Ankle-Knee Prosthesis with Powered Ankle and Energy Transfer for CYBERLEGs α-Prototype.

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***Abstract*—Restoring natural walking for amputees has been increasingly investigated because of demographic evolution, leading to increased number of amputations, and increasing demand for independence. The energetic disadvantages of passive prostheses are clear, and active prostheses are limited in autonomy. This paper presents the simulation, design and development of an actuated knee-ankle prosthesis based on a variable stiffness actuator with energy transfer from the knee to the ankle. This approach allows a good approximation of the joint torques and the kinematics of the human gait cycle while maintaining compliant joints and reducing energy consumption during level walking. This first prototype consists of a passive knee and an active ankle, which are energetically coupled to reduce the power consumption.**

I. Introduction

Worldwide there are many people who fall victim to a lower limb amputation for various reasons, such as cardiovascular diseases, trauma, malignancy or congenital limb defects. The number of people who undergo an amputation has risen over the past decades, also because of the demographic evolution [1]. It is clear that thigh-level amputations, known as transfemoral amputations, are extremely challenging for the amputee, as two joints are lost. This increases not only the metabolic energy that is needed to walk and to perform other tasks like climbing stairs, but also the cognitive load intrinsic to those tasks. A solution that is currently selected for a considerable number of transfemoral amputees is to combine an ankle prosthetic module and knee prosthetic module, as almost no integrated ankle-knee prostheses are available on the market.

For the ankle module, there are a lot of options ranging from conventional feet, to so called "Energy-Storing-and-Releasing"-feet or ESR-feet and active prostheses. Conventional feet like the SACH-foot [2] are able to restore basic walking capabilities by improving stability and providing a roll-over to make walking more comfortable. The ESR-feet like the Seattle foot and many more [3], [4] are different from the conventional feet because they store energy early in the gait cycle and release this energy at push-off when it is needed to move the body forward. This energy, however, is not enough to closely approximate healthy ankle gait because ankles generate work during a normal gait cycle. More advanced passive prostheses are the CESR-foot, which stores the ground impact energy and releases it at push-off [5], and the AmpFoot 1.0, which provides a different stiffness during loading and during push-off, to better approximate the healthy gait [6]. Active prostheses use motors or pneumatic muscles [7] to provide this additional energy. Examples are the MIT Powered Ankle-Foot prosthesis [8], the SPARKy-prostheses [9], [10] and their commercially available counterparts, the iWalk BiOM foot [11] and SpringActive Odyssey [12], and the VUB AmpFoot 2.0 [13].

Current knee prostheses are generally passive devices, designed as dampers which dissipate some of the energy available in the knee. Because the knee joint primarily dissipates energy during the gait cycle, they are able to decently approximate human walking kinematics. The few powered prosthetic knee modules use a motor to approximate the human gait, either by changing the damping or by providing extra energy at the joint [14]. This additionally makes it possible to walk on slopes or take stairs, conditions where the knee produces energy rather than simply dissipating it. One issue with these modules is that they also consume energy during normal level walking.

There are no integrated transfemoral prostheses on the market, and only a few in research phase. Fully passive prosthesis are being investigated in a few research labs such as the University of Twente [15], and the VUB where a passive device with energy coupling between the ankle and the knee that allows knee flexion during weight acceptance is being studied [16]. The idea behind this energy coupling is to use part of the energy dissipated in the knee joint (13J for an 80kg person) to make up for the energy need in the ankle (-18J). A fully active prototype with a powered knee and powered ankle has been developed at the Vanderbilt University [17]. No efforts have been made to combine powered joints with the idea of energy recuperation from the knee to other joints. Investigation in this area could be beneficial, as combining these aspects maintains energy efficiency and gains the advantages of active prostheses, such as natural slope walking. This research fits in the CYBERLEGs project.[1] The novelty of the first prosthesis in this project, described in this paper, will be that it has an active ankle joint and energy transfer from the knee joint to the ankle.

Section II describes the working principle and simulation of the ankle actuator, Section III does the same for the knee mechanism and in Section IV the actual realization of the concept is explained.

