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A powered prosthetic knee joint inspired from musculoskeletal system

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ABSTRACT

This paper reports on a powered prosthetic knee joint powered by artificial muscles. A musculoskeletal system integrating artificial and biological muscles was simulated. The gait cycle was divided into seven modes. Based on the results of the simulation, the artificial muscles were pressurized to provide the biological knee torque. Analysis of the gait trials of an amputee showed the timing of artificial muscles was similar to EMG of biological knee muscles. This paper is an initial step forward to implement the concept of biomimetic approach in prosthetic knee technology.

1. Introduction

A major source of disability among the patients who have lost their limb is the lower extremity amputation, and specifically, transfemoral amputation. Approximately, 1.9 million people are affected by limb loss, and 400,000 of them have above the knee amputation [1]. Due to important role of prosthetic knee in transfemoral ambulation, its design and function are particularly important [2]. One of the deficiencies of current prosthetic knees that prevent them from achieving this control is their inability to introduce new energy into locomotion. As a result, they cannot emulate certain functionalities of biological knee in situations like level ground walking, sit to stand maneuver, and stair/slope accent [3].

In regard to the level ground walking, many experimental studies have shown the energy expenditure of transfemoral amputees is higher than transtibial amputees and able-bodied (for example [4,5]). The inability of the knee to add net power prevents the prosthetic knee from emulating the normal angle of the biological one particularly in the stance phase. This leads to an asymmetric gait which contributes to higher energy expenditure of transfemoral amputee locomotion [6]. Furthermore, our simulation studies showed that due to absence of functionality of muscles which are lost during transfemoral amputation, the residual muscles efforts will be higher compared to a normal joint [7], which may also

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contribute to the higher energy expenditure of transfemoral amputee gait. Therefore, it is speculated that if actuators able to deliver power, are added into a prosthetic knee joint, transfemoral amputee gait will improve.

To address the above issue, some studies designed and constructed active prosthetic knee joints [3,8–10]. In these studies, the strategy used to activate the actuator was based on considering the prosthesis as a machine which was separated from the body, and was designed to fulfill a requirement related to position [3,8–10].

However, it is expected that the artificial prosthetic limbs will progress toward biomimetic limbs that function as intimate extensions of human body [11]. As a result, their actuators might have activities like biological muscles. For this purpose, the activation of actuators of a prosthetic knee might be designed based on dynamics of musculoskeletal system.

Although the concept of biomimetic prostheses has been proposed in the literature [11,12], the authors are not aware of a previously reported biomimetic prosthetic knee joint. Therefore, as an initial step toward biomimetic prosthetic knee technology, the goal of this study was to implement and evaluate a biologically inspired approach to power a prosthetic knee, which is based on activation of biological muscles in musculoskeletal system. In comparison to previous studies that used electric motor [3,8,9] or cylinder-piston actuator [10], in the presented biomimetic knee prosthesis, one kind of McKibben pneumatic muscle, Festo artificial muscle (Festo Corporation, Germany) was used. While according to the literature this actuator exhibits characteristics similar to biological muscle [13,14], its structure and force exertion characteristics were most relevant to this study: like biological muscle, it has an origin and an exertion point; to apply force, contracts [14], and has high force/volume ratio [15]. These features make it possible to consider this actuator as an extension of body to design the activation pattern based on dynamics of musculoskeletal system.

2. Materials and methods

2.1. Lower extremity model

Based on a biomimetic approach, assuming the artificial leg as an integrated part of the residual limb, we carried out a computer simulation to determine the forces of artificial muscles for normal hip and knee angles. The model is schematically shown in Fig. 1. Only the motions in the sagittal plane were considered to be important, and it was assumed that no rotation happens at the ankle joint. Therefore, the model was a linkage, for which, hip and knee were functional joints, and it was comprised of residual and prosthetic limb. For intact biological muscles, the attachment coordinates in the reference skeletal frames were based on the data reported by Delp [16]. For transected biological muscles, assuming a myodesis, in which the new attachments of muscles end are fixed to the amputated tip of the bone, the distal attachments of the transected muscles

