phase as shown inFig.8. This instant of the gait also corresponds to the maximum moment produced about the knee.

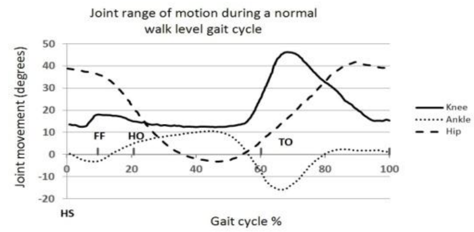


Fig.8. Joint range of motion during normal level walk gait cycle adapted from [10].

With reference to Fig.8**,** at 16.5 % of gait cycle, the hip, knee and ankle joints are found at 29^oextension, 17^o flexion, and 12^odorsiflexion, respectively. Using this information and the anthropometric data presented in [12][13], two-dimensional sagittal free body diagrams are drawn in SolidWorks. The first free body diagram Fig.9(a) consists of the entire subject and two external forces namely, the body weight W, and the ground reaction force F_g. For static equilibrium, the body weight force is aligned with the ground force reaction and the body weight does not generate any moment about the center of pressure.Thus, the ground friction force is null. As shown in Fig.9(b), the second free body diagram consists of the entire subject with exception of the lower legand foot with three external forces namely, the femoral-tibial contact force J,the quadriceps femoris muscle force vector Q and the body weight minus the lower leg and foot W_G. During a single stance of the gait cycle, one foot supports the entire body weight and the weight of the foot and lower leg do not contribute to the moment about the knee joint.

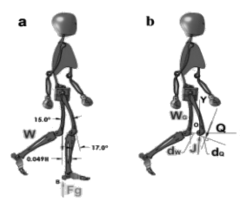


Fig.9. Free body diagram of subject during walking

With reference to Fig.9 (b), O corresponds to the femoral-tibial contact point and the center of rotation of the knee joint, d_W is the perpendicular distance between the vector force W_G and point O and d_Q is the perpendicular distance between the vector force Q and point O. Using the Solid Works model of the subject, the location of the center of mass, orientation of quadriceps muscle and perpendicular distances between Q and O and between W and O are automatically determined. The center of gravity of the lower leg and foot is located at 61% of the length of the leg measured from the knee joint. Applying an equilibrium condition at this instant of the gait, the quadriceps femoris muscle force is subsequently determined:

$$W_g = BW - W_L$$

where BW is the total body weight and W_L is the weight of the lower leg and foot. From[18]

$$W_L = 0.061BW \ or \ W_g = 0.939BW$$

From Solid Works model, d_{Wg} is 0.0436H where H is the total height of the body. According to [12], d_Q is equivalent to 45.54 mm for a knee flexion of 17°. Applying equilibrium conditions about point O with positive moments in the clockwise direction:

$$\Sigma M = 0$$

$$Q = \frac{W_g \times d_{wg}}{dq} = \frac{0.939BWx\ 0.049H}{0.02679H} = 1.72\ BW$$

Applying this result to a subject of 1.7 meters height and 736 Newton (75 Kg) body weight, the required quadriceps force is 1,263.1 N.

Stair ascending & sits to stand movements

Similar to the static analysis during walking, a two-dimensional sagittal free body diagram is considered to determine the required quadriceps force during stair ascending (Fig.10) and sits to stand movements (Fig.11).

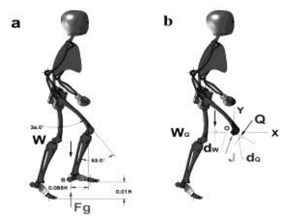


Fig.10. (a) Free body diagram 1 of the subject during stair ascent (b) Free body diagram 2 of the subject during stair ascent

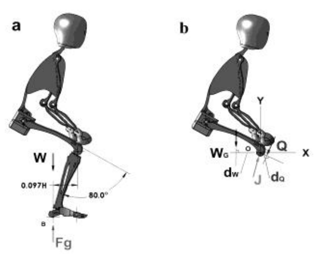


Fig.11. (a) Free body diagram 1 of subject during sit to stand (b) Free body diagram 2 of subject during sit to stand

Static analysis results indicated that the required quadriceps force for ascending stairs at 53° knee flexion is 2431.1 N and the required quadriceps force for sit to stand at 90° knee flexion is 1283.5N.

IV. Discussion

Biomechanics analysis showed that the maximum required quadriceps force during normal gait, ascending stairs, and sit to stand movement is 2434.1 N. Whereas the Fluidic Muscle DMSP/MAS (40 mm) produces a force that surpasses this requirement; the prosthesis apparatus cannot accommodate the muscle size. Based on anthropometric data of the human body, lower limb prostheses can be no more than approximately 40cm in length, 11cm in diameter and 3.5kg in weight. Given that a PAM expands radially by approximately 200% when it is fully contracted and assuming an antagonistic muscle setup, the maximum PAM diameter is limited to approximately 4cm. Thus, the maximum unstressed diameter of the PAM is about 2cm.

Using the muscle static force *F*model given by [1], to achieve a muscle force of 2434.1N that is required by powered lower limb prostheses while confining the size of the PAM, an operating pressure of 1.63 MPa is required. Currently, the maximum operating pressure of existing PAMs is limited to 0.83 MPa. Thus existing PAMs would fail to satisfy powered lower limb prosthesis requirements and consequently a new PAM design must be developed to sustain this higher pressure.This can be achieved by designing new end fixtures and a selection of combination of braid and bladder material that permits the muscle to operate at a very high pressure while withstanding extreme muscle pulling forces.

V. Conclusions

The objective of this paper was to validate the PAM as an actuator for powered transfemoral prostheses. A literature survey was first conducted to identify currently available PAM designs and analyze their limitations for actuating powered transfemoral prostheses. This led to a comprehensive study of knee biomechanics to characterize the actuation requirements for lower limb prostheses. Using Solid Works and anthropometric data, a three-dimension model of a human subject was created to achieve full understanding of human gait and also to perform a static force analysis of a subject during high energy movement namely, walking, ascending/descending stairs and sit to stand movements.

Having identified the actuation requirements for knee joint, it was clear that no currently available PAM would meet these criteria. The author has recommended the design of end fixtures and a selection of a combination of braid and bladder material that permits the muscle to operate at a very high pressure while withstanding extreme muscle pulling forces.

VI. References

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