II. Ankle concept

In this section, the concept and architecture of the ankle joint of the prosthesis is described. Biomechanical data from

[1]The CYBERnetic LowEr-Limb CoGnitive Ortho-prosthesis. The project aims for the development of an artificial cognitive ortho-prosthesis system for the replacement of the lost lower limb of dysvascular transfemoral amputees and to provide assistance to the sound limb. The final prototype will allow the amputee to walk, use stairs and move from sit-to-stand and stand-to-sit with limited cognitive and energetic effort. www.cyberlegs.eu

the literature shows that a healthy ankle produces energy during walking. To closely approximate this behavior with a prosthesis, the ankle must also have an energy source. An obvious solution to this is to connect an electric actuator to the joint.

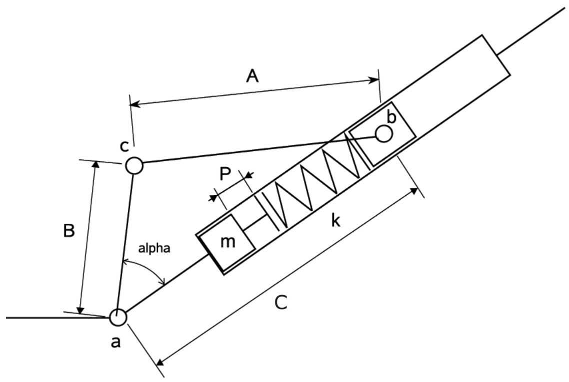


Fig. 1: Configuration of a MACCEPA using rigid linkages. To see the link with the actual ankle prosthesis design, see Fig. 10

A MACCEPA actuator, a variation on previous designs of this variable compliance actuator used for biologically inspired robots [18], was designed for the ankle joint. The advantage of this actuator is that the compliance can be varied without changing the equilibrium position of the actuator. Also, the compliance of the actuator enables it to buffer the energy and thus provide higher peak power outputs at the joint than if the joint were directly actuated. The difference with respect to earlier designs is that rigid bars have been used instead of cables. With this design, problems due to the cables that appeared earlier, such as attachment and durability issues, were avoided.

The design shown in Figure 1 consists of two rigid linkages (ac and cb), one sliding bushing at b and three rotating points a, b and c, a linear spring with stiffness k and a precompression mechanism for the spring. The length P represents the precompression of the spring. The angular position of lever arm ac (the angle α) is what will be driven by the position equilibrium motor located at point a. By changing this angle, the length C changes and the spring is compressed or decompressed, causing a torque around point a, which represents the ankle joint. The length C can be written as function of α, B and A:

$$C(\alpha) = B\cos\alpha + A\left[1 - \left(\frac{B}{A}\sin\alpha\right)^2\right]^{1/2} \tag{1}$$

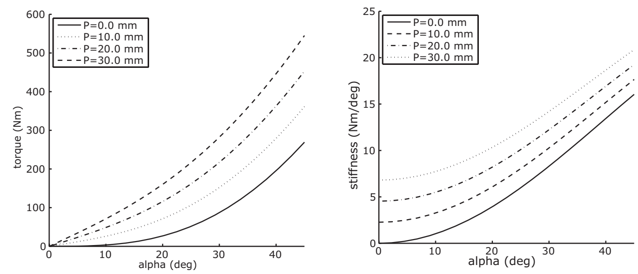
It is clear that when α=0, C=B+A. The torque around point a, caused by the actuator is equal to the product of this length C and the force acting on b, perpendicular to ab:

$$T(\alpha) = C(\alpha) f(\alpha, P) \tag{2}$$

This force is a function of the angle α but also depends on the precompression length P:

$$f(\alpha, P) = \frac{kB(P + A + B - C(\alpha))\sin\alpha}{A\left[1 - \left(\frac{B}{A}\sin\alpha\right)^2\right]^{1/2}} \tag{3}$$

With these equations the torque-angle characteristic and the stiffness as a function of the moment arm angle can be calculated. These change when the pretension value changes as can be seen in Figure 2. It is clear that the pretension does not change the rest position: the torque is 0 when the angle α is 0° for all pretension values (α=0° is equilibrium angle), while the stiffness increases with pretension.



