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Motor Electrical Damping for Back-Drivable Prosthetic Knee

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Abstract— **The paper presents a model and analysis of a back-drivable knee prosthesis. In this context, the investigation into the design, modelling and analysis of a back-drivable semi-active prosthetic knee is presented. A mathematical model has been developed for evaluating the electrical damping characteristics of the DC motor in passive mode. The analysis shows that a single actuator could be suitable to work in active mode to provide mechanical power and in passive mode as a damper dissipating energy.**

Keywords- **Prosthetic Knee; Transfemoral Amputee, Electrical Damping.**

I. INTRODUCTION

Every year, thousands of people around the world lose their lower limbs due to circulatory and vascular problems, complications associated with diabetes or cancer, and trauma. It is difficult to obtain the exact number of lower limb amputees worldwide as many countries do not give attention to amputees' records or to the cause of the amputation [1]. In the UK, the total number of new referrals to the UK's prosthetics service centres is about 5,000 per year according to National Amputee Statistical Database (NASDAB) [2].

The loss in mobility following amputation results in a degradation of the quality of life of the amputees as it affects many aspects of their personal and professional lives. Hence, there is a real need to develop an efficient prosthetic to restore lost mobility and to assist amputees in actively participating in their activities of daily living. Over the last few decades, a technological revolution in the prosthetic industry has taken place as a consequence of state-of-the-art advancements in materials, electronics, sensing, and actuators. Currently available lower limb prostheses can be divided into three main groups: *purely mechanical*, *actively damping controlled* and *actively powered* prostheses.

Purely mechanical prostheses depend only on mechanical components such as linkages, locking mechanisms and passive hydraulic/pneumatic cylinders while *actively damping controlled* controls the damping/braking of the knee flexion and extension. This damping effect is generated either by a hydraulic, pneumatic swing control, magnetoroheological fluid as in the REHO knee , or combination of hydraulic stance and pneumatic swing control as in the Hybrid Knee [3]. More advanced intelligent actively controlled prostheses were recently presented, such as the Orion microprocessor knee [4] and the Genium microprocessor knee . These advanced prostheses use a wide variety of sensors to measure the load transfer and the knee angle in order to determine the knee damping/braking requirements during stance phase and swing phases. However, these prostheses cannot provide the positive power required during some tasks or walking phases as for early push off during level ground walking and ascending stairs. *Actively powered prostheses*, such as the Victhom knee [5], commercially known as the Power Knee [6] and distributed by Ossur are fully actuated. These prostheses are powered using either DC motors [7-9], or pneumatic actuators [10].

The passive dynamic walking machines which was introduced in 1990 by McGeer [11] will help us to understand how above-knee amputees with purely passive or actively damping controlled prostheses can walk by controlling the movements of the residual limbs. These passive walking systems are more efficient than powered bipedal walkers as their movements are sustained by the dynamic swing of the limbs rather than powered actuators. The consequence is that completely passive dynamic walking machines powered only by gravity can walk like humans on modest inclines and with a small initial impulse providing an excellent natural gait on slopes without using actuators and relying solely on gravity, inertia and energy transfer between the segments of the walking machine. This produces very energetic and efficient walking based on just the machine dynamics without the need for a complex control or actuation system.

Some researchers developed semi-powered bipedal walkers based on passive dynamic walkers with just a little additional energy provided to the joints to enable walking on level ground [12]. These types of robots walk more naturally, use less energy, and require less control in comparison to fully powered walking robots. This shows that amputee also can drive the prosthetic knee via energy transferred from residual limbs to the prosthetic knee and produces movement. The amputee's hip is thus considered the main engine and power source for voluntary control of the prosthesis. However, this requires more metabolic energy and mental effort in comparison to healthy subjects.

