Design and Evaluation of Pneumatic Artificial Muscle for Powered Transfemoral Prostheses

Marc Doumit* Jaime Murillo Agata Lawrynczyk Natalie Baddour

Department of Mechanical Engineering, University of Ottawa, Ottawa ON, K1N 6N5, Canada

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Abstract

Ideal prostheses are artificial limbs that permit physically impaired individuals freedom of movement and independence. Transfemoral prostheses have been drastically improved in the last two decades; however, despite advancements in technology and medicine, they are incapable of generating net power about joints to allow patients to regain their original mobility and improve their quality of life. From a biomechanics perspective, currently available devices assist patients by absorbing and dissipating energy and thus hinder them from achieving daily activities. Historically, one of the main challenges for powered medical assistive devices has been the actuation system that mimics the characteristics of the biological musculoskeletal system. A medical device must be light, small, and cosmetically appealing, but it must also generate forces and torques that are proportional to the individual’s weight. For industrial applications, many forms of actuators have been developed, such as electrical motors and hydraulic and pneumatic cylinders. However, none of these actuators have been shown to be optimal for powered transfemoral prostheses. Pneumatic artificial muscles (PAMs) are candidates for powering medical assistive devices. This study first conducts a comprehensive investigation of knee biomechanics. The failure of existing PAMs to meet the actuation requirements for powered transfemoral prostheses is then demonstrated. Based on the determined size, weight, kinetic, and kinematic requirements of knee articulation, a PAM design is proposed and evaluated. The proposed PAM is experimentally validated to meet the biomechanical requirements of a human knee during walking, stair ascent, and sit-to-stand movement.

Keywords: Pneumatic artificial muscle (PAM), Knee biomechanics, Powered transfemoral prostheses

1. Introduction

Powered transfemoral prostheses require actuators to achieve their desired outcome. Unfortunately, many typical forms of developed actuators, such as electrical motors, hydraulic and pneumatic cylinders, and piezoelectric actuators, have fallen short of meeting actuation requirements of transfemoral prostheses. A pneumatic artificial muscle (PAM) is a pneumatically powered actuator whose behavior is significantly different from those of other types of actuator due to its unique structure. As shown in Fig. 1, a typical PAM consists of an elastic bladder wrapped by a tubular braided mesh. The bladder and mesh are held in place via an attachment to an end fixture on each side of the muscle. This configuration allows for simple operation of the PAM. Specifically, the PAM is inflated through an end fixture, which causes radial expansion of the muscle diameter due to the steepening of the braid angle, and a simultaneous shortening of the muscle length. This action yields a muscle contraction. If the muscle contraction is resisted, the PAM produces a substantial pulling force.

Figure 1. PAM prototype in deflated (top) and pressurized states.

According to previous studies [1-4], PAMs offer a combination of properties that makes them very appealing for medical assistive devices compared to other forms of actuator. These desirable properties include light weight, large pulling force, non-linear stiffness, compliancy, simple mechanical connection, adequate speed, and large contraction distance [4]. In spite of these advantages, no previous study has quantitatively validated PAMs for powering transfemoral prostheses.

1.1 Pneumatic artificial muscle

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 by only two companies, namely Shadow Robot [5] and Festo Corporation [6]. Similar to the majority of PAMs, 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 end fixtures. 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 clamps 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 fixtures. Shadow Robot 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).

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 [5]. The largest muscle size, which corresponds to 30 mm and 290 mm in unstressed diameter and length, respectively, generates a maximum force of 686.5 N.

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 (3D) 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, the Fluidic Muscle DMSP/MAS [6] incorporates a press fit design for the end fixture, which permits the muscle to withstand a large gage pressure and mechanical load.

Similar to a typical PAM, the Fluidic Muscle by Festo generates a maximum pulling force at full extension that decreases as the muscle contracts. Festo states that the effective operating range of the Fluidic Muscle is up to 15% of the contraction length. The largest force is 6000 N for DMSP40, which has an unstressed diameter of 40 mm and can be operated at a maximum pressure of 0.62 MPa (90 psi).