(a) (b)

Fig. 2: Torque and stiffness of the developed $MACCEPA$-design.

A few similar concepts and configurations were investigated, but the one described is relatively easy to fit into the shape constraint of an average healthy ankle and the torque characteristic can be reached with acceptable forces and link lengths. For the simulation of the ankle joint, the required motor power necessary to follow the desired torque trajectory was calculated by first identifying the required position of the moment arm, ac, at every moment in time by means of Equation 2. The desired torque trajectory was determined from biomechanical data of healthy gait that is shown in Figure 3 [19].

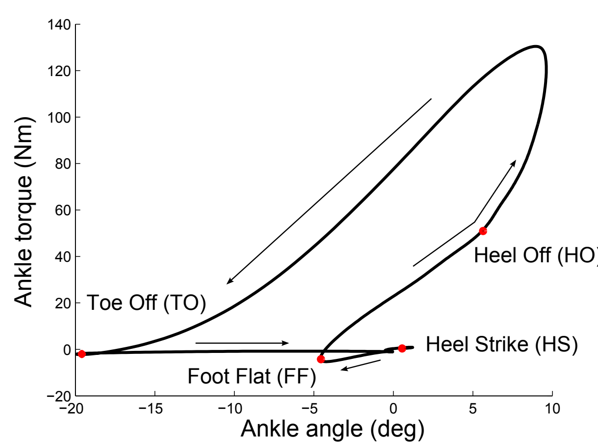


Fig. 3: Ankle kinematics for a healthy 80 kg human [19].

From the desired moment arm angle trajectory, the desired moment arm velocity can be calculated. Multiplying this moment arm speed with the joint output torque at that time gives a value for the desired motor output power. By using an iterative method varying the spring constant and the pretension length, the required peak power can be minimized. Increasing the pretension length will increase the peak power but on the other hand greatly reduces the required velocity because of the increased stiffness. The power, torque, and position characteristics are shown in Figure 4.

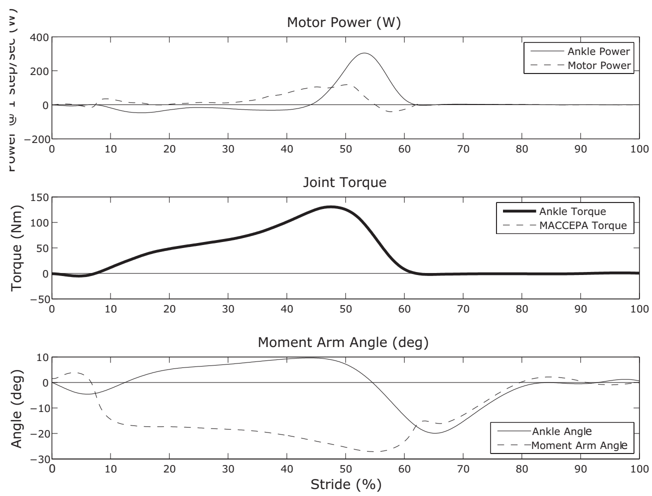


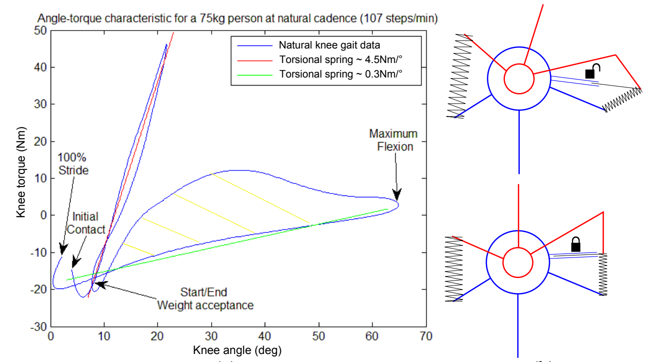
Fig. 4: Power, torque and position characteristics of the MACCEPA actuator.