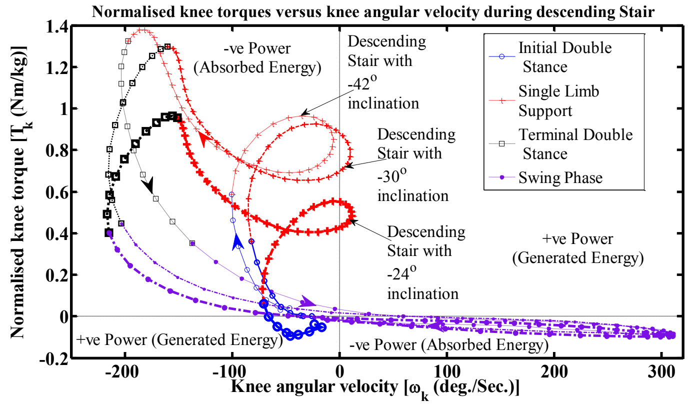
This paper introduces the conceptual framework and methodology for developing a back-drivable prosthetic knee with the capability of using DC motor as either an actuator or as a damper when it is required.

II. BIOMECHANICAL CONSIDERATIONS FOR THE DESIGN

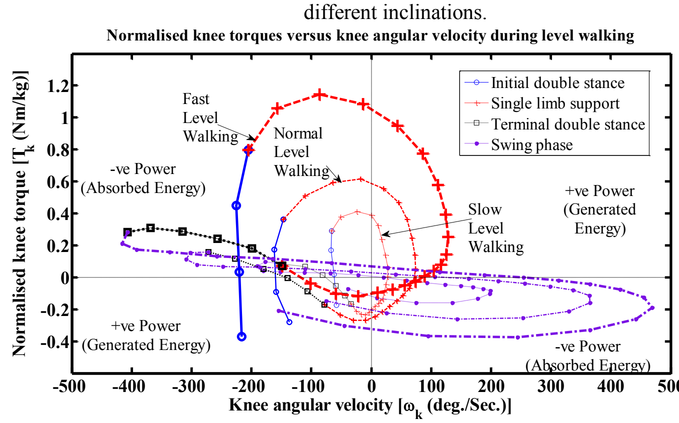
The proper selection of the appropriate actuator for the system under consideration is a significant issue because if the actuator power is underestimated, the prosthetic knee will not be able to deliver the required motion. However, if the actuator power is overestimated, the prosthetic knee will be heavier. Hence, it is required to accurately estimate of the total energy consumption, maximum peak torque, continuous rated torque, and maximum velocity for a healthy human knee joint during daily activities, such as walking on level ground at different speeds and descending/ascending stairs with different inclination angles. Normative clinical gait data for the knee joint of healthy subjects from previous research works [13, 14] was used to calculate the knee performance requirements. The ascending and descending stairs data in [13] were collected from 10 healthy male subjects for 42°, 30°, and 24° inclination angle of the staircase. The level ground walking data in [14] were collected from 19 healthy subjects for slow, normal walking speeds and from 17 healthy subjects for fast walking speeds.

In order to know when a driving or a damping torque is required during daily activities relations between the normalised knee torque (T_k) (Nm/kg) and the knee angular velocity (ω_k) were derived as shown in Figure 1 based on the gait data from [13, 14]. It is observed that most phases of stair descent produce negative energy as shown in Figure 1a, which means that this energy should be recovered by regenerative braking or dissipated by damping or by mechanical impedance. The regenerative braking might help in energy regeneration by converting the mechanical energy into electrical energy instead of dissipating it into heat by using mechanical braking or damping.

It is clear that the greater the staircase inclination angle, the more negative energy is present which needs to be absorbed. However, very small portions of the single limb support and initial double stance phases require positive energy, which means that an actuation source is required at these times. On the other hand, all the phases in stair ascending except knee extension in the swing phase require positive energy, which should be provided by an actuator. This indicates that a damping effect is an important issue during descending stairs, but the actuation effect is more important during ascending stairs. At fast walking speeds, the required energy is higher than for normal level ground walking. It is noticed that most of the phases generate negative energy during level ground walking except for very small portions of the initial double stance, single limb support, and swing phases as shown in Figure 1b. This knee torque-velocity diagram helps to decide when the regenerative braking/damping or actuation source is needed and at which sub-phases of the daily living activities.



(a) Knee torque-angular velocity diagram during descending stairs with different inclinations.



(b) Knee torque-angular velocity diagram during level walking at different speeds.

Figure 1: Knee torque-angular velocity diagram during descending stairs and level ground walking.