1.2 PAM analytical models

Analytical models are critical for predicting muscle behavior to evaluate PAM for powered transfemoral prostheses prior to designing and fabricating PAM prototypes. Assuming that the muscle retains a cylindrical shape during contraction, its geometry is governed by the enfolding braid with a helix form. Defining θ as the braid angle, \( L_{\text{un}} \) as the uncoiled fiber length, and \( n \) as the number of fiber revolutions over length \( L_{\text{un}} \), the relationship between muscle diameter \( D \) and muscle length \( L_{\text{un}} \) is:

\[
D = \frac{L_{\text{un}} \sin \theta}{\pi n} \quad (1)
\]

\[
L_{\text{un}} = L_{\text{m}} \cos \theta \quad (2)
\]

The PAM static model is obtained by balancing the longitudinal forces developed due to the muscle gage pressure \( P \). Assuming that the muscle retains a cylindrical shape and that the wall thickness:

\[
F = -P \left[ \frac{L_{\text{m}}^3 - 3L_{\text{m}}L_{\text{un}}^2}{4\pi n^2} \right] \quad (3)
\]

1.3 Biomechanics of knee joint

The knee is one of the largest and most complex joints in 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 when walking and ascending stairs, respectively. While transferring a substantial load from the upper body, the knee joint allows an impressive relative motion in three anatomical planes, namely the sagittal, coronal, and transverse planes. The most significant amount of motion occurs in the sagittal plane, where the angular displacement ranges from 0° to 140° from maximum extension to maximum flexion, respectively. According to a previous study [7], the angular displacement of a knee joint for typical daily activities is 117°. Based on the angular displacement of the knee in the sagittal plane, the contraction distance of the quadriceps muscle can be determined as a function of its moment arm (i.e., the distance between the center of rotation of the knee and the patella-femoral contact). Measurements of this distance can be obtained from radiological imaging [8].

Contrary to the kinematics of the knee joint, the muscle forces that drive the knee are challenging to determine and highly dependent on the adopted model. Using an electromyogram-driven model, Winby et al. [9] estimated the quadriceps, hamstrings, gastrocnemius, and tensor fascia latae muscle forces during the stance phase of the gait cycle in healthy adults. According to this model, the quadriceps force is the greatest during early stance, specifically during the weight acceptance phase. Similarly, Messier et al. [10] used inverse dynamic techniques along with kinematic and kinetic data to calculate net joint forces and moments at the knee joint, which were then applied to a biomechanical model to estimate the knee muscle forces. Although the population under study was overweight adults with knee osteoarthritis, it was found that with weight loss, the quadriceps muscle force remains the same, while the hamstrings muscle force decreases [10].

1.4 Anthropometric model

For medical assistive devices, size and mass are critical design requirements. Due to the symmetry of the human body, balance has to be considered when designing a replacement for an amputated limb. A successful actuator should not only reproduce the required kinetics and kinematics but should also be confined to the mass and volume characteristics of the sound limb. According to a previous study [11], the mass of each segment of the lower limb can be expressed as a function of the total body weight \( W \). The same study suggested that assuming a subject with a body mass of 75 kg, a transfemoral prosthesis should not surpass the lower leg mass of approximately 3.5 kg. This is a significant design challenge for powered prostheses, which include actuators and a power supply. The gravity center of each body part depends on its mass, form, and length. The
2. Knee static force analysis

Although the results for quadriceps muscle force are applicable to normal bipedal locomotion, it was deemed imperative for this research to develop an analytical biomechanical model in order to validate PAMs for powered transfemoral prostheses. Knee static force analysis during gait was conducted by developing a 3D Solid Works model of a human subject using anthropometric data from the literature [11,12]. Thus, the dimensions of the model shown in Fig. 2 are proportional to those of a standard human body. To adequately evaluate the quadriceps force and subsequently the moment about the knee, force analysis was performed at five percent intervals during the stance phase of the gait cycle for level walking, stair ascent, and sit-to-stand movement.

