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Towards a Smart Semi-Active Prosthetic Leg: Preliminary Assessment and Testing

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Abstract: This paper presents a development of a semi-active prosthetic knee, which can work in both active and passive modes based on the energy required during the gait cycle of various activities of daily livings (ADLs). The prosthetic limb is equipped with various sensors to measure the kinematic and kinetic parameters of both prosthetic limbs. This prosthetic knee is designed to be back-drivable in passive mode to provide a potential use in energy regeneration when there negative energy across the knee joint. Preliminary test has been performed on transfemoral amputee in passive mode to provide some insight to the amputee/prosthesis interaction and performance with the designed prosthetic knee.

Keywords: Bionics, Biomedical systems, Transfemoral amputee, Smart Bio-leg, Prosthetic knee, Prosthetic systems integration, Biomechatronics.

1. INTRODUCTION

Thousands of lower limb amputations are carried out around the world every year due to complications of diabetes, circulatory and vascular disease, trauma, or cancer (Cristian, 2005). The amputation process is one of the methods when reconstruction surgery cannot provide an adequate solution or the injury to the limb is too severe to recover. The lower limb amputation results loss in mobility of individuals which degrades the quality of amputees' life and lack in functional performance to maintain their activities of daily livings (ADLs). After the amputation was performed the most important goal of healthcare provider must be focused on amputee's return to his/her routine ADLs in the shortest period of time. After the recovery of a subject, a prosthetic leg is an essential assistive device to recover some of missing locomotion functions. During rehabilitation process, the physician assesses the amputee potential level of functional mobility and ability to use lower limb prosthesis and then classifies the lower limb amputees into a range of K-levels by providing a score (K0, K1, K2, K3, K4) (Hordacre et al., 2014). This classification classifies the transfemoral amputees (TFA) from do not have the ability to ambulate safely without assistance (K0) to who exceeds basic ambulation skills and exhibits performance of high impact stress activities (K4). This type of classification helps prosthetist/physician as a guiding rules for prosthetic prescription.

The lower limb prosthesis is defined as a device that substitutes the function of a missing limb either due to amputation or a congenital defect (Pitkin, 2010). The commercial lower limb prosthesis consists of off-the-shelf components and a custom-made socket that are attached to the user's residual limb (stump) as shown in Figure 1. The main prosthetic components of a TFA are socket, prosthetic knee and prosthetic ankle. The purpose of the prosthetic socket is to establish connect between the amputee's residual limb and the prosthesis in order to transmit forces from/to the stump (residual limb) to/from the prosthesis (amputee/prosthesis interaction) while the lower limb prosthesis is used to transfer the weight of the amputee to the ground and provide the main requirements for mobility (prosthesis/environment interaction).

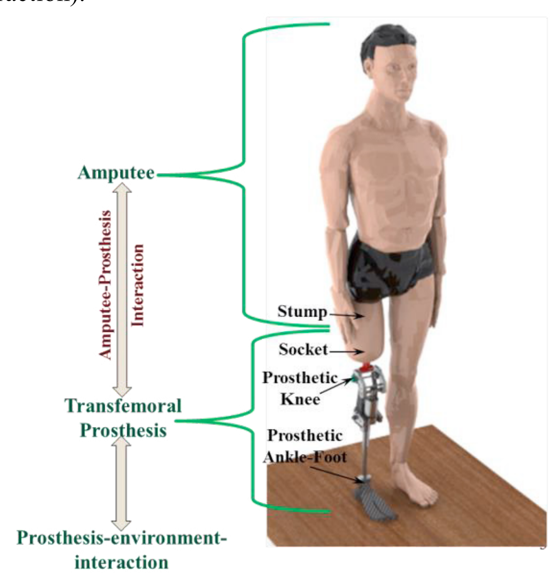

Fig. 1. Prosthetic leg components for transfemoral amputees (TFA).

Prostheses stability and comfort which partially maintained through proper fitting of the socket to the stump, is quite critical for amputees; hence, a well-constructed prosthesis system is important to provide stability and comfort ADLs.

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The TFA gait is associated with compensatory mechanisms that leads to the gait asymmetry due to lack of movements in the knee and the ankle joints to overcome the functional losses. One of the challenges an amputee experiences is to find an efficient prosthetic leg to decrease the gait asymmetry and metabolic energy consumption. Hence, developing a proper prosthetic leg system can help in reducing such compensatory efforts required from amputees to ambulate. In this paper, a development and preliminary testing of a semi-active instrumented prosthetic knee is presented.

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 passive*, *active-damping controlled* and *powered* controlled prostheses.