Based on these data, it is noted that the maximum extreme knee flexion angle is 104° for stair ascending. Also, that the total net energy is negative for level ground walking and descending stairs while it is positive for ascending stairs. Also, it was noticed from Figure 1b that positive power assistance is required only during some sub-phases of level ground walking, and that the majority of the phases require either damping or regenerative braking effect.

III. MECHANICAL DESIGN

The movement of human joints is affected by the action of muscles that are generally arranged in opposite pairs. As one muscle contracts and becomes shorter, the other relaxes and becomes longer. Human muscles are the main internal actuators for human movement and have both active (contractile component) and passive components (dampers and springs) according to the Hill model [15, 16]. The knee muscles perform concentric contractions when the knee accelerates and perform eccentric contractions when the knee decelerates to brake. In early swing and terminal swing, the knee muscles perform eccentric contractions to brake/damp the knee velocity [17]. This means that the muscle provides two contractions modes concentric (provide assistance to accelerate the movement) and eccentric contractions (provide resistance, braking, or damping to decelerate the movement). The proposed actuation mechanism should behave as closely to a human muscle as possible, but it is difficult to achieve the same performance as the natural muscle due to the limitations of the currently available technologies related to actuator size to power ratio.

To determine which actuator type and mechanism is suitable for developing the back-drivable prosthesis, a comparison between different types of actuation sources including electrical, hydraulic, and pneumatic was carried out. This comparison was based on power/weight ratio, bandwidth, controllability, operating speed, noise, damping/braking effect, energy recovery, overall system efficiency, maintenance, pollution, and indoor/outdoor activities. This comparison showed that electrical sources are more suitable for the prosthetic knee, as these are easy to use in indoors/outdoors applications and have higher efficiencies and bandwidth. Moreover, the regenerative braking concept is easily implemented with electrical sources. However, DC motors have high speeds and small torques, so a transmission mechanism is required to reduce the speed and increase the applied torque.

Among transmission mechanisms, the ball screw has the highest efficiency in both forward (driving) and reverse (back-driving) directions. The back-driving efficiency is related to the driving efficiency as shown in equation (1). Therefore, if the forward efficiency is less than 50%, the transmission mechanism cannot back-drive and it is referred to as a self-locking mechanism. Hence there is a need to select a high forward efficiency mechanism in order to back-drive with less resistance. A closed kinematic chain mechanism was developed with a low friction and high lead angle ball screw to back-drive easily with high efficiency.

$$\eta_{back-drive} = 2 - \frac{1}{\eta_{forward}} \tag{1}$$

Where:

$\eta_{back-drive}$, back-driving or reverse efficiency.

η_{forward} , driving or forward efficiency.

In the closed loop configuration shown in Figure 2, the motor (R) rotates the ball screw and the nut (S) moves linearly. This linear movement of the nut (S) drives the prosthetic knee due to the thrust force on the nut. On the other hand, if a thrust force is applied to the nut due to the rotation of the knee under effect of dynamic coupling or external moments/forces, the nut undergoes a linear movement to back-drive the ball screw and the brush DC motor. The efficiency of the selected ball screw is 95% and is not changed with applied loads as in a harmonic drive, but it varies with the screw's helix angle, which is a function of both the screw diameter and the screw pitch (p). According to equation (1), the ball screw exhibits different efficiencies in driving and back-driving directions with the driving efficiency having a higher efficiency than that of back-driving. In this mechanism, the knee's torque and speed varies relative to the mechanism transmission ratio (r) at constant motor's torque and speed. According to the geometry of this closed loop configuration, the transmission torque arm (r) can be calculated based on the mechanism geometry shown in Figure 2. By applying the sine and cosine rules in ΔABC, the equation (2) is found:

$$r = \frac{xL_1}{L_{screw}} sin\beta = \frac{xL_1}{\sqrt{L_1^2 + x^2 - 2xL_1 cos\beta}} sin\beta \tag{2}$$