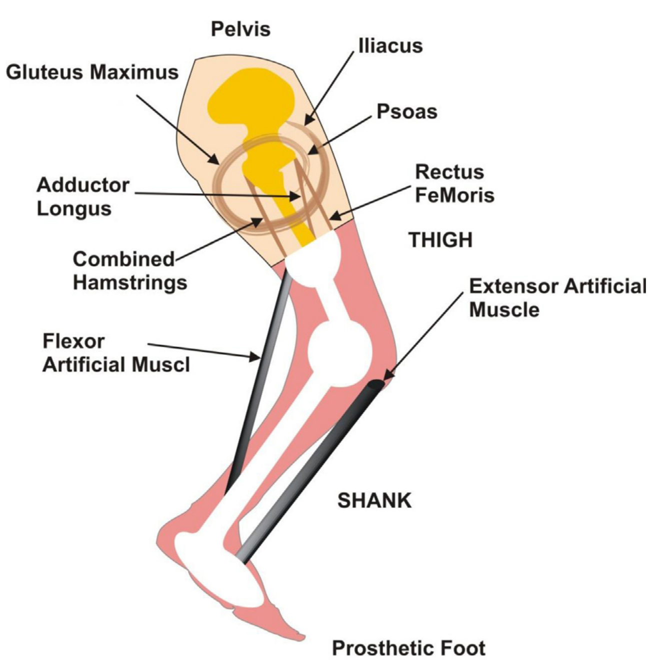

Fig. 1 – Lower extremity model which integrates residual limb and the prosthesis.

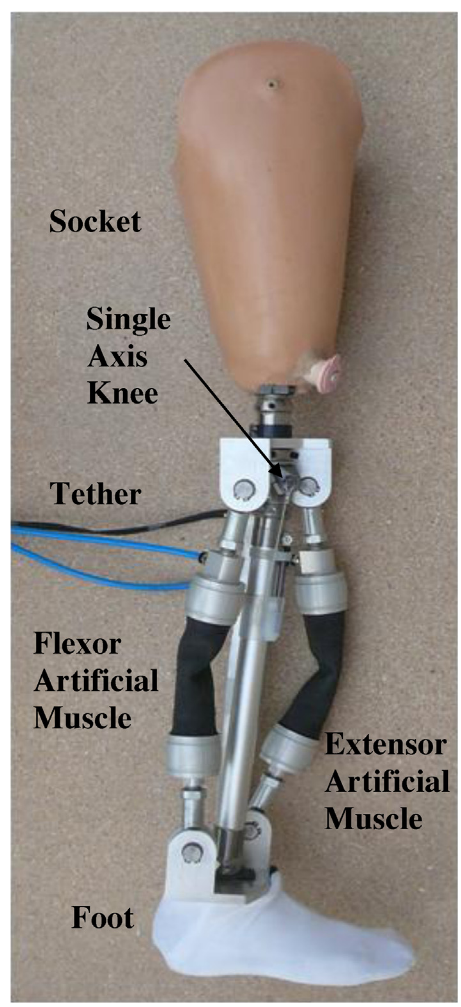

Fig. 2 – The structure of the prosthetic limb, showing the antagonist artificial muscle structure.

were changed [17,18]. Two antagonist Festo muscles provide the desired torque at knee joint, which exert force when they are pressurized. Based on a static optimization approach [19], for normal hip and knee angles [20], the forces of all muscles were calculated to produce normal torque at these joints [21]. To control the force that Festo muscles exert, the equation proposed by Tondu and Lopez [22], was used. This equation using measurable parameters of the muscle, relates its force and pressure together. The model is mathematically described in Appendix A.