III. Knee concept

A. Knee joint

In this section, the concept and architecture of the knee joint of the prosthesis is described. During normal walking of an able-bodied person, a knee joint primarily dissipates energy [19]. One of the periods of time where energy dissipation occurs is when the knee must slow the lower leg at the end of the leg swing phase. In most passive knee-prostheses, this energy is dissipated by using a damper. If this energy is stored, for example in a spring, it is not dissipated and it can be used in an other phase of the gait cycle. This is the purpose of the energy transfer mechanism in the knee joint of the prosthesis. The stored energy will be used in the ankle to reduce the torque that the motor has to provide. Apart from this, the prosthetic knee should of course also provide a good approximation of the torque behavior of a healthy knee.

The knee behavior can be subdivided in two parts: first the weight acceptance phase, characterized by a high joint stiffness, and the flexion phase, where there is a high knee flexion of about 60° and a low torque to prevent the leg from collapsing. The knee behavior can roughly be approximated by using two springs placed between the lower leg and the upper leg. One spring can provide all of the negative torques such that the other spring will only be loaded in one direction (for example only in tension), which aids mechanical construction (like the green line in Figure 5). The stiff spring used for the weight acceptance (like the red line in Figure 5) must be disengaged after the weight acceptance phase so the knee can flex and provide sufficient ground clearance for the swing phase. This requires a locking-mechanism to unlock the spring at one side so that it no longer exerts a torque around the knee.



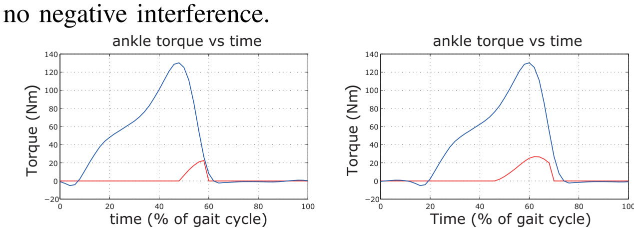
(a) (b)

Fig. 5: (a): Approximation of the knee kinematics [19] by using 2 torsional springs (the yellow shaded torque will be provided by the energy transfer mechanism and will be linked to the ankle), (b): schematic of the working principle of the weight acceptance spring (red=upper leg, blue=lower leg).

The approximation with these two springs is not ideal, as can be seen in Figure 5. Between the end of the weight acceptance and maximum flexion, a higher torque is needed around the knee joint to prevent the knee joint from collapsing at this point during stance phase. At this point, a second locking mechanism locks in another stiff spring, placed between the knee and the ankle. This energy transfer mechanism will provide the necessary stiffness at the knee and, because it is also connected to the ankle, transfer stored energy to the ankle where it can be used for push-off. The yellow shaded area in Figure 5(a) is what will be captured at the knee and transferred to the ankle.

B. Energy transfer

There is a difference in phase between the available energy at the knee and the necessary energy at the ankle. The problem with this is that the energy partially comes too late for the push-off at the ankle. This can be seen in Figure 6, where the ankle torque of healthy gait is compared to the torque from the energy transfer mechanism. In the first graph the actual situation is shown where the energy transfer is limited. The energy that can be gained in this scenario is just over 2 J per step for a person of 80 kg. The main reason for this small value (which is still a drop of 10% in energy expenditure) is that the knee flexion continues when the ankle joint starts to dorsiflex to provide ground clearance. The motor would have to compensate this unwanted torque caused by the energy transfer, reducing the energy gain even more. This is solved by unlocking the transfer mechanism early enough so there is no negative interference.



(a) (b)

Fig. 6: Healthy ankle torque (blue) and energy transfer torque without (a,red) and with late push-off (b,red). The ankle characteristic has been shifted to increase the energy transfer.

