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Design of a Fully-Passive Transfemoral Prosthesis Prototype

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Abstract— **In this study, we present the mechanical design of a prototype of a fully-passive transfemoral prosthesis for normal walking. The conceptual working principle at the basis of the design is inspired by the power flow in human gait, with the main purpose of realizing an energy efficient device. The mechanism is based on three elements, which are responsible of the energetic coupling between the knee and ankle joints. The design parameters of the prototype are determined according to the human body and the natural gait characteristics, in order to mimic the dynamic behavior of a healthy leg. Hereby, we present the construction details of the prototype, which realizes the working principle of the conceptual mechanism.**

I. INTRODUCTION

In both robotics and bio-mechanics, the research in the design of lower limb prostheses is of great interest due to their crucial impact to human life. Therewith, one of the open challenges is the design of a prosthesis that provides proper mobility and energy efficiency, in terms of required metabolic energy and external actuation.

Existing lower limb prostheses can be classified as passive, controlled and powered. Passive prostheses exploit the dynamics of walking thanks to their special kinematic configuration. However, with this kind of prostheses, amputees consume a large amount of extra metabolic energy (approximately 60%) to compensate the lack of energy transfer from the lost muscles [1]. Controlled prostheses have internal actuators that control their dynamics. Therefore, these prostheses are better in terms of mobility, such as the ones presented in [2], [3]. Powered prostheses are capable of injecting power into the gait cycle in order to support push-off generation and to reduce the extra metabolic energy consumption [4], [5], [6], [7], [8]. Commercial transfemoral prostheses, as a representative of these three groups, are also available, such as the Mauch GM, 3R80 (passive), RheoKnee, Smart Adaptive and C-Leg (micro-processor controlled), and the PowerKnee (powered).

Recently, some of the design studies focused on the transfemoral prosthesis with energy storage capabilities in order to reduce the power consumption. For example, in [9] and [10], energy storage and release are provided by using an adjustable spring. Electrically powered transfemoral prosthesis, which includes a spring in parallel to the ankle motor unit, has been reported in [11]. In [12] and [13], we proposed the design for a transfemoral prosthesis mechanism based on the evaluations of the power flow in natural human gait.

In this paper, we present the mechanical design of a fully-passive transfemoral prosthesis prototype for normal walking [14]. The conceptual design is mainly based on mimicking the energetic behavior of the human gait in order to improve the efficiency in terms of metabolic energy consumption. To derive such kind of mechanism, a power analysis of human gait is exploited. By analyzing the relations between the energy absorption intervals occurring during the human gait, a working principle of the conceptual mechanism with three distinct elements is presented. Finally, we illustrate the construction details of the mechanism and the final assembly.

II. POWER FLOW IN THE HUMAN GAIT

The origin of the conceptual design of the energy efficient transfemoral prosthesis lies within the analysis of the power flow of the natural human gait [15], as shown in Fig 1. In the figure, it is possible to identify one main power generation interval G and three main absorption intervals: A_1 during pre-swing, A_2 during swing, A_3 during roll-over. Moreover, the figure highlights three instants, i.e. heel strike, push-off and toe-off, and three phases:

- Stance: the knee absorbs a certain amount of energy during flexion and generates as much as the same amount of energy for its extension. In the meantime, the ankle joint absorbs energy, represented by A_3 in the figure, due to the weight bearing.
- Pre-swing: the knee starts absorbing energy, represented by A_1 in the figure, while the ankle generates the main part of the energy for the push-off, represented by G, which is about the 80% of the overall generation.
- Swing: the knee absorbs energy, represented by A_2 in the figure, during the late swing phase, while the energy rate in the ankle joint is negligible.

Note that, in the healthy human gait, the knee joint is mainly an energy absorber whereas the ankle joint is mainly an energy generator. Moreover, there is almost a complete balance between the generated and the absorbed energy, since the energy for push-off generation, i.e. G, is roughly equals the total energy absorbed in the three intervals $A_{1,2,3}$.

III. CONCEPTUAL DESIGN OF THE PROSTHESIS

In this Section, the conceptual design of the transfemoral prosthesis is presented, as proposed in [14]. First the functions of each element are described and, then, the working principle of these elements is explained.