The first step in accomplishing an efficient design is to find all possible assembly modes and the significant parameters affecting mechanism performance. The assembly modes are investigated by differentiating the transmission arm expression of equation (2) relative to β as follows:

$$\frac{\partial r}{\partial \beta} = \frac{xL_1 \cos\beta}{\sqrt{L_1^2 + x^2 - 2xL_1 \cos\beta}} - \frac{x^2 L_1^2 \sin^2\beta}{(L_1^2 + x^2 - 2xL_1 \cos\beta)^{3/2}} \tag{3}$$

In equation (3), β is the only variable that changes with respect to prosthetic knee rotation (θ_k). Hence, $\partial r / \partial \beta = 0$ with value of the maximum transmission arm (r_{max}) is used to find β^* which defines the angle ∡CAB for the maximum transmission arm (r_{max}). Solving equation (3), there are four solutions, indicating that the mechanism could be assembled in four different configurations. Based on analysis by the authors in previous research [18, 19], the assembly modes shown in Figure 3 and Figure 4 are selected to provide high transmission profile (transmission arm) with respect to the changing of the knee angle. The non-linear transmission ratio of this mechanism was designed to provide the mechanism with high torque and low speed at small knee angles, which occur during the stance phase, while providing low torque and high speed at large angles, which occur during the swing phases.

The essential machine elements of the mechanism are shown in Figure 3 and Figure 4. These elements are summarised as: DC motor, transmission mechanism, bearings, ball screw and timing belt arrangements. The prosthetic knee mechanism consists of seven main links, which are: the link that is attached to the amputee's socket, the carbon fibre frame, the ball screw, the ball screw nut, the ball screw bearing, the motor and the timing belt. This prosthetic knee uses a 42V DC motor which drives the ball screw through a 1:1 timing belt. The DC motor is not connected directly to the ball screw for three reasons. The first is that the DC electrical motor bearing supports large radial loads and very small axial loads. The second is that the effective screw length should be as small to produce a better transmission ratio as described in [18, 19]. Therefore, if the motor was connected in series with the ball screw arrangement, the overall effective screw length (L_{screw}) would increase. The third reason is that if more torque was required from the prosthetic knee, then only one timing pulley would be needed to change the reduction ratio from 1:1 to other ratios.

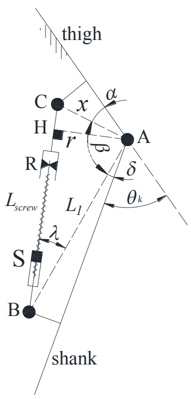


Figure 2: Schematic diagram for closed loop kinematics chain.

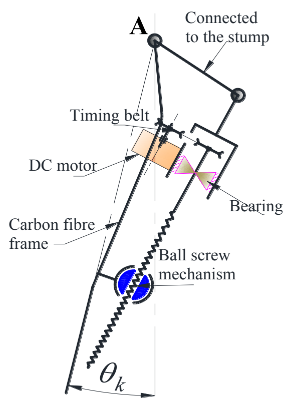


Figure 3: Schematic diagram for the proposed knee.

Figure 4: 3D CAD model.

IV. ELECTRICAL SYSTEM

The proposed control system for the prosthetic knee has two operating modes: a passive and an active one. A four quadrant amplifier based on MOSFET for example, can be used to drive a DC motor and allow it to drive or brake the load in both the forward and reverse directions. In the driving quadrants, a driving torque is applied by the motor to accelerate or sustain the load while in the braking quadrants a resistive torque is applied to decelerate and slow down the load. On the other hand, the DC motor can be used also to dissipate the kinetic energy through external fixed resistor as in dynamic braking. The other way to dissipate energy is to use a variable controlled resistor (VCR) to and hence slow down the load movement and dissipate the kinetic energy. This approach essentially involves using an external controlled resistor to dissipate the energy and to provide an adaptive electrical impedance damping torque without an expenditure of high level of electrical power. This approach was used in some haptic devices [20, 21]. Although this approach dissipates the energy to heat in the external controlled resistor, it is easier to implement. This simple approach is used in this study, for preliminary trials only, to validate the proposed concept.

The electrical damping created by controlling an external resistor is considered as a source of resistance torque for knee rotation. Specifically, electrical damping, like its mechanical counterpart, can dissipate mechanical energy and increase the rotational resistance. In order to study this effect, a simplified mathematical model of a DC motor, shown in Figure 5a, was derived using Equation (4).