2.2. *Mechanical structure*

Fig. 2 shows the prosthetic knee in a labeled photograph. It includes a single axis knee joint and two Festo pneumatic muscles. The joint is an OttoBock (Otto Bock OrthopaK dische GmbH and Co., Duderstadt, Germany) single axis one model 3A15, for which the brake has been deactivated. This joint has an extension-assist spring that applies passive torque. The Festo artificial muscles which are model MAS-40, are attached to thigh and shank by two brackets, via eye bolts. The upper and lower brackets were designed to withstand a 2000 N force applied by each actuator. Safe stress conditions were verified using ANSYS finite element software version 12.1. The results of this analysis indicated that if the brackets are made from 7000 series aluminum, a minimum safety factor of 3.0 will be provided. The prosthetic foot is a Solid Ankle-Cusion Heel (SACH) type. The pressure inside each muscle is controlled via a Festo proportional pressure regulator model MPPE-3-1/4-10-010-B. The air required to pressurize muscles is provided by an external compressor.

2.3. *Control strategy*

To pressurize the agonist–antagonist artificial muscles with values calculated by the model described in Section 2.1, the gait cycle was divided into seven distinguishable modes shown in Fig. 3. If the mode in which ambulation takes place is identified, the appropriate pressures for each artificial muscle can be determined. The criteria for each mode are heel-ground contact, toe-ground contact, and the sign of knee velocity.

2.4. *Sensors, electronics, and controller interface*

Three sensors are implemented in the prosthetic leg. Two of them are on–off switches that are placed under the sole, and being used to determine whether the heel and the toe are in contact with ground or not. The third sensor is a rotational potentiometer that is used to measure knee flexion velocity.

To acquire signals with minimum noise from sensors, a data acquisition card was custom-designed and constructed which along with LABVIEW software version 9 was used to read the signals from sensors, process them, and send output signals to pressure regulators. Fig. 4 shows a schematic of the data acquisition card and its hardware.

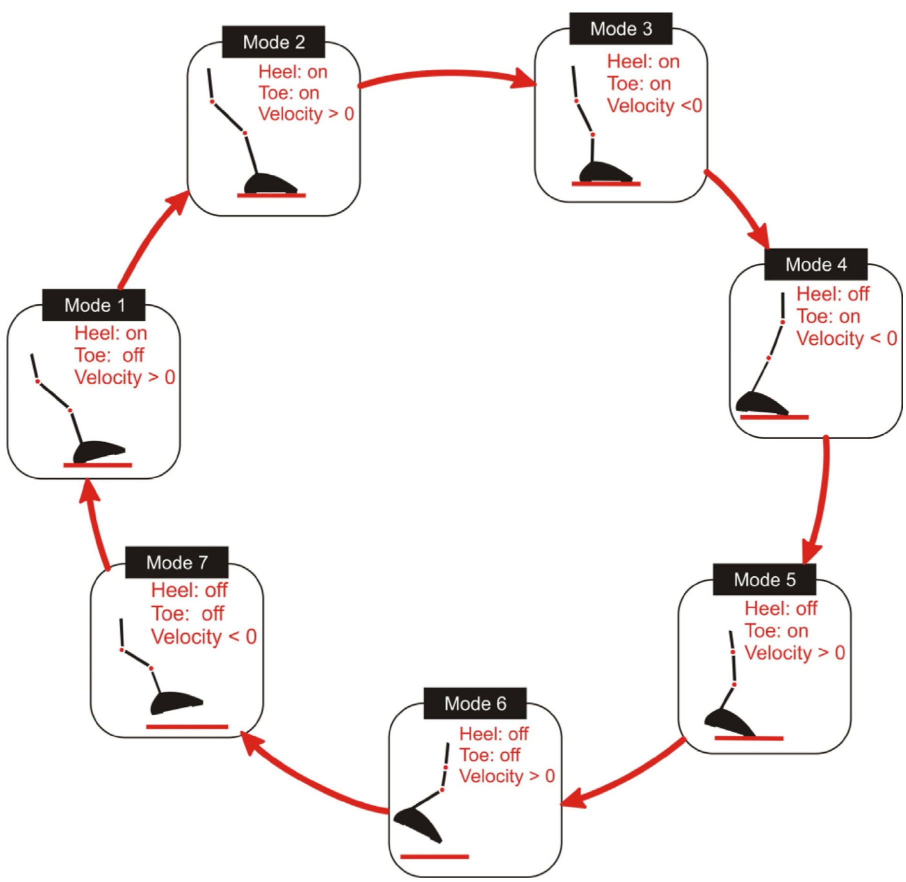

Fig. 3 – The finite state criteria according to which the gait cycle has been divided. Each mode is distinguished by knowing whether heel and toe are in contact with ground, and the sign of knee flexion velocity.

