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POSITION AND WEIGHT ACTIVATED PASSIVE PROSTHETIC KNEE MECHANISM

Tyagi Ramakrishnan
Department of Mechanical Engineering
University of South Florida
Tampa, Florida 33620
Email: tyagi@mail.usf.edu
Phone: 813-999-6669

Christina-Anne Lahiff
Asgard Kaleb Marroquin
Kyle B. Reed
Department of Mechanical Engineering
University of South Florida
Tampa, Florida, 33620
Email: kylereed@eng.usf.edu
Phone: (813) 974-2385

ABSTRACT

The human knee is a complex and robust system. It is the most important joint for human gait because of its immense load bearing ability. The loss of such an important joint often makes it difficult for a person to ambulate. Because of this and the resulting unnatural application of forces, many trans-femoral amputees develop an asymmetric gait that leads to future complications. Prosthetic knees are required to be well-designed to cope with all variabilities. There have been many prosthetic knee designs, some more complex than others. This paper describes the design and preliminary testing of a novel passive position and weight activated knee locking mechanism for use in lower limb prosthetics. This knee mechanism is designed to be a simple and economical alternative to existing knee mechanisms. The mechanism utilizes the dynamics of the user to lock the knee during stance and unlock during the swing phase. The presence of one moving component and a simple assembly makes this design a good base for customization. Results from testing the knee mechanism shows trends that are different from a normal human knee, which is to be expected. The prosthetic knee is designed to have low friction during swing of the shank and, hence, the flexion and extension angles and angular velocities are larger compared to a normal knee. The kinematics show a cyclic trend that is highly repeatable. Further refinement and testing can make this mechanism more efficient in mimicking a normal knee.

Introduction

Human gait can be defined as a synchronized and periodic advancement of each leg propelling a person forward [1]. It is a complex process involving the coordination of various muscle groups belonging to different parts of the lower extremity. The balance that holds the complex process of human gait together is diminished when a person has a limb amputated. Every part of the lower extremity contributes towards a stable gait, especially the joints. The ankle and knee joint are responsible for load bearing, articulation, and the overall dynamics of gait [2]. Hence, removing the knee and ankle joints during trans-femoral amputation severely affects the person's gait [3]. One way to counteract the changed gait pattern is to improve the prosthetic design, specifically at the knee joints. Since there are an estimated seven million trans-femoral amputees across the world [4], it is important to keep the economics in mind during the design phase so the prosthetic can be low cost and simple, both of which are met by passive knee mechanisms.

Prosthetic knees can be broadly characterized as passive and active mechanisms [5, 6]. Active mechanisms are state of the art and are designed to mimic the knee and ankle joint effectively [7]. In many comparison studies related to walking, such as stair ascent, walking on a slope, and performing ambulatory movements [3, 8, 9], active knees have shown lower metabolic strain than passive knees. Many active knees have variable settings that allow the user to adjust their prosthetic to the terrain and condition of their environment. However, all these advantages of active knees are expensive and many trans-femoral amputees have to resort to inexpensive passive knees [10].

There are five kinds of passive knee locking mechanisms, namely: manual, poly-centric, single axis, weight activated, and knee with exterior hinges [11, 12]. Manual locking mechanisms are generally used by amputees who have minimal capacity for movement, K0 – K2 in the amputee K levels (K is an arbitrary letter assigned by HCFA) [12, 13]. Amputee K levels are specified to categorize amputees on their ability to rehabilitate and is also taken into consideration when choosing a prosthesis. Manual locking allows the amputee to achieve more stability from the knee joint, since they cannot control the prosthesis in any other form due to the lack of ambulatory muscles. Poly-centric knees are a popular choice for passive knee mechanisms [14]. Poly-centric knees are generally made of 4, 5, and 6 bar mechanisms [11, 15] where the instantaneous center of the mechanism shifts during the gait cycle and locks based on the position of the shank with respect to the thigh in the gait cycle. Poly-centric knees also offer better control of the swing to the amputee. Single axis systems are simple mechanisms, but are not as commonly used as poly-centric knees. Weight activated knee mechanisms are often coupled with single axis knees to provide better locking [12]. This mechanism utilizes the user's weight to lock the knee during stance phase. The weight-actuated mechanisms often rely on links that are connected with an intricate pattern to either guide high friction surfaces to mesh or apply brakes when the weight is acted upon the system. The constant contact of the components results in high friction leading to more wear of the internal components. Knees with an exterior hinge type mechanism were used earlier in the development of prosthetic knees and they resembled an orthotic device.