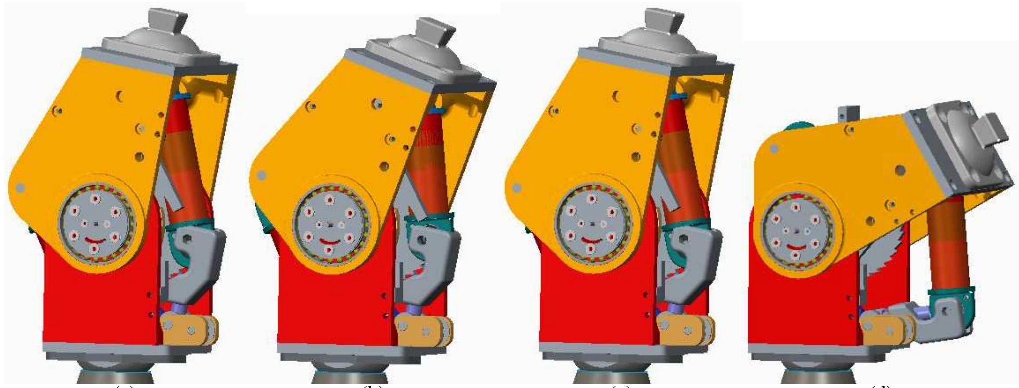

(a) (b) (c) (d)

Fig. 7: Operating sequence of the weight acceptance mechanism in the prosthetic knee. (a): Initial contact, (b): Maximum torque during weight acceptance, (c): End of weight acceptance, (d): Maximum flexion.

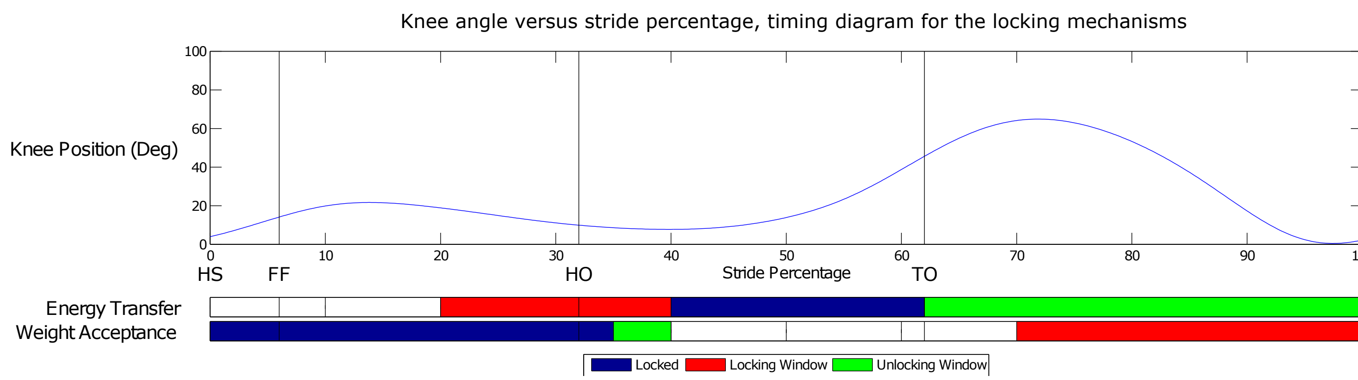


Fig. 8: Overview of the knee angles and timing of locking mechanisms.

A possible way to increase the transferred energy is to delay the ankle push-off by a short time period. This reduces the difference in phase and greatly increases the amount of energy that can be transferred. Also, because this energy is now provided at the moment where the ankle torque is the highest, the reduction in torque that the motor has to provide is bigger. In Figure 9, the reduction in power that the motor has to provide in order to match the ankle torque is shown. The power peaks are lower and there is an overall drop in energy usage of about 30 % (7 J reduction compared to a total consumption of 22 J per step).