Purely passive prostheses depend on mechanical systems such as a polycentric knee joint, four bar linkages, locking mechanisms and passive hydraulic/pneumatic cylinders. This type of prostheses requires a significant voluntary control effort from amputees. *Active-damping controlled* prostheses were introduced during the 1990s with the release of the Intelligent Knee (Nabtesco, Japan), the Intelligent Prosthesis (IP) (Zahedi, 1998) (Chas. A. Blatchford & Sons, UK), and the C-Leg (Otto Bock, German). For example, the C-leg (Otto Bock) controls the damping effect using a hydraulic cylinder, and monitors the knee flexion and extension by means of an angle sensor. Other commercial prostheses use either a pneumatic swing control unit; such as the smart IP (Chas. A. Blatchford & Sons, UK), magnetoroheological fluid stance and swing control unit as in the REHO knee (Össur, Iceland) , or combination of hydraulic stance and pneumatic swing control. More advanced intelligent active-damping controlled prostheses were recently presented, which are adjusting the damping torque during the gait using microprocessor, such as Orion microprocessor knee (Zahedi et al., 2005) and the Genium microprocessor knee (OttoBock). The microprocessor prostheses use a wide variety of sensors to measure the load transfer and the knee angle in order to determine when knee flexion and extension is needed and to avoid buckling the knee during stance phase and provide safe progression during walking.

In the case of *active-damping controlled prostheses*, above-knee amputees often compensate for the loss of function in both the knee and the ankle by regulating the transferred energy via the residual limb. This is acceptable during most level ground walking phases and while descending stairs, as the net energy required from the knee is negative and needs to be absorbed. 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. *Powered prostheses*, such as the Victhom knee (Bedard, 2004, Bedard, 2006, Bédard and Roy, 2008), commercially known as the Power Knee and distributed by Ossur are fully actuated. These prostheses are powered using either DC motors (Fite et al., 2007, Sup et al., 2008, Goldfarb, 2013, Goldfarb et al., 2013, Shultz et al., 2014), or pneumatic actuators (Sup and Goldfarb, 2006). Although these prostheses are able to supply positive power, they consume more power than the human joint (Unal et al., 2014). The reason for this is that they use an external power source to generate motion which deteriorates the overall dynamic performance of the system (Unal et al., 2014) in addition to the energy conversation efficiency of the actuation system while the walking process of humans is considered to be mechanically energy-efficient cyclic activity as consequences of comfortable dynamics interaction between human segments. Also, the human muscles show high levels of activity during stance phase, and less activity during the swing phase for level ground walking (Collins et al., 2005). With regard to robotic systems, for example, Honda's ASIMO, which is a completely powered robot, represents 'specific cost of transport' of 3.23 (Hobbelen and Wisse, 2007, Collins and Ruina, 2005) to travel unit distance while the Cornell efficient semi-powered biped expends 0.20 (Collins and Ruina, 2005, Collins et al., 2005) the same 'specific cost of transport' as humans. The passive dynamic walking concept was introduced in 1990 by McGeer (McGeer, 1990b, McGeer, 1990a) such 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 cycle based on just the machine dynamics without the need for a complex control or actuation system.

This explains how above-knee amputees with purely passive or actively damping controlled prostheses can walk by controlling the movements of the residual limbs. Energy is transferred from the residual limbs to the prosthetic knee and produces movement of the prosthetic knee due to the dynamic coupling effect. 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. McNealy et al. (McNealy and Gard, 2008) have shown that energy of the hip joint in TFA has been increased compared with able-bodied subjects . Despite the technological advancement in the prostheses sector, the lower extremity prosthetic legs still have long way to fully emulate human biological limb functionality and provide efficient functional artificial limb.

This paper introduces the mechatronics system design and development of an instrumented semi-active prosthetic leg. This prosthetic knee has back-driveable capability to operate passively in unactuated phase depending on the amputee-prosthesis-environment system dynamics in addition to providing assistive power in actuated phase when positive energy is required. Initial testing of the prosthetic leg on transfemoral amputee in unactuated phase was presented in this paper.

2. BIOMECHANICAL CONSIDERATIONS FOR EFFICIENT DESIGN

Bipedal walking is the human body's natural method for moving from one location to another and is usually the most convenient way to travel distances. Bipedal walking uses a repetitious sequence of limb motions to move the body forward while maintaining postural stability. It is hypothesized that energy consumption is minimized during walking at a self-selected speed (Donelan et al., 2002). As walking is frequently intertwined with performing routine ADLs, the loss of a limb is extremely debilitating and will reduce significantly the individual's quality of life. The concept of energy is an important factor in designing, controlling and developing an efficient prosthetic leg for TFA as the inability to deliver the required power at the right instant significantly impairs the lower limb prosthesis' capacity to restore TFA mobility. Some of the daily life tasks require net positive power; such as ascending stairs, while others require net negative power at the knee joint; such as descending stairs.