Design

The position and weight activated knee locking mechanism is designed to be simple and can serve as an alternative to poly-centric and single axis knee mechanisms. The knee mechanism is designed to utilize the user's dynamics to function, which makes the knee ideal to be used by trans-femoral amputees in the K3 and K4 level. The amputees in the K3 and K4 level are more mobile and have more residual limb muscles, which means they require a prosthesis that can enable them to use their motion effectively. This knee mechanism can also be prescribed to people who have undergone knee disarticulation [16]. Because the target population has more abilities, the research study tested the knee mechanism on able-bodied subjects using the prosthetic simulator depicted in Figure 1.

The knee mechanism consists of two major components: the femur, depicted in Figure 2(f), and the shank, depicted in Figure 2(d). The femoral component consists of the housing and a spur gear rack, depicted in Figure 2(e). The knee housing holds the knee assembly together and also has slots in which bearings, depicted in Figure 1(g), can translate vertically. The shank component consists of a spur gear with a through hole at the center. Both femur and shank elements have stoppers that come in contact, at knee strike, to have the shank assume a precise locking position. The shank can be extended by means of a coupler, depicted in Figure 2(c), and extender, depicted in Figure 2(b). The roll-over shape foot, depicted in Figure 2(a), is secured to the extender and it is defined by the curve followed by the foot's center of pressure points when they are transformed from a general coordinate system to a knee-ankle based coordinate system [17]. The foot utilized for this design is a laser cut rigid piece of delrin, which is an acetal homopolymer that has a similar tensile strength as aluminum but has a lower shear strength. This curved foot allowed the design to function without an ankle. The curvature of the foot allows the person to rock forward, which simulates a downward slope and also gives the effect of plantar-flexion, although no force is generated by the foot itself. The complete assembly is fitted on an able-bodied person using a custom built knee brace, depicted in Figure 2(i).

The knee locking mechanism is designed to be a simple passive system consisting of only one moving part. Figure 3 shows the various positions the knee assumes during a gait cycle. Comparing the positions from Figure 3 to the gait cycle depicted

FIGURE 1. The prosthetic simulator used to test the position and weight-actuated knee locking mechanism.

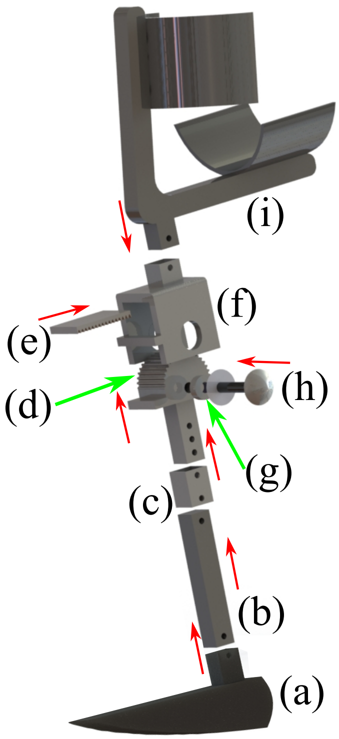

FIGURE 2. Exploded view of the prosthetic simulator: Red arrows indicate assembly direction and green arrows indicate specific components: (a) prosthetic foot based on roll-over shape, (b) extender component, (c) coupling, (d) shank with spur gear, (e) femoral spur gear rack, (f) femoral housing, (g) bearing, (h) shaft, and (i) knee brace

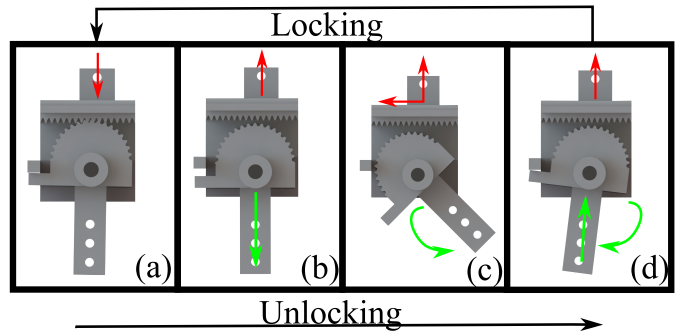

FIGURE 3. Knee positions: Red arrows indicate the application of force and green arrows indicate motion. (a) The knee is locked when the shank gear meshes with the femoral gear, which happens because of the user's weight acting on the knee. (b) The knee unlocks when the user's weight does not act on it, and the shank gear disengages from the femoral gear and slides down the slot. (c) The shank rotates about the bearing as the user swings their residual limb. (d) As the user reaches the end of their swing, the shank swings back like a pendulum and hits the stopper to assume the position for locking. The locking cycle then begins as the user applies their weight on the prosthesis.

in Figure 4 provides a better understanding of the working of the knee mechanism. The knee assumes the position in Figure 3(a) when it is locked. The locking occurs when the weight of the user is applied on the mechanism, as indicated by the red arrow in the figure, which causes the spur gear of the shank to mesh with the spur gear rack of the femur. The knee is locked while the user applies their weight on the prosthesis during stance phase as seen in Figure 4(h), (g), and (f). When the user's weight is taken of the knee mechanism, which occurs just after toe off in Figure 4(e), the shank spur gear unmeshes with the femoral spur gear rack. The slot in the femoral housing allows the bearing of the shank to translate 5 mm vertically, depicted in Figure 3(b). This marks the beginning of the swing phase for the prosthesis when the shank is free to rotate as seen in Figure 3(c) and correspondingly in Figure 4(d) and (c). The shank utilizes the motion of the user's residual limb to swing like a pendulum. When the user's residual limb reaches the extended position, the shank returns and the stopper of the shank makes contact with the stopper of the femoral housing as seen in Figure 4(b) at knee strike and in closer view in Figure 3(d). Knee strike occurs just before heel strike and the shank assumes its position to traverse back up the slot to mesh with the femur when the user applies their weight on the prosthesis at heel strike as depicted in Figure 4(a). The knee is then back to the locked position as seen in Figure 3(a). The knee strike and heel strike occur at a close interval and, hence, there is no bounce back in the prosthetic knee mechanism. This cycle continues for every stride of the gait cycle.

Results and Discussion

Testing on the knee mechansim was conducted on a single subject who is experienced with walking on the prosthetic simulator under IRB Study #Pro00016724. The knee mechanism is designed to have constant periodic kinematics during every stride. The study was conducted in the computer assisted rehabilitation environment (CAREN) by Motek Medical®, which is a state of the art rehabilitation environment consisting of a Bertec® split belt treadmill, a MOOG® motion base (*MB-E-6DOF/12/1000KG*) with six degrees of freedom (DOF), a ten-camera Vicon® (*Edgewood, NY*) infrared motion capture system, Bertec® force plates (*FP4060-08-1000*), and a panoramic display for full visual immersion. The subject's motion was captured using reflective markers placed on specific locations on the subject's body. For this study, we utilized the lower limb human body model [18] to place the reflective markers.

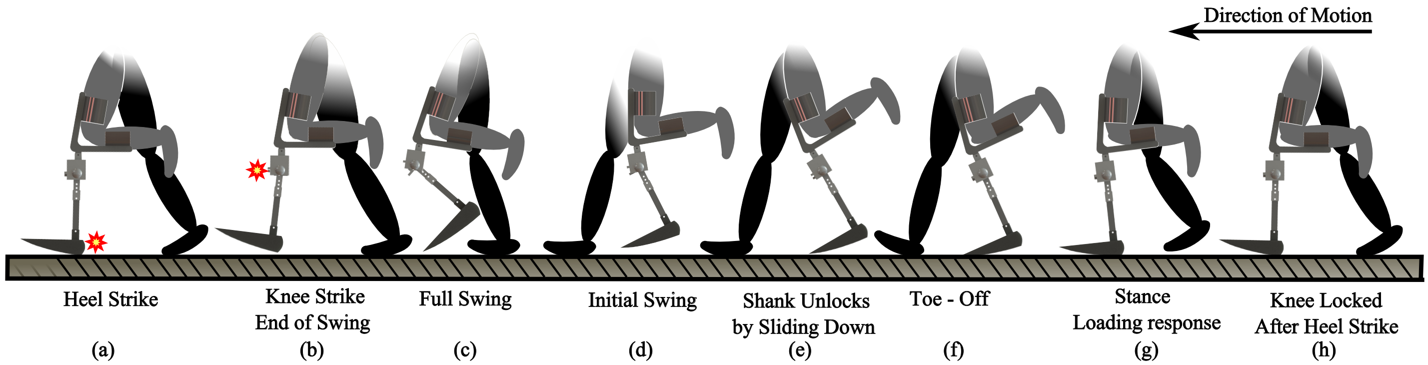

FIGURE 4. Example of walking with the prosthetic simulator.

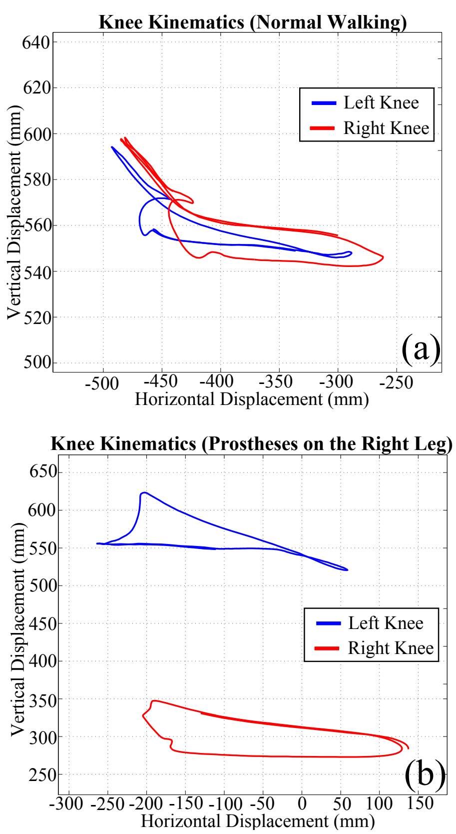

FIGURE 5. Knee Kinematics: (a) The motion of the knees at baseline normal walking is traced. (b) Knee motion is shown with the prosthesis on the right leg. The prosthetic knee was slightly lower than the normal knee to accommodate the prosthetic simulator.

The data obtained from the motion capture was analyzed using a custom Matlab script. The kinematics, absolute angles during gait cycle, and angular velocity during the gait cycle were analyzed. Figures 5(a), 6(a), and 7(a) depict normal walking for the three cases while Figures 5(b), 6(b), and 7(b) depict the cases with the prosthesis. The prosthesis was worn on the subject's right leg for this study.

The plots for the knee kinematics showed an interesting trend. In the case of normal walking, seen in Figure 5(a), the motion of both knees are symmetric and the pattern of the motion is identical, which is to be expected. An expected difference in the pattern of knee kinematics was observed when the prosthesis was worn. In the plot it is seen that the prosthetic knee is lower than the intact knee, which was done to accommodate the knee brace that is worn by the user to lock their knee, depicted in Figure 2(i). The motion of the prosthetic knee is seen to follow the trend that was described in the design section. The shank translates vertically in the femoral housing which is seen as the space between the stance (bottom half of the curve) and swing (top half of the curve) phases. This regular cyclic pattern is not observed in the intact knee curve since the design is not designed to mimic the human knee exactly.

The differences are also seen in the absolute knee angles, measured with respect to the ground. The shank of the knee mechanism is designed to operate with the least amount of friction as possible. Hence, the shank is allowed to swing freely like a pendulum during swing phase. Since this is a passive system, the user has to rely on their dynamics and timing their steps correctly in order to perform a stable gait. This freedom to freely rotate has produced some interesting results as seen in Figure 6(b) where we can see that the prosthesis generates a greater angle during flexion, at about 20% of the gait cycle, than the baseline right leg seen in Figure 6(a). The knee angles are fairly consistent after heel strike, at about 50%, in both cases which means the prosthetic knee locks successfully. We can also see that the intact (left) knee is compensating by gradually flexing with a quick extension.

Similar to the knee angles, the angular velocity profiles for the prosthesis show a larger magnitude of angular velocity during flexion, at about 20% of the gait cycle, and during extension, at

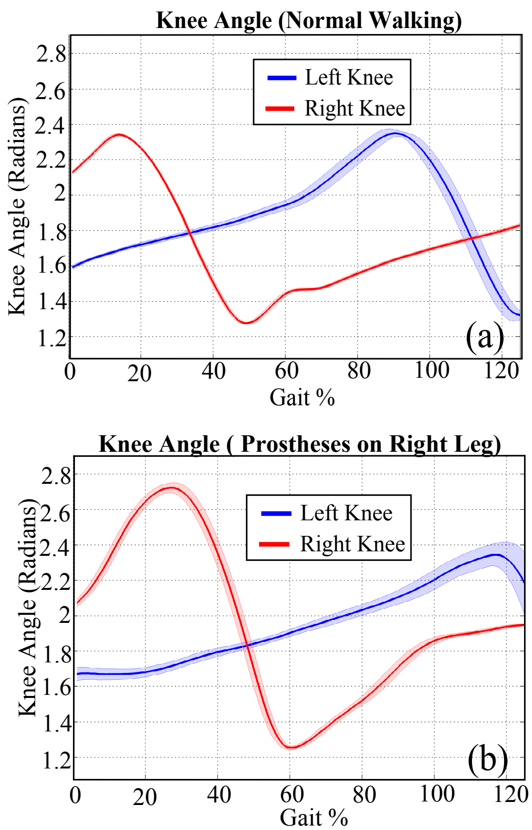

FIGURE 6. Knee angle in radians: (a) Knee angles for normal walking recorded as "baseline." (b) Knee angles with the prosthesis on the right leg shows the shank of the prosthesis has a larger knee flexion angle, and the intact knee compensates by keeping the flexion to a minimum.

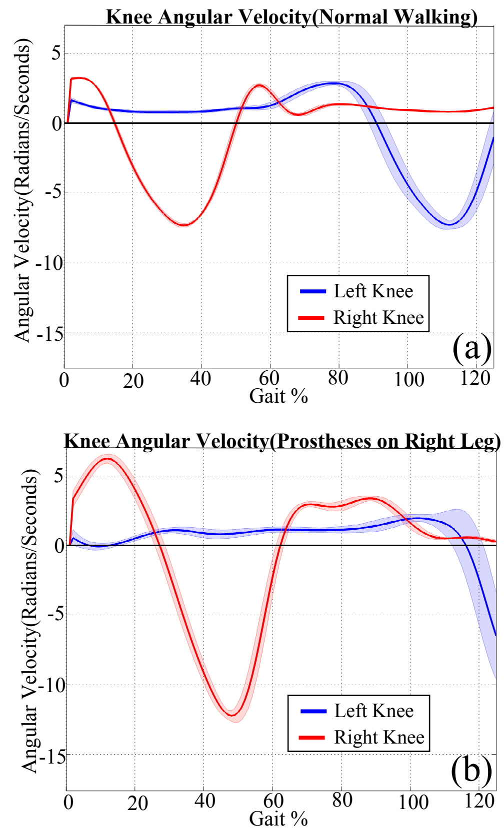

FIGURE 7. Knee angular velocity in radians/seconds: (a) Angular velocity of the knees during normal walking recorded as "baseline." (b) Angular velocity of the knees with the prosthesis on the right leg shows the shank of the prosthesis has a larger magnitude of angular velocity from flexion to extension and vice versa, while the intact knee compensates with low angular velocity and a longer stance phase.

about 50% of the gait cycle, as seen in Figure 7. The higher magnitude of angular velocity of the prosthesis can be attributed to low resistance to rotation of the shank. The intact knee in the prosthetic trial showcases a more prolonged stance phase as opposed to the profile generated by normal walking, seen in Figure 7(b).

Conclusions

The passive position and weight activated knee mechanism is a robust mechanism, surviving rough treatment. The results presented in this paper can be seen as a preliminary step in the introduction of an alternative prosthetic knee mechanism. This passive knee mechanism was featured in a preliminary analysis of asymmetric knee location study [19], but no analysis was performed pertaining to the knee mechanism. The knee mechanism will be used in future studies to confirm the effect of knee location for correcting dynamic asymmetries in trans-femoral amputees [20]. A smaller version of the knee mechanism can also be adapted to robots and bipedal walkers such as passive dynamic walkers, that specifically model asymmetric gait [21, 22].

Future version could include springs and dampers in the knee mechanism to improve the dynamics and reduce forces acting on the residual limb of the amputee. Further, including traditional active systems such as motors and pistons in conjunction with springs and dampers can generate joint forces at only specific stages of the gait cycle. This embodiment of semi-active systems can retain the knee and ankle mechanisms as a low cost option and improve battery life of the prosthesis. Adding semi-active mechanisms at the ankle and foot can aid the performance of the knee during push off and help in ground clearance.

The passive prosthetic knee mechanism showcased in this paper is designed to serve as an alternative to current knee systems to K3 and K4 amputees. The simple passive design can be made better simply by using low cost state of the art materials and processes. This system is designed to be customized as per the requirements of the user and is flexible when it comes to adding new components to improve the control, weight distribution, and locking. Further testing is required to demonstrate the inherent differences between the proposed prosthetic knee mechanism and current passive knee systems.

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