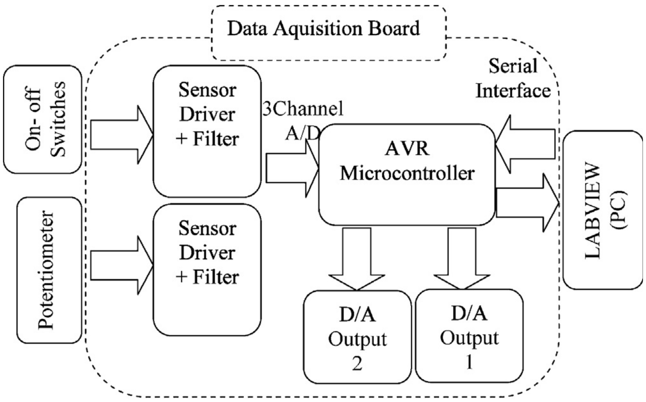

Fig. 4 – A schematic and the hardware of the data acquisition card.

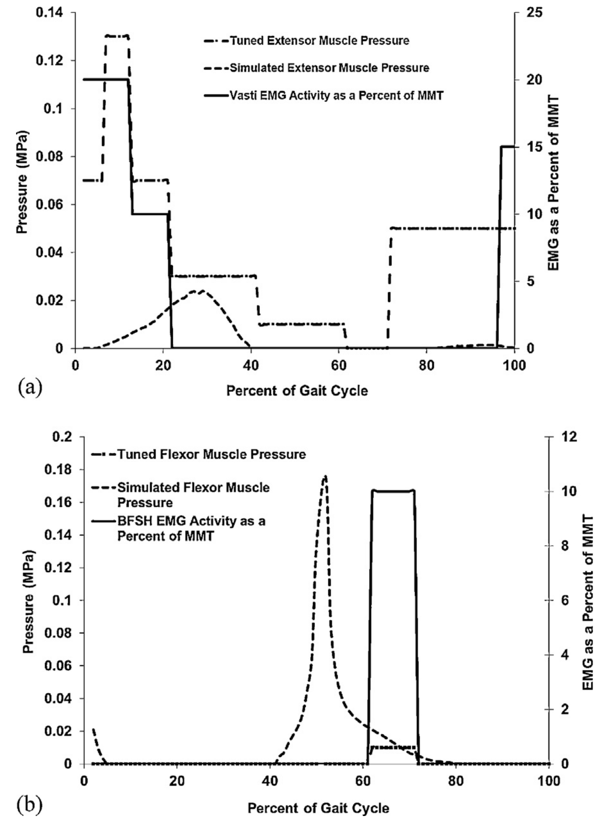

Fig. 5 – The timing of the activation of artificial and biological muscles. (a) Extensor muscle; (b) flexor muscle.

The LABVIEW interface program calculates knee flexion velocity, and based on the sign of this velocity and signals sent from heel and toe on–off switches, determines the current mode of gait, calls the artificial pressures that pertain to the calculated gait mode, and sends the appropriate signals to the pressure regulators.

3. Results and discussion

To evaluate the performance of the powered prosthetic knee, an amputee was asked to wear it, and then his gait was evaluated. He was 47 years old, 183 cm high, with 75 kg weight. Kinematic data of the lower limb during walking were measured by a motion analysis system (WINanalyze 1.4, 3D, Mikromak Gmbh, 1998, Germany). A high speed camera (Kodak Motion Corder, SR-1000, Dynamic Analysis System Pte Ltd., Singapore) was used to record the two-dimensional motion of the body segments taken at 125 frames s^{-1}. Three markers were used to record knee angle during gait. An inverse dynamics method was used to calculate hip and knee torque from recorded hip and knee angles [23].