Figure 2. Free-body diagrams of subject during level walking developed in Solid Works.

2.1 Static analysis during level walking

The hip, knee, and ankle joint angles from a previous study [7] were used for the stance phase of the gait cycle at five percent intervals. Using this information and the developed 3D model, two-dimensional (2D) sagittal free body diagrams were developed. The first free body diagram of Fig. 2 assumes single support and consists of the entire subject and two external forces, namely the body weight \( W \) and the ground reaction force \( F_g \). For static equilibrium, it was assumed that the body weight force is aligned with the ground reaction force and thus the body weight does not generate any moment about the foot center of pressure. In Fig. 2, the second free body diagram consists of the entire subject without one lower leg, instead showing three external forces, namely the femoral-tibial contact force \( J \), the quadriceps muscle force \( Q \), and the body weight minus the lower leg weight \( W_l \).

With reference to Fig. 2, \( O \) corresponds to the femoral-tibial contact point and the center of rotation of the knee joint, \( d_p \) is the perpendicular distance between the body weight vector \( W_s \) and point \( O \), and \( d_q \) is the perpendicular distance between the quadriceps muscle force vector \( Q \) and point \( O \).

For each knee flexion angle, \( d_q \) was obtained from the literature [8]. The location of the body center of mass, and thus \( d_w \), was also obtained from the literature [13]. Applying an equilibrium condition at any point of the gait cycle, the quadriceps muscle force \( Q \) is determined as:

\[
W_s = BW - W_l
\]

where \( BW \) is the total body weight and \( W_l \) is the weight of the lower leg. From a previous study [11]:

\[
W_l = 0.061BW \quad \text{or} \quad W_s = 0.939BW
\]

For a sample calculation, arbitrarily picking 10% of the gait cycle and assuming single support, according to a previous study [8], \( d_q \) is equivalent to 0.04601 m for a knee flexion angle of 17.6°. With reference to Fig. 2, \( d_w \) is 0.1322 m for a subject 1.7 m in height. Applying equilibrium conditions about point \( O \) with positive moments in the clockwise direction yields:

\[
\sum M = 0
\]

\[
Q = \frac{W_s \times d_w - F_s \times d_q}{d_q} = \frac{0.939BW \times 0.1322}{0.04601} = 2.698BW
\]

The required quadriceps force for a subject with a body weight of 75 kg (736 N) is thus 1985.1 N.

Referring to the same point of the gait cycle once again, but assuming the more realistic scenario of double support, \( F_s \) from the contralateral leg will contribute to the net moment about the point \( O \) on the knee joint of interest. The center of pressure of the contralateral foot is estimated to occur mid-foot. Assuming that the transfer of weight from one foot to another occurs linearly between 0 to 15% of the gait cycle, \( F_s \) is found to be approximately 48 N for a 75-kg subject and occurs at a distance \( d_q \) of 0.3541 m from the point \( O \). Applying an equilibrium condition at that point with positive moments in the clockwise direction, the quadriceps force \( Q \) is determined as:

\[
\sum M = 0
\]

\[
Q = \frac{W_s \times d_w - F_s \times d_q}{d_q} = \frac{690.87 \times 0.1322 - 48 \times 0.3541}{0.04601}
\]

\[
Q = 1612.80N
\]

This process was repeated for every 5% increment of the stance phase of the gait cycle. The results are shown in Fig. 3. The maximum calculated quadriceps forces are 3792 and 1647 N at the beginning of the gait cycle for the single and
double support scenarios, respectively. The quadriceps force obtained from the biomechanical model based on the double support assumption more closely matches the results obtained by Messier et al. [9] and Winby et al. [10].