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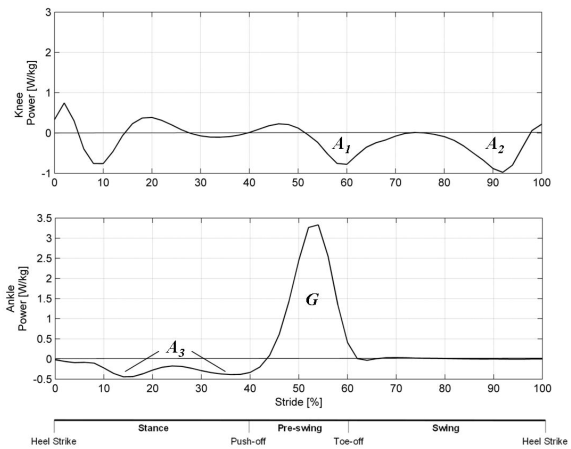

Fig. 1. The power flow of the healthy human gait normalized in body weight in the knee (upper) and the ankle (lower) joints during one stride [15]. The areas $A_{1,2,3}$ indicate the energy absorption, whereas G indicates the energy generation. The cycle is divided into three phases (stance, pre-swing and swing) with three main instants (heel-strike, push-off and toe-off).

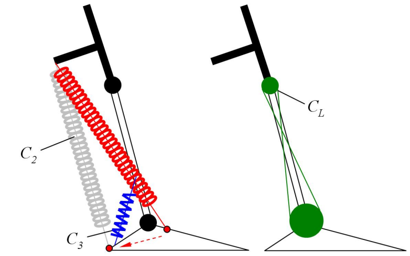

Fig. 2. Conceptual design of the proposed mechanism (given separately for better interpretation) - The design presents three storage elements: $\mathbb{C}_2$, one elastic element between the foot and upper leg; $\mathbb{C}_3$, one elastic element between the foot and lower leg (left); $\mathbb{C}_L$, a linkage system between the knee and ankle joints (right).

The proposed design relies on the energy transfer between the knee and ankle joints and, therefore, on three elements. Therefore, as summarized in Fig. 2, we introduced:

- one linkage system $\mathbb{C}_L$ that couples the knee and ankle joints, responsible for the contribution of A_1 to the ankle push-off generation.
- one elastic element $\mathbb{C}_2$, that energetically couples the knee and ankle joints. This element is responsible for the absorption of A_2 and the transfer of it to the ankle joint and for a part of absorption A_3 during stance phase.
- one linear elastic element $\mathbb{C}_3$, which is responsible for the main part of the absorption A_3.

The working principle of the conceptual mechanism with these elements has been explained and analyzed in [14] and is given briefly in the following parts.

A. Energy transfer during pre-swing phase

By analyzing the power flow on the knee and ankle joints during the pre-swing phase, the challenge for the realization of this part can be easily seen. In this phase, the knee joint absorbs the kinetic energy of the lower leg due to ankle push-off. Since the generation and absorption take place simultaneously, we employ a linkage mechanism $\mathbb{C}_L$ that energetically couples the knee and ankle joints in order to contribute to the push-off generation. By this connection, the push-off torque is transferred from the ankle joint to the knee joint with a transmission ratio tuned according to the kinematic relation between the knee and ankle joint. Therefore, this kinematic constraint realizes the natural walking kinematics while it is also providing an energy contribution, namely A_1, to the ankle push-off generation from the knee joint as a consequence of closed-loop kinematic chain.

B. Energy storage during swing phase

For the storage purpose, a coupling element $\mathbb{C}_2$ is employed during the swing phase. This element stores the kinetic energy of the lower leg (A_2) during the swing motion and transfers it to the ankle joint by changing its attachment point on the foot through a trajectory, which keeps constant the length of the spring [12].

C. Energy storage during stance phase

During the stance phase, i.e. while ankle is in dorsiflexion motion, a braking torque is applied to the ankle in order to bear the weight of the body. Instead of dissipating the energy by using a damping system, we propose a design in which the storage elements $\mathbb{C}_3$ provide the brake torque and, therefore, store the energy (A_3) during the stance phase to be delivered for the ankle push-off. $\mathbb{C}_2$ is also employed after mid-stance to build up a braking torque around the ankle joint in order to achieve the torque profile of the natural ankle joint.