The mean value of simulated pressure was used as preliminary pressure for each artificial muscle. The amputee's gait was analyzed using the motion analysis system. Then, during gait, the pressure inside each muscle was tuned until the amputee felt comfortable and also a biological knee angle was emulated by the amputee. For example, during pre-swing he felt that the prosthesis was too stiff to start the swing phase easily. So, the pressure inside the extensor muscle was reduced at that stage.

Fig. 5 shows the preliminary and tuned pressures for both artificial muscles. The difference between two values is mainly due to the fact that the curves used to calculate initial pressure were based on knee and hip angles that pertained to a group of subjects [20], and therefore, they differ from those calculated for the subject who participated in this study. From mode 1 to mode 5, the extensor muscle is activated. As heel strikes the ground (mode 1), this muscle starts contraction, which prepares the knee for loading response. Then, the knee continues flexion, and the sole of the foot touches ground. During this period, the extensor muscle remains activated to bear the applied load. When both heel and toe are in contact with ground (mode 2), the activation of this muscle is increased to be able to maintain the stability of the flexed prosthetic knee under body weight. After maximum stance knee flexion, the knee starts extension (mode 3). As this happens, the activation of the extensor muscle is decreased. As knee continues extension, the heel leaves ground (mode 4) and the activation of extensor muscle is further decreased. Then, while toe is in contact with ground, the knee starts to flex (mode 5). At this point, the activity of extensor muscle is again decreased. As knee continues flexion, the toe also leaves ground (mode 6). While this happens, the extensor muscle is deactivated and flexor muscle is activated until knee reaches its maximum swing flexion. After maximum knee flexion (mode 7), the flexor muscle is deactivated, and extensor muscle is again activated.

Also, in Fig. 5 the function of the artificial extensor and flexor muscles have been compared with that of the EMG activity of vasti muscles and short head of biceps femoris,

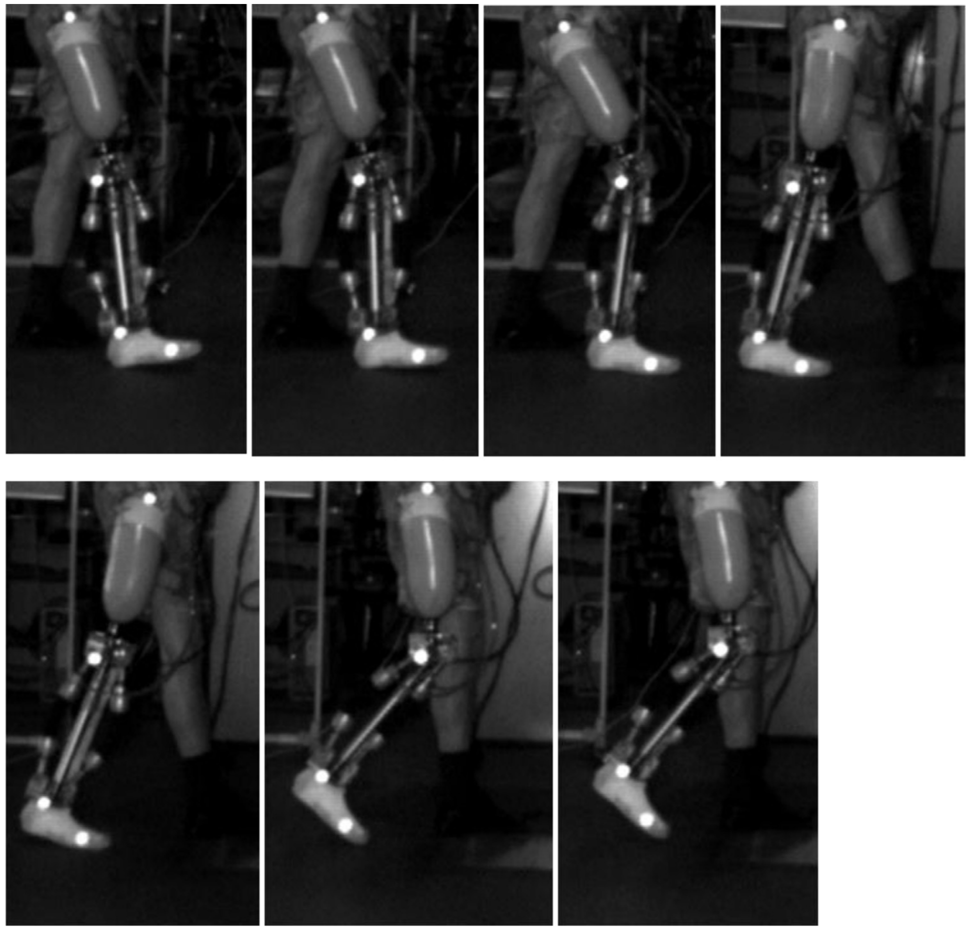

Fig. 6 – Motion of the prosthetic knee during sequence of walking modes. Each image represents the beginning of a mode in Fig. 3.

respectively. The EMG activities are step function approximations to the intramuscular EMG data reported by Perry [20]. As can be seen from these figures, the function of artificial muscles is similar to that of biological ones. That is, from initial contact to mid stance, both the artificial extensor and vasti muscles are active. The maximum activity of both muscles happens at the end of loading response when the body weight is transferred onto the forward limb. However, from mid stance to pre-swing (approximately from 20 to 40 percent of gait cycle) the artificial extensor is active, but vasti muscles are not. This is possibly caused due to the inability of prosthetic foot to emulate the dorsiflexion of the biological one. During normal walking, the knee extension after loading response happens passively as a reaction to forces which move the body forward over the supporting foot [20,21]. The prosthetic foot, which is a SACH type, cannot dorsiflex after heel strike, as it happens in healthy foot. As a result, the forward progression of the ground reaction force line does not happen fast enough to cause knee extension after loading response, and the artificial knee extensors should remain active throughout mid stance to inhibit knee collapse. At terminal swing, both the vasti muscles and artificial extensor are active, which prepares limb for heel strike. Similar to short head of biceps femoris, the artificial flexor is activated during initial to mid swing.

Fig. 6 shows a sample sequence of prosthetic knee motion, and Fig. 7 shows the angles obtained from three gait analysis trials. As it can be seen in these figures, the powered prosthetic knee is capable of emulating the kinematic features of the normal biological knee. Specifically, it is capable of providing knee flexion during stance. In order to prevent knee collapse, the transfemoral amputees avoid knee flexion when they wear

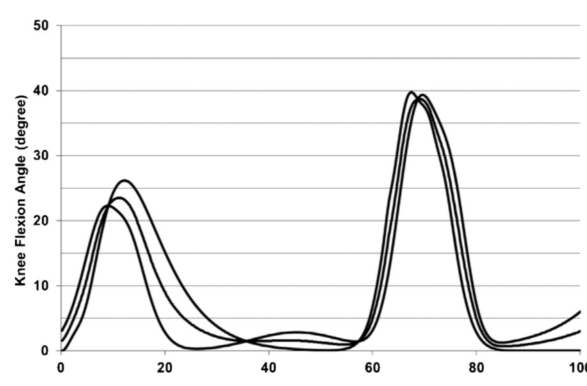

Fig. 7 – The prosthetic knee angle recorded in the gait analysis experiment. These curves show the results obtained from three trials of gait analysis.

most of passive prosthetic knees [21], which possibly leads to higher energy expenditure. The ability of providing flexion during stance with a powered knee may lead to lower energy expenditure in comparison to the passive prosthetic knees. However, in comparison to the biological knee, the prosthetic knee stance flexion is relatively large. This may be due to the inability of the prosthetic foot ankle to emulate the plantar flexion/dorsiflexion moments of biological ankle during loading response, which obligates the amputee to flex the prosthetic knee more, until foot flat is achieved.

The current drawback with using artificial pneumatic muscles is that they need a pressure source, and consequently the prosthesis will be tethered. However, new technologies have been invented in which an onboard component of the prosthesis can deliver the required pressure to the prosthesis [10,24]. The electronic system can be designed and constructed to be implemented as an onboard part of the prosthesis, as well.

In future work, the control strategy should be linked with the body. In its current version, the prosthesis is controlled through signals that are sent by controller (PC), but an interface between body and prosthesis may be designed to control the prosthesis. Some possible methods for that interface include surface electromyographic interface, implantable peripheral nervous system interface, and implantable central nervous system interface [25].

In this paper, a SACH foot was used along with the constructed powered knee. As mentioned above, this foot cannot mimic the dorsi/plantar flexion of the biological knee. Regarding the importance of this motion during normal walking [20], to emulate the normal gait as much as possible, the prosthetic foot should be able to mimic this motion as well. Design and construction of a powered ankle joint that is able to do so, is under way by the authors.

4. Conclusion

In this paper, we presented a novel biomimetic prosthetic knee joint powered by pneumatic artificial muscles. These muscles look and behave mechanically similar to biological muscles and can deliver the required torque to the prosthetic knee. Test results showed that the powered knee can produce a to a degree gait characteristics that are similar to normal gait. The findings of this study show that pneumatic artificial muscles are worthwhile candidates to consider further investigation to produce new generations of prosthetic limbs that emulate the biological ones in both appearance and functionality.

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Appendix A

In this section the mathematical equations governing the lower extremity model are described. For normal hip and knee angles, the muscle forces were calculated to produce normal torques at these joints i.e. the equation below was enforced:

$$\begin{bmatrix} \tau_H \\ \tau_K \end{bmatrix} - \begin{bmatrix} (\tau_H)_{\text{exp.}} \\ (\tau_K)_{\text{exp.}} \end{bmatrix} = 0.0 \quad (1)$$

where $\boldsymbol{\tau}_H$ and $\boldsymbol{\tau}_K$ are the sum of calculated torques of all muscles around hip and knee joints, respectively, and $(\tau_H)_{\text{exp.}}$ and $(\tau_K)_{\text{exp.}}$ are recorded biological torques of the joints. To compute the torque of each muscle, its force was multiplied by its moment arm about a joint. The moment arm was calculated using the partial velocity method [16].

Since in this project, the prosthetic knee was designed for normal walking, each muscle can be assumed as an ideal force generator [19]; therefore, the force exerted by each muscle was:

$$F^m = a^m F_0^m \quad (2)$$

where F^m is the force applied by each muscle, F_0^m is the maximum isometric force of each biological muscle (taken from Delp [16]), and a^m is the activation of each biological muscle. For artificial muscles, F_0^m is the maximum force that each muscle can apply, and a is a coefficient multiplied by its maximum force. For all muscles, the value of a^m is bounded between 0 and 1.

Since the number of unknowns (muscle forces) was larger than the number of equations, an optimization approach was followed. The optimization criterion is based on the fact that during normal walking, forces between muscles are distributed so that the normal stress at each muscle is minimized [19]. Mathematically the criterion is to minimize the function below:

$$J = \sum_{m=1}^{MN} (a_m)^2 \quad (3)$$

where J is the performance criterion and MN is the number of muscles. Using MATLAB programming language, Eqs. (1)–(3) were solved numerically to obtain muscle forces.

Calculating the forces of artificial muscles, the pressure inside each of them was determined. Festo artificial muscles exert force when they are pressurized. To control the force that they exert, the relation proposed by Tondu and Lopez, which includes parameters that can be easily measured, was used [22]:

$$F = (\pi r_0)^2 P[a(1-\varepsilon)^2 - b] \quad 0 \le \varepsilon \le \varepsilon_{\max} \quad (4)$$

in which r_0 is the initial artificial muscle radios, P is pressure and

$$\varepsilon = \frac{l_0 - l}{l_0} \qquad a = \frac{3}{\tan^2(\alpha_0)} \qquad b = \frac{1}{\sin^2(\alpha_0)}$$

where l_0 is muscle initial length, l is muscle length, and $\boldsymbol{\alpha}_0$ is the muscle mesh wave angle.

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