Using Kirchhoff's Voltage Law 'KVL':

$$V_a(t) = L_a \frac{di_a}{dt} + i_a R_a + e_b = L_a \frac{di_a}{dt} + \frac{R_a}{k_m} T_m + k_g \omega_m \tag{4}$$

Where:

$V_a(t)$	, applied voltage to the motor's armature (**V**).
L_a	, motor's armature inductance (H).
i_a	, motor's armature current (A).
R_a	, motor's armature resistance (Ω).
e_b	, back electromotive force (EMF) voltage (V).
k_g	, generator constant (V/(rad/sec)).
ω_m	, motor angular velocity (rad/sec).
T_m	, motor torque (Nm).
k_m	, torque constant (Nm/A).

In this work, the dynamic braking concept is studied by using a controlled variable braking resistor between the motor terminals to damp and dissipate the energy as shown in Figure 5b. Equation (4) is then reformulated after the addition of the external resistor (R_{ex}) and considering $V_a(t) = 0$ gives equation (5).

In dynamic braking mode:

$$T_{elec\to m} = T_m = -\frac{k_m k_g}{(R_a + R_{ex})} \omega_m = D_{elec\to m} \omega_m \tag{5}$$

Where:

$T_{elec\to m}$	, electrical impedance torque generated by the motor (Nm).
R_{ex}	, external resistor (Ω).
$D_{elec\to m}$	, electrical damping coefficient (Nm/(rad/sec)).

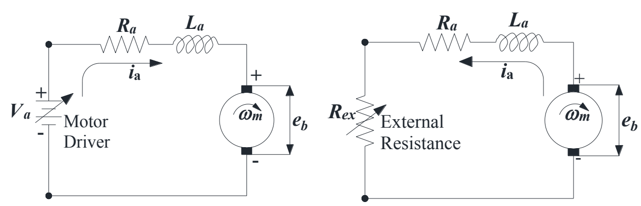
Equation (5) shows that the electrical impedance torque ($T_{elec\to m}$) behaves similarly to a mechanical damper that is linearly proportional to the motor speed (n_m). The electrical damping coefficient ($D_{elec\to m}$) changes when the external resistance between the motor terminals (R_{ex}) changes. Hence, controlling (R_{ex}) provides the system with an electrical impedance which can be used as a linear damper. However, it will be limited to the rated torque of the motor at rated speed. This relationship also shows that the electrical impedance does not have any effect when the motor terminals are open circuit ($R_{ex} = \infty$), and the electrical damping coefficient ($D_{elec\to m}$) is high at very small resistances and decreases as the resistance increases. The effect of this electrical damping coefficient on the knee impedance torques is reflected to the knee joint A in Figure 3 as follows:

$$D_{elec \to k} = \left(\frac{2\pi r}{p}\right)^2 \frac{D_{elec \to m}}{\eta_{back-drive}} \tag{6}$$

Where:

$D_{elec \to k}$, electrical damping coefficient reflected to joint A (Nm/(rad/sec)).

$\eta_{back-drive}$, back-driving efficiency from the motor to joint A.



(a) Motor driving in active mode. (b) Dynamic braking and electrical damping in passive mode.

Figure 5: Operating modes.

The electrical damping coefficient effect reflected to the knee joint ($D_{elec \to k}$) is shown in Figure 6 with respect to the knee angle and the external resistance. It is clear that the reflected electrical damping coefficient ($D_{elec \to k}$) is changing with respect to the knee angle because of the changing of arm transmission ratio (r) according to equation (2). The maximum electrical damping coefficient ($D_{elec \to k}$) occurs at $R_{ex} = 0$ with the knee angle at r$_{max}$ (θ_k^***)**. The impedance torque ($T_{elec \to k}$) that is produced due to electrical damping and the effect on knee impedance is calculated based on the following equation:

$$T_{elec \to k} = D_{elec \to k} \omega_k \tag{7}$$

Where:

$T_{elec \to k}$, Electrical damping torque reflected to joint A (Nm).

ω_k , Knee angular velocity (rad/sec).