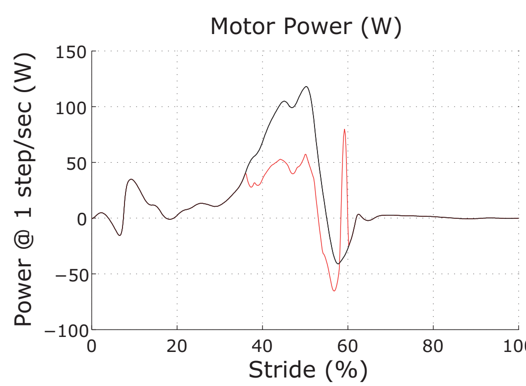


Fig. 9: Motor power production in order to match the ankle torque with (red line) and without energy transfer (black line) from the knee.

IV. Prosthesis design

The implementation of the designed MACCEPA actuator can be seen in Figure 10. The selected 200W Maxon motor and 14:1 gearhead are placed in parallel with the lower leg shank. A helical gearing with another 10:1 reduction connects the motor to the two ankle joint moment arms. The spring is placed centrally between the moment arms for a symmetrical load. The power of the motor is this high (compared to a necessary peak of about 80W in Figure 9) to be able to compare operation with and without energy transfer, and to maintain a high enough speed during the swing phase despite the high gear ratio. In the picture an elastomer spring is used, this has been replaced by a steel spring due to friction losses. The 8W precompression motor from Maxon and 1621:1 gearhead are placed below the spring. It is attached to an ACME lead screw with a 3 mm pitch which compresses the spring and is located in the ankle axis.

The prosthetic knee shown in Figure 11, has two locking mechanisms and two springs, as explained in Section III. The operating sequence is illuminated in Figures 7 and 8. In the first phase, beginning at heel strike, the weight acceptance starts. The angle of the knee joint is about 10° and the weight acceptance mechanism is locked by means of a ratchet: a cable coming from the knee axis prevents the bar mechanism with the spring to move as long as the ratchet is locked (Figure 7(a)). As the stance phase progresses, the knee angle increases

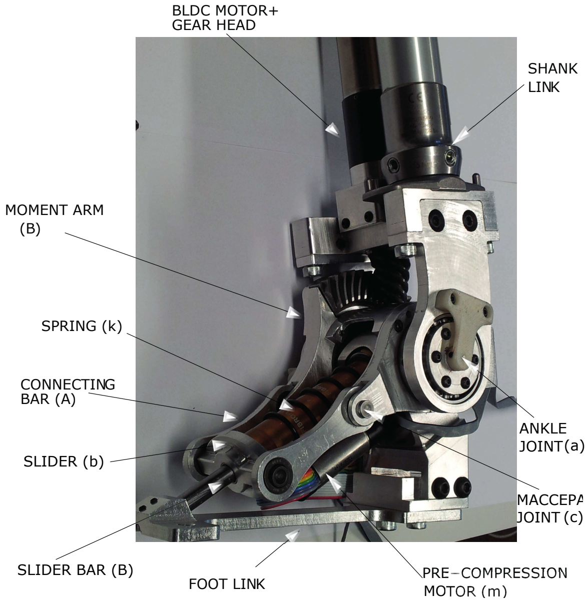


Fig. 10: Implementation of the MACCEPA actuator. Letters between brackets refer to schematic of the MACCEPA actuator in Fig. 1

to 20° and because the bar mechanism is immobilized, the spring is compressed, providing the necessary stiffness around the knee (Figure 7(b)). The angle returns to 10° and the weight acceptance mechanism is unlocked (Figure 7(c)). This can be accomplished with little effort because the ratchet is not highly loaded at this moment.

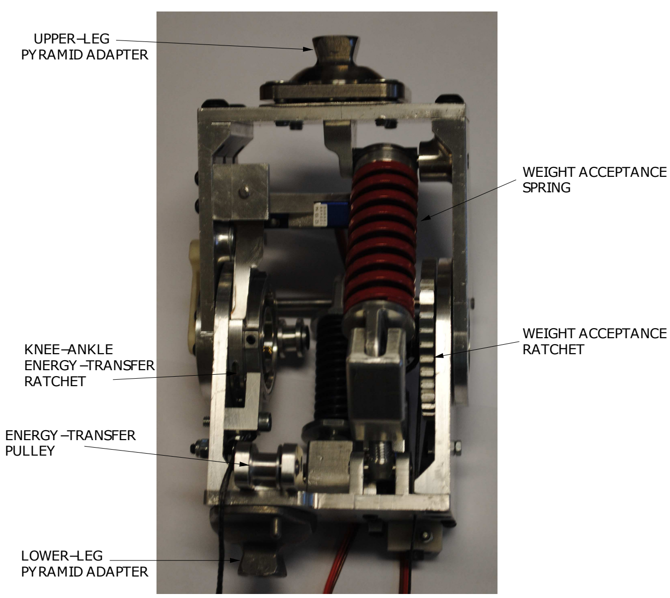


Fig. 11: Back view of the prosthetic knee with two locking mechanisms.

After this, the energy transfer mechanism is locked. This mechanism consists of two lever arms, whose positions can be adjusted, a ratchet with one locking position which can also be tuned and two pulleys to guide the cable connecting the lever arms. While the knee angle increases to 65°, the weight acceptance mechanism moves out of the way since the ratchet is not locked (Figure 7(d)). The energy transfer mechanism provides a torque at the knee, and pulls on the ankle to reduce the amount of torque that the ankle motor needs to provide (Figure 12). The ratchet unlocks automatically when the knee reaches a certain angle which can be adjusted. When the leg is swung forward, both the energy transfer and the weight acceptance mechanism return to their initial positions.

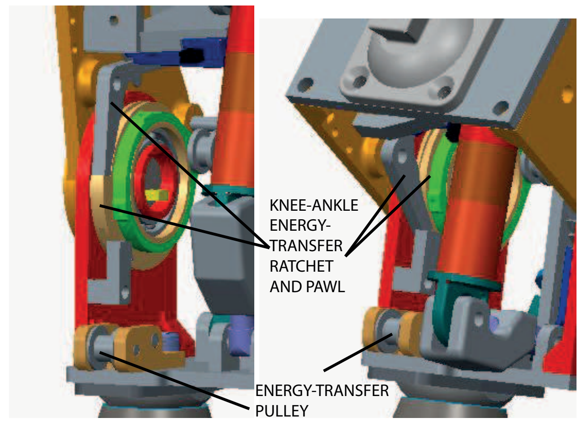


Fig. 12: Operation of the energy transfer mechanism in the prosthetic knee. Left: Beginning of energy transfer. Right: End of energy transfer.

The range of rotation for the prosthetic ankle ranges from around -45° to 60°. At the ends of this range however there is the risk of the helical gears detaching, so that the safe operating region ranges from approximately -25° to 40°. This is still sufficient compared to biomechanical human gait data [19]. For level walking, a healthy ankle rotates from -20° to about 10°. Since future versions of the prosthesis will also incorporate walking on slopes and stairs, the range of rotation is higher than this. From [20] we learn that for stair ascent the range of motion is from -10° to 23° and for descent from -27° to 28°. The range of rotation for the prosthetic knee joint is limited from 0° to 90°, since this is what is needed to be able to walk on level ground and sit down with the prosthesis. Active stair ascent and decent, where the maximum angle can increase up to 105° is not within the scope of the first prototype.

V. Conclusion

In this paper the design and realization of a novel active integrated transfemoral prosthesis with energy transfer is described. The prosthesis is part of the CYBERLEGs system and is currently being tested on test subjects both independently and as part of the system (with a pelvis module which allows weight transfer between the legs and an orthosis for the sound leg). The prosthesis provides the necessary torques at the ankle and the knee for a 80kg person walking on level ground and transfers energy from the knee to the ankle to reduce the work of the motor. The stiffness of the ankle joint can be adjusted to better fit the needs of different amputees. The total weight of the prosthesis is under 5kg, electronics included but without batteries. This is less than the weight of an actual leg and can be further reduced by optimizing the design. The prototype is fully assembled as can be seen in Figure 13 and ready to be tested in the coming months.

Fig. 13: Completely assembled prosthesis on mannequin.

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