![Figure 3. Quadriceps force for level walking during stance phase of gait cycle for single and double support.](image)

The maximum moment contributed to the knee joint by the quadriceps muscle during the stance phase of the gait is:

\[ M = Q \times d_q \]

\[ M = 3792 \times 0.02966 = 112.47 \text{Nm} \quad \text{(single support)} \]

\[ M = 1647 \times 0.0475 = 78.25 \text{Nm} \quad \text{(double support)} \]

Performing clinical gait analysis using a 3D musculoskeletal model, Kim et al. [14] experimentally determined the knee joint moment throughout the gait. According to their study, for a subject with a body weight of 75 kg (736 N), the maximum moment about the knee joint is approximately 56.9 Nm, which occurs at 16.5% of the gait cycle. While it is hard to compare the net knee moment obtained experimentally and the analytical moment calculated based on one muscle contribution (i.e., quadriceps muscle), the results obtained by Kim et al. [14] also more closely match the biomechanical model based on the double support assumption. The results obtained with a single support assumption are high; however, they do represent a possible gait scenario for a patient and thus are considered in the PAM evaluation.

2.2 Static analysis during stair ascent

Similar to the static analysis during walking, a 2D sagittal free body diagram is considered to determine the required quadriceps force during stair ascent (Fig. 4). The hip, knee, and ankle joint angles at 5% intervals of the gait cycle for this movement were obtained from the literature [14]. The location of the body center of mass [15,16] and the \( d_q \) and \( d_s \) values during double support were also determined for the calculation of the required quadriceps force.

The greatest required quadriceps force during stair ascent occurs at about 15% of the gait cycle. For an individual 1.7 m in height and 736 N in body weight, applying equilibrium conditions about point \( O \) with positive moments in the clockwise direction yields:

\[ \sum M = 0 \]

\[ Q = \frac{W_g \times d_s}{d_q} = \frac{0.939BW \times 0.0821}{0.05613} = 1.37BW \]

![Figure 4. Free-body diagram of subject during stair ascent developed in Solid Works.](image)

The required quadriceps force for a subject with a body weight of 75 kg (736 N) is thus 1008.32 N.

The moment contributed to the knee joint by the quadriceps during this instance of the gait for the single and double support cases is:

\[ M = Q \times d_q \]

\[ M = 1011 \times 0.05613 \]

\[ M = 56.7 \text{Nm} \]

Experimental results obtained by Kim et al. [14] for the knee moment during a normal stair ascent gait cycle show that the maximum knee moment is 1.008BW and occurs at 17.6% of the cycle during the weight acceptance phase. Applying this proportional value to a 75-kg subject, the maximum knee moment is 75.6 Nm. Similar to the previous static analysis, the discrepancy between the analytical model and experimental results is attributed to the assumptions made for the biomechanical model.

2.3 Static analysis during sit-to-stand movement

Lastly, a similar static analysis is considered to determine the required quadriceps force during the sit-to-stand movement. The body weight is assumed to be equally distributed to both lower limbs. Therefore, the load acting on each knee joint is:

\[ W_{s1} = \frac{BW}{2} - 0.061BW \]

\[ W_{s1} = 0.439BW \]

\[ W_{s1} = W_{s2} \]

or 323N for an individual with a \( BW \) of 736 N (75 kg). Using data from a kinematic analysis [17], the lower body joint angles for the sit-to-stand movement at 5% intervals were used to create a Solid Works model (Fig. 5). Applying the equilibrium conditions about the femoral-tibial contact point \( O \) gives:

\[ \sum M = 0 \]
\[ Q = \frac{W_d \times d_m}{d_q} \]

where \( d_{mq} \) is determined by assuming that the body center of mass acts directly over the center of pressure, which occurs in the heel of the foot.

![Free-body diagram of subject during sit-to-stand movement developed in Solid Works.](image)

The greatest required quadriceps force for a 1.7-m tall individual with a BW of 736 N (75 kg) is 1005 N and occurs at about 35% of the movement pattern. This corresponds to a knee joint angle of 82°.