Figure 7 shows the upper and lower boundaries of the knee electrical impedance torque ($T_{elec \to k}$) with respect to knee angle (θ_k) and speed (n_k). The upper and lower boundaries of the impedance torque are defined by the minimum (short circuit ($R_{ex} = 0$)) and maximum (open circuit ($R_{ex} = \infty$)) values of the external resistor. The impedance torque is linear to the knee angular speed at constant angle as shown in Figure 7. This electrical impedance torque does not have any effect on the knee if the motor terminals are open circuit or if the knee is stationary. This phenomenon could be helpful in using the DC motor as a linear damper similar to the hydraulic one used in the C-Leg or the pneumatic one to dissipate the negative energy generated in the prosthetic knee during walking and descending stairs. Moreover, the regenerative braking concept can be used in addition to store this energy instead of dissipating it as heat in the resistor.

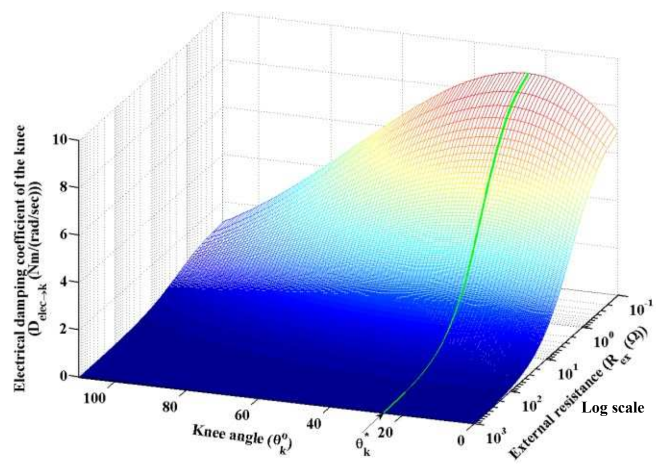


Figure 6: Changing of the electrical damping coefficient for the knee versus knee angle and external damping resistance.

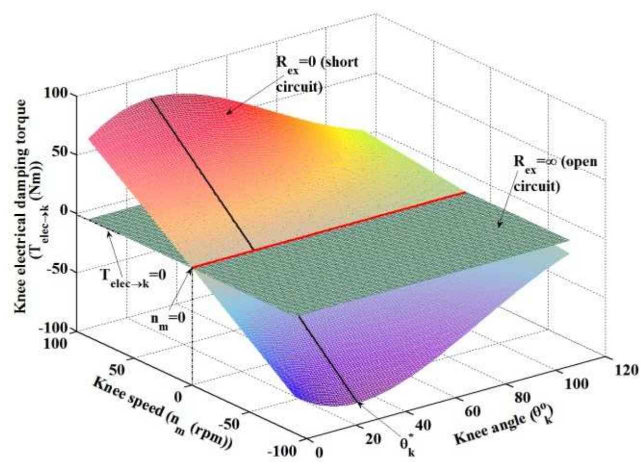


Figure 7: Boundary of the knee resistance damping torque versus knee speed and knee angle.

The ability of the DC motor when back-driven during the passive mode to work as an electrical power generator offers the potential for generating electrical energy and also dissipating the mechanical energy. This electrical impedance decelerates the knee movement in the passive mode, and is nearly proportional to the back-driving speed of the knee at constant knee angle. This concept can help to damp the movement of prosthetic knee in the passive mode to overcome the excess movement.

V. Conclusion

This paper pointed out that during the majority of level ground walking phases and while descending stairs, negative energy is generated which needs resistance damping source in order to dissipate it. On the other hand, positive energy is required during some sub-phases of level ground walking and while ascending stairs to assist human knee movement.

Hence, a back-drivable prosthetic knee was developed to maintain all the advantages of passive prostheses during descending stairs and walking on level ground while adding the capability to drive the knee easily during some sub-phases of level ground walking and ascending stairs. This will also lead to reduction of the energy consumption of both amputee and the prosthesis. Also, the DC motor of the prosthetic knee can be used as a linear damper to dissipate the negative mechanical energy in passive mode. This damping effect can be controlled by a variable controlled resistance circuit.

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