At the end of the stance phase, the storage elements $\mathbb{C}_2$ and $\mathbb{C}_3$ are loaded and, therefore, are ready to release the total energy of absorption phases (A_2,A_3) for the ankle push-off.

Note that, the linkage mechanism $\mathbb{C}_L$ is only active between push-off and toe-off, i.e. in the pre-swing phase. The stance storage element $\mathbb{C}_3$ is only active during the stance phase. Therefore, there is no undesirable interference of the storage elements during the motion. Since the activation and deactivation of the storage elements can be realized when the velocities of the related joints (during walking) are zero, ideally no dissipation is present.

Fig 3 shows the animation of the 3D CAD model during one stride. Frames (1-5) represent the stance phase till push-off generation, with only the ankle elastic element $\mathbb{C}_3$ active during this interval. Frames (4-6) represent the rollover phase and at frame (6), the configuration of the sliding elastic element $\mathbb{C}_2$ changes for the transfer of stored energy during swing phase to the ankle push-off generation. At push-off (6), the linkage mechanism $\mathbb{C}_L$ is activated and the torque created around the knee joint is transferred to the ankle joint with the natural gait relations between the ankle and knee joints. After toe-off (8), the linkage mechanism disengages and the slider goes to front (9). Frame (11) shows the dorsiflexion of the ankle for a sufficient ground clearance and the start

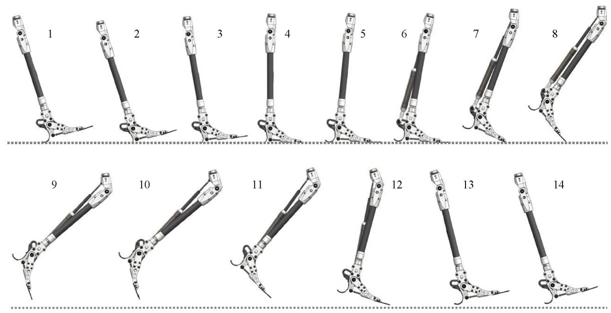

Fig. 3. Animation of one stride (from heel-strike to heel-strike) of the 3D CAD representation of the prototype.

of swing phase energy storage, which goes up to frame (14). The stride is finishing at frame (14) with heel-strike.

IV. CONSTRUCTION OF THE PROTOTYPE

This Section provides the details of the prototype construction according to the working principle.

A. The Knee and Lower Leg

The CAD drawing of the total prototype is depicted in Fig 4. The knee joint is a single axis connection, executed with two ball bearings. The upper (01) and lower part (02), are made of a 6061 aluminium alloy. A small damper is fitted inside the knee to damp the knee extension at the end of the swing phase. In case in which the swing momentum becomes too high, due to excessive hip flexion, this damper prevents hitting at full knee extension. The damper is also used to set the degrees of hyperextension during stance.

B. The Linkage Mechanism

The linkage mechanism, $\mathbb{C}_L$, is a physical coupling between the knee and ankle joints. It couples the knee flexion to ankle plantar-flexion and vice versa. The coupling is mechanically activated at push-off and deactivated at toe-off. The linkage system consist of two gears (03)-(04), and a pulley (05) which are integrated in the knee. From the knee, a steel cable (08) runs through the shank to a second pulley (09) in the ankle. This cable is flexible enough to run over the small diameter guidance wheels (06) that feed the cables into a lightweight carbon fiber shank. The steel cable can be tensioned by a fine threaded socket that slightly extends the shank by the rotation of a nut (10). The connection between the second gear (04) to the first pulley (05) is made elastic by a specially designed flange coupling, which gives the possibility to set the stiffness of the mechanism according to the user needs for normal walking.