The moment contributed to the knee joint by the quadriceps during this instance of the movement is:

\[ M = Q \times d_q \]
\[ M = 1005 \times 0.03734 \]
\[ M = 37.5 Nm \]

Experimentally obtained knee joint moments during the sit to stand movement were determined by Kim et al. [14] for two cases, namely with and without arm support. The maximum moment values of the knee joint were determined to be 0.77 and 0.88 Nm/kg or 57.75 and 66 Nm for a subject with a body weight of 75 kg with and without arm support, respectively. Similar to the previous analysis, the discrepancy between the theoretical and experimental results is attributed to the assumptions made in the biomechanical model. Moreover, the analytical values obtained are significantly affected by the positions of the arms, head, and trunk, which can alter the body center of mass, which in turn varies the location of the center of pressure and finally the counterbalanced moment or the quadriceps force.

3. PAM design requirements

Based on the findings of the static force analysis and the size restraints, five PAM design parameters were identified: (i) PAM pulling force \( F_m \) to achieve the required knee moment, (ii) PAM contraction distance \( \Delta L \) to achieve the required knee angular displacement, (iii) PAM weight \( W_m \) to limit the weight of the prosthesis, (iv) PAM diameter \( D \), and (v) PAM length \( L \) to confine the size of the prosthesis.

Biomechanics analysis showed that the maximum required quadriceps forces during normal gait, stair ascent, and sit-to-stand movement are 3792 and 1647 N, with corresponding maximum knee moment \( M_k \) values of 112.47 and 78.25 Nm for single and double support, respectively. Assuming a prosthesis moment arm that is similar to the average biological moment arm of 4.1 cm, the PAM must produce a maximum pulling force \( F_m \) of:

\[ F_m = \frac{M_k}{R_s} \]
\[ F_m = \frac{112.47}{0.041} = 2743.17 \text{ N (single support)} \]
\[ F_m = \frac{78.25}{0.041} = 1908.53 \text{ N (double support)} \]

As stated earlier, the angular displacement of a knee joint for typical daily activities such as walking and ascending stairs is approximately 117°. To achieve this angular displacement about the knee prosthesis and assuming a hinge type of joint with an average moment of arm of 4.1 cm, this would require a PAM contraction distance \( \Delta L \) of:

\[ \Delta L = \theta \times r = \frac{117^\circ \times \pi \times 4.1 \text{ cm}}{180} = 8.37 \text{ cm} \]

Based on anthropometric data presented earlier, powered knee prostheses can be no more than approximately 40 cm in length, 11 cm in diameter, and 3.5 kg in weight. Given that a PAM expands radially by approximately 200% when it is fully contracted and assuming an antagonistic PAM setup, the maximum PAM diameter is limited to approximately 4 cm. Thus, the maximum unstressed diameter \( D \) of the PAM should be confined to 2 cm. Moreover, the maximum PAM length \( L \) must be limited to approximately 30 cm, which would yield a PAM contraction ratio \( C_r \) of:

\[ C_r = \frac{8.37}{30} = 27.9\% \]

Furthermore, based on identified masses of existing passive prostheses and the mass of a sound limb, the proposed PAM mass should be limited to 250 g. Adhering to the design specifications of an unstressed PAM length and diameter of 30 cm and 2 cm, respectively, and a typical mesh braiding angle \( \theta_m \) of 20°, the uncoiled fiber length \( L_{unm} \) from Eq. (2) is 40.9 cm. From Eq. (3), to achieve a force \( F \) of 2743.17 or 1903.53 N while restricting the unstressed muscle diameter to 2 cm, the pressure \( P \) should be:

\[ -P = \frac{4F \pi n^2}{E_{unm} - 3L} \]
\[ P = 166 \text{ psi} \text{ (single support)} \]
\[ P = 115 \text{ psi} \text{ (double support)} \]

Thus, to meet the biomechanic requirement of a knee joint, the PAM must operate at a pressure range of 1.15 MPa (167 psi).
to 1.66 MPa (240 psi), and achieve a contraction ratio of no less than 27.9%. Currently, none of the PAMs designed by Festo or Shadow Robot meets these criteria and thus they would fail to actuate a knee joint throughout daily activities.

4. PAM prototype design

The PAM is a device that converts gas pressure into a high pulling force. It mainly consists of three components: the internal flexible tube or bladder, the braided sleeve, and the end fixtures. It was deemed important to focus the design effort on the end fixtures, since this has been the weakest link of existing PAM designs. During high operating pressures, the end fixtures must sustain high mechanical loads while connecting muscle components together and sealing the high-pressure gas inside the muscle. Moreover, the muscle end fixtures transfer muscle force to the apparatus and act as a conduit for gas flow.

The first conceptual design is a simple cylindrical arrangement of three parts (Fig. 6). Part 1 has an external cone to hold the bladder and the sleeve together and a centered longitudinal hole acting as a conduit to inflate and deflate the muscle. Part 2 has a hexagonal external shape to ease the process of assembly and an internal cone that serves to press the bladder and sleeve together. Part 3 is a commercial nut used to exert the initial force on part 2.

The braided sleeve and the internal tube have the same length and are held together with the end fixture between the two conical and parallel surfaces of parts 1 and 2. When the nut (part 3) is tightened, the generated force \( F_n \) moves part 2 toward part 1 and consequently increases the normal force \( N \) which acts perpendicularly to the sleeve and bladder. To avoid slippage in the end fixtures, the frictional force \( F_f \) must be greater than or equal in magnitude and opposite in direction to the sleeve and bladder tension \( T \).

With the intention of increasing the contact area between the sleeve-bladder and the end fixtures and ultimately increasing the load capacity of the muscle, a variation of the first design is proposed.

Conceptual design 2 consists of successive waved chambers between the two conical and parallel surfaces of parts 1 and 2 to accumulate material and restrict it at the periphery of each cavity. This eliminates relative motion of the sleeve bladder material between parts 1 and 2. Consequently, design 2 retains higher tension loads in section A, but the fabrication of these chambers is more complex and thus its manufacturing cost would be higher. A third conceptual design is proposed to reduce the number of circular cavities and add a hexagonal form at the end of part 1 in order to reduce the manufacturing costs while maintaining its desirable characteristics.

This third design keeps the self-retaining feature of Design 1 and the circular cavity from Design 2, which acts as a high-pressure ring where the force \( F \) is longitudinally perpendicular to the sleeve-bladder surface. The hexagonal external forms in parts 1 and 2 are used as a means of holding the end parts when assembling the end fixtures. Figure 8 displays the assembled parts of the end fixtures for both sides of the muscle. Side 1 is the part of the muscle that supplies and releases the gas and Side 2 is completely sealed and usually connected to the load.

![Figure 7. Isometric contraction results of muscle 14P-10S-L300.](image)

![Figure 8. PAM experimental results compared to quadriceps force requirements during level walking from biomechanical model for knee flexion angles.](image)

5. PAM prototype evaluation

To evaluate the PAM prototypes, two experimental setups were designed and built. These setups were designed according to the requirements of three different types of muscle contraction achieved by a skeletal muscle: 1) isometric muscle contraction, which takes place when the muscle generates force, but its length remains constant, 2) eccentric muscle contraction, which occurs when the force generated by the muscle is inefficient to hold the applied load and thus the muscle length extends during contraction, 3) concentric muscle contraction, which happens when the muscle force is greater than the applied load and thus the muscle length decreases as the muscle contracts.
The isometric contraction and eccentric contraction test were performed using a tensile testing machine (model 4482, Instron, USA), where the PAM was installed at a specific position between the load cell at the moving crosshead and the base beam.

For isometric contraction, the PAM was restrained to a specific length and then inflated to a specific pressure. The muscle pressure and the muscle force were measured by a pressure transducer and by the Instron machine load cell, respectively. These measurements were recorded using a data acquisition system (LabVIEW, National Instruments, USA) running on a personal computer (PC). For the eccentric contraction, the PAM was initially inflated at a specific pressure and then mounted in the tensile testing machine at its maximum contraction length. While retaining a constant muscle air mass during the test, the muscle was gradually pulled at a rate of 10 mm/min until it reached the muscle unstressed length \( L_o \). PAM length elongation resulted in a decrease in muscle volume and thus an increase in muscle pressure. The muscle force and the muscle length were measured by the Instron machine load cell and displacement sensor, respectively. Moreover, the muscle pressure was measured by a pressure transducer and all measurements were recorded using a data acquisition system (LabVIEW).

For the PAM concentric contraction test, the muscle was suspended vertically within a setup while being fixed at one end and attached to a mass at its other end. As the muscle was inflated, muscle contraction lifted the weight until it reached force equilibrium or the maximum contraction distance of the PAM prototype. The muscle contraction distance was monitored using a linear variable displacement transducer (LVDT) sensor. The concentric contraction setup was controlled by a software program using LabVIEW running on a PC.

6. Prototype testing

The prototype, which consisted of a bladder made of natural latex rubber and a braided sleeve made of polyethylene terephthalate (PET) monofilament yarns, was first tested for gas retention over a period of five days. Experimental results showed that the PAM pressure remained constant at 0.62 Mpa, which confirms the sealing properties of the joint achieved by the end fixtures.

The second prototype test consisted of an isometric PAM contraction used to determine the capacity of the end fixtures to retain the sleeve-bladder when the muscle is loaded. The muscle was installed in the Instron tensile machine at its unstressed length and pressurized to increment the pulling force generated by the PAM until the muscle failed. A maximum pulling force of 3700 N was achieved; however, PAM failure occurred when its membrane slipped out of the end fixtures.

Table 2 presents the values of parameters of the tested prototypes for unstressed and contracted states.

Table 2 show the PAM achieved a contraction distance and contraction ratio of 90 mm and 32%, respectively. These results exceed the kinematic requirements for powered lower limb prostheses, which are a contraction distance and contraction since the end fixtures were not able to hold the braided sleeve to material failure, an alteration to the fixture design was prompted. A small cup form was added between parts 2 and 3 to eliminate PAM membrane slipping from the end fixtures at high pulling load. Moreover, part 2 was machined to match the surface of the new cup. The cups increase the sleeve grabbing area and break the continuity of the slipping path. Experimental testing demonstrated the failure of the braided sleeve after sustaining a pulling load of beyond 9000 N. In this test, the ends of the sleeve were in good condition, which demonstrates that the fixtures can withstand forces that are larger than the ultimate strength of the sleeve and bladder materials combined.

Throughout the experiments, it was observed that as the muscle contracted, muscle radial expansion resulted in hemispherical deformation at the end of the muscle. It was observed that the bladder material tended to bulge out from the braided sleeve, resulting in premature bladder rupture at high operating pressures. To prevent sleeve fabric deformation, a coat of liquid rubber was applied to the muscle covering the hemisphere area at each of its ends. This creates an elastic film that maintains the symmetry and uniformity of the fabric in the hemisphere critical area.

7. Results

For the experimental analysis, PAM prototypes of different materials and sizes were built in the laboratory. The prototypes were of the same design and consisted of a braided sleeve, a bladder, and end fixtures. To identify the PAM prototype, a designation code was established, as shown in Table 1.

<table>
<thead>
<tr>
<th>PAM prototype designation codes</th>
<th>Unstressed sleeve diameter (mm)</th>
<th>Sleeve material</th>
<th>Bladder internal diameter (mm)</th>
<th>Bladder material</th>
<th>Unstressed muscle length (mm)</th>
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</thead>
<tbody>
<tr>
<td>14P-108-L300</td>
<td>14 mm</td>
<td>PET</td>
<td>10 mm</td>
<td>Silicone</td>
<td>300</td>
</tr>
</tbody>
</table>

7.1 Concentric muscle contraction

For the concentric muscle contraction test, two PAM prototypes (14P-10S-L300 and 14P-12R-L300) were constructed. An unstressed prototype length of 300 mm and a sleeve diameter of 14 mm were selected to satisfy the PAM size constraint defined earlier. The PAM prototypes were connected to a load through a steel cable that wrapped around a pulley to generate a torque and an angular displacement when the muscle contracted. The prototypes were inflated to a pressure of 1.42 MPa (200 psi) and contracted from an unstressed muscle length \( L_o \) of 300 mm to the maximum contraction distance.

The results in ratio of 83.7 mm and 27.9%, respectively.

7.2 Isometric muscle contraction

For the isometric muscle contraction test, two prototypes
Instron tensile testing machine and then inflated to a pressure of 1.42 MPa (200 psi). Although a higher pressure could have been applied to the PAM, the selected pressure was deemed sufficient based on the pressure range identified earlier. Figure 9 displays the isometric contraction test results of the muscle 14P-10S for contracted muscle lengths of 200 to 300 mm.

The experimental results in Fig. 7 show the PAM pulling force as a function of its contracted length. A maximum muscle force of 2700 N was obtained at an operating pressure and contracted muscle length of 1.42 MPa and 300 mm, respectively. While this result demonstrates that the muscle is capable of generating the maximum force required during gait, it is critical to show that the PAM can sustain the required force as it contracts throughout the three identified daily activities. As stated earlier and shown experimentally in Fig. 7, the muscle force magnitude decreases as the muscle contracts. Figure 8 displays the required force calculated based on the developed biomechanical model and the PAM force based on the experimental results and an average moment of arm of 4.1 cm for level walking.

With the exception of the loading response phase (i.e., 0 to 15% of the gait cycle) for the single support scenario, operating at a pressure of 1.42 MPa (200 psi), the designed PAM satisfies the quadriceps force requirement during level walking.

Figure 9 displays the required force calculated based on the developed biomechanical model and the PAM force based on the experimental results and an average moment of arm of 4.1 cm for stair ascent. Operating at a pressure of 1.42 MPa (200 psi), Fig. 9 confirms that the designed PAM satisfies the quadriceps force requirement throughout the stair ascent movement.

Figure 10 displays the required force calculated based on the developed biomechanical model and the PAM force based on the experimental results and an average moment of arm of 4.1 cm for sit-to-stand movement. With the exception of large flexion angles (i.e., over 85°), which correspond to large contraction distances, operating at a pressure of 1.42 MPa (200 psi), the designed PAM satisfies the quadriceps force requirement during sit-to-stand movement.

By increasing the operating pressure beyond 1.42 MPa (200 psi), the designed PAM can achieve the required forces during all three activities for both support cases; however this was deemed to be not the optimal solution. A slight deficiency of the muscle force can be compensated instead by an increase of moment arm of the prosthesis. This would be a prosthesis design parameter as the moment of the arm must be variable to optimize the produced moment while not compromising the angular displacement achieved by the muscle contraction. Similar to a biological knee, the moment of the arm must vary as a function of the knee flexion angle to meet its biomechanical requirements.

8. Conclusion

This study evaluated a PAM as an actuator for powered transfemoral prostheses. A comprehensive study of knee biomechanics based on a literature review was conducted to characterize the actuation requirements for lower limb prostheses. Using Solid Works and anthropometric data, a 3D model of a human subject was created to achieve a better understanding of human gait and to perform a static force analysis of a subject during high-energy movements, namely walking, stair ascent, and sit-to-stand movement.

Based on the identified actuation requirements for the knee joint, it was clear that no currently available PAM meets these criteria. A design of end fixtures and a selection of a combination of braid and bladder materials that permits the muscle to operate at a very high pressure while withstanding extreme muscle pulling forces were conducted. Numerous prototypes were fabricated and experimental setups were designed and built for experimental testing. Finally, concentric,
eccentric, and isometric muscle contraction tests were performed, demonstrating the capability of the designed PAM to satisfy the biomechanics requirements of a human knee during walking, stair ascent, and sit-to-stand movement.

References