In this section, the amount of negative and positive energy required to be dissipated or generated from able-bodied knee and ankle joints during ADLs are studied. The authors used normative gait data from three different references (Bovi et al., 2011, Riener et al., 2002, Winter, 1991) to estimate the amount of normalised positive and negative energy required at the knee and ankle joints during level ground walking and stair ascending/descending as shown in Figures 2 and 3. The analysis showed that the amount of normalised negative energy per person weight dissipated through the knee joint are 0.2576±0.021 J/kg and 1.123±0.22 J/kg during level ground walking respectively while the positive energy required from the knee to ascend stairs leg over leg is about 0.612±0.066 J/kg. This explains why most transfemoral amputees can descend stairs and perform level ground walking activities using passive or adaptive damping controlled prostheses while they cannot ascent stair leg over leg.

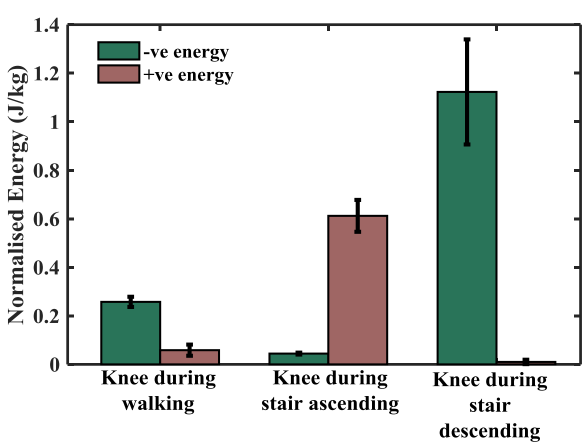

Fig. 2. Amount of positive and negative energy required from knee joint during ADLs.

The ankle is quite important to deliver power during level ground walking and stair ascending as shown in Figure 4. Also, the energy required from the ankle joint to assist able-bodied subjects during level ground walking are more than four times the positive energy required to assist the knee joint as shown in Figure 4. It is clear from these figures that the joints especially knee do not require to be powered continuously and also the knee joint can be used to harvest and recover the negative energy during level ground walking and stair descending and then increase the battery life span in lower limb prosthesis before recharge it.

These points are need for highly versatile and energy efficient lower limb prostheses that can replicate the biological behaviour of the biological human leg. The development of a biomimetic artificial limb that can bring natural comfort to the user entails integration of different scientific approaches, biomechanical, mechanical and electrical, each of which have decisive role in improving the user experience. Therefore, there is a real need for developing a semi-active transfemoral prosthesis that can deliver the required positive energy when it is needed, and regenerate energy during other gait sub-phases.

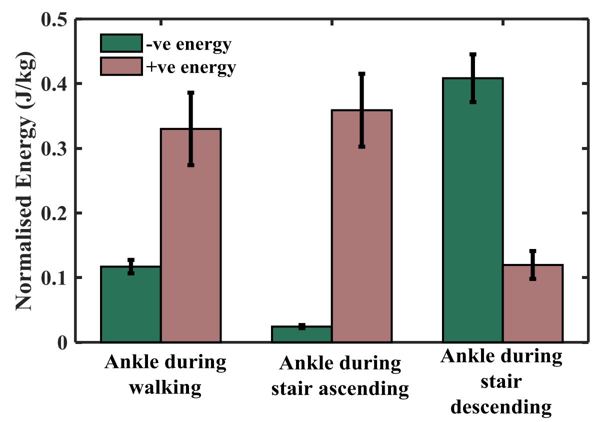

Fig. 3. Amount of positive and negative energy required from ankle joint during ADLs.

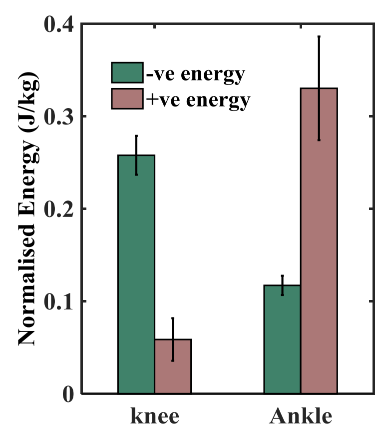

Fig. 4. Amount of positive and negative energy required from knee and ankle joint during level ground walking.

3. PROSTHTIC LEG SYSTEM DESIGN CONFIGURATION

An optimally designed prosthesis and control which can mimic the human biological system. Activities of daily living (ADLs) (e.g. normal walking, stair/ramp ascending and descending, etc.) are the automatic functions that occur every time we take a step and do not require any active concentration on our part. The request for a movement is initiated by the human brain and from there the signals travel through the nervous system to the muscles to actuate the lower limb joints. The human body is equipped with numerous sensors to provide feedback about the progression of the motion and the success of the steps during walking. Muscle and nerve are constantly interacting and update the brain to maintain functional the movement of the lower limb. Also, the human control