C. The Foot and Ankle

The foot is made of two parts: the heel (11), which is rigidly connected to the shank, and the forefoot (12), which is connected to the shank by a rotational joint that form the ankle joint (9). The ankle springs $\mathbb{C}_L$ (13) are fitted in two cylinders that extend from 0° to 10° dorsi-flexion. The connection ends are executed with a length adjustable ball socket rod end and a shaft with sliding

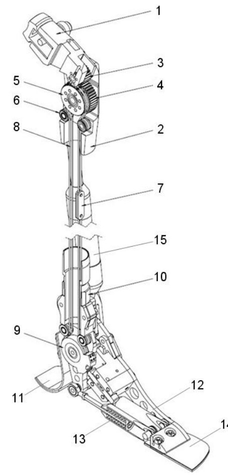

Fig. 4. CAD representation of the fully-passive transfemoral prosthesis prototype.

bushings. The fore foot and heel are made of carbon-fiber sheets. FEM calculations have been done with a flexural module of 125000 MPA related to 45° layered UD carbon-fiber material. The heel is designed to flex between 7.8 mm and 15.4 mm according to the leg angle at heel-strike. The toe (14) bends a maximum of 30°, which is derived from a ground pressure plot [16]. A peak ground reaction force occurs during push-off, corresponding to a toe angle of 30°.

For designing the ankle springs $\mathbb{C}_3$ for stance phase storage, the amount of available energy is determined from the roll-over during stance in natural gait, which is 0.13 J/Kg. This amount of energy is sufficient to compress the two springs over a length of 10 mm. Together with a moment arm length of 55 mm towards the ankle joint, the springs deliver an additional 98 Nm of push-off torque.

The lower attachment point of $\mathbb{C}_2$ is held in position by a hook. The slider is released at 6° roll-over with a small pin that pushes down the slider hook by dorsi-flexion of the forefoot. The slider trajectory is designed as adjustable by a left/right hand threaded stud. Therefore, the angle can be set such that the slider reaches the end of the trajectory at zero velocity. In case the angle is set incorrectly, a rubber is fitted to prevent hitting at the end of the trajectory. There is also a groove to hold the slider in position during push-off generation, until all the energy is released.

D. The Coupling Element

The coupling element $\mathbb{C}_2$ (15) is employed for storing energy during the swing phase and transfer it to the ankle joint for push-off generation. Furthermore, this element is also employed for creating a natural moment profile around the ankle joint during stance phase. In order to get the natural swing braking torque around the knee joint, a progressive elastic element is designed to compensate the decreasing moment arm length. In this way, the prosthesis follows the natural knee torque profile more closely and still allows to have a zero moment arm length at the end of the swing phase. Having a zero moment around the knee and ankle joints at heel-strike is considered to be important for safety purpose. The combination of design parameters i.e. spring dimensions and required stiffness values, resulted in the usage of a progressive spring fitted inside a Ø30 mm tube (15). There is also another spring into this system for knee full-flexion stop. If the user delivers a redundant hip flexion during pre-swing, instead of flexing the knee over 65°, which is a limit for normal walking that we set for the design, the knee brakes the shank by compressing this spring. Following the full-flexion of the knee, this energy will be released to the shank during swing phase.

E. Prototype Specifications

The prototype is designed for an 80 kg user with a height of 1.80 m. The total weight of the prosthesis is 2.49 kg and the specifications of the prototype are depicted in Table I.

TABLE I

SPECIFICATIONS OF THE PROTOTYPE.

	Knee	Ankle	Shank	Foot
Range [°]	$-5°/100°$	$15°/-25°$	–	–
Torque [Nm]	-16	123	–	–
Length [m]	–	–	0.47	0.26
Height [m]	–	–	–	0.07
Mass [kg]	–	–	1.44	1.05

As it can be seen from the table, the foot is relatively heavy compared to the shank and knee sections. This is a design choice in order to have the center of mass (CoM) at the distal side from the knee joint, so, many parts are placed into the foot. This is necessary to obtain the similar swing dynamics with a different weight from the natural human leg, which is around 4.9 kg for an 80 kg person. With the current weight distribution, the moment of inertia around the knee-axis of the lower leg and the foot is 0.3 kgm^2. In case the prosthesis moment of inertia is lower than the natural human leg, according to the working principle, less energy is stored during swing phase. This results in a lower amount of energy for push-off. Since most of the push-off energy is used to propel the trunk forward, but not the lower leg, a light weight prosthesis will not significantly reduce the amount of energy required for push-off. However, a light weight prosthesis is desired for the user's comfort. Therefore, we designed the prosthesis in a way that both of the objectives are achieved. The final assembly of the prototype is depicted in Fig. 5.

V. CONCLUSIONS AND FUTURE WORKS

In this study, we presented the design of a fully-passive transfemoral prosthesis prototype for normal walking as a realization of the conceptual design proposed in [14], inspired by the power flow in natural human gait. The conceptual mechanism that consists of three elements for the absorption intervals in the healthy human gait is realized according to the working principle. The design parameters of the prototype have been determined according to the human body and gait characteristics. Construction details of the mechanism have been given and final assembly has been presented. Further developments and improvements will be

Fig. 5. Final assembly of the prototype.

implemented after testing the prosthesis on healthy subjects and amputees.

REFERENCES

[1] R. Waters, J. Perry, D. Antonelli and H. Hislop, "Energy Cost of Walking Amputees: The Influence of Level of Amputation", *Journal of Bone and Joint Surgery*, vol. 58A, pp. 42-46, 1976.

[2] J.H. Kim and J.H. Oh, "Development of an Above Knee Prosthesis Using MR Damper and Leg Simulator", *IEEE International Conference on Robotics and Automation*, 2001.

[3] H. Herr and A. Wilkenfeld, "User-adaptive Control of a Magneto Rheological Prosthetic Knee", *Industrial Robot: An International Journal*, vol. 30, pp. 42-55, 2003.

[4] F. Sup, A. Bohara and M. Goldfarb, "Design and Control of a Powered Transfemoral Prosthesis", *International Journal of Robotics Research*, vol. 27, pp. 263-273, 2008.

[5] F. Sup, H.A. Varol, J. Mitchell, T. Withrow and M. Goldfarb, "Design and Control of an Active Electrical Knee and Ankle Prosthesis", *IEEE RAS/EMBS International Conference on Biomedical Robotics and Biomechatronics*, pp. 523-528, 2008.

[6] W.C. Flowers, "A Man-Interactive Simulator System for Above-Knee Prosthetics Studies", *PhD Thesis*, MIT, 1973.

[7] D. Popovic and L. Schwirtlich, "Belgrade Active A/K Prosthesis", in de Vries, J. (Ed.), *Electrophysiological Kinesiology, International Congress, Excerpta Medica*, pp. 337-343, 1988.

[8] S. Bedard and P. Roy, "Actuated Leg Prosthesis for Above-Knee Amputees", *7314490 US Patent*, 2003.

[9] A. Rovetta, T. Chettibi and M. Canina, "Development of a Simple and Efficient Above Knee Prosthesis", *IMECE International Symposium on Advances in Robot Dynamics and Control*, 2003.

[10] M. Canina and A. Rovetta, "Innovatory Bio-robotic System for the Accumulation of the Energy of Step in a Limb prosthesis", *International Workshop on Robotics in Alpe-Adria-Danube Region*, 2003.

[11] F. Sup, H.A. Varol, J. Mitchell, T. Withrow, and M. Goldfarb, "Self-Contained Powered Knee and Ankle Prosthesis: Initial Evaluation on a Transfemoral Amputee", *IEEE International Conference on Rehabilitation Robotics*, 2009.

[12] R. Unal, R. Carloni, E.E.G. Hekman, S. Stramigioli and H.F.J.M. Koopman, "Conceptual Design of an Energy Efficient Transfemoral Prosthesis", *IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2010.

[13] R. Unal, S.M. Behrens, R. Carloni, E.E.G. Hekman, S. Stramigioli and H.F.J.M. Koopman, "Prototype design and realization of an innovative energy efficient transfemoral prosthesis", *IEEE RAS/EMBS International Conference on Biomedical Robotics and Biomechatronics*, 2010.

[14] R. Unal, S.M. Behrens, R. Carloni, E.E.G. Hekman, S. Stramigioli and H.F.J.M. Koopman, "Towards a Fully Passive Transfemoral Prosthesis for Normal Walking", Internal Report, University of Twente, 2011.

[15] D.A. Winter, *The Biomechanics and Motor Control of Human Gait: Normal, Elderly, and Pathological*, University of Waterloo Press, 1991.

[16] N. Hayafune, Y. Hayafune, H. A. C. Jacob, "Pressure and force distribution characteristics under the normal foot during the push-off phase in gait", *The Foot*, vol. 9, pp. 88-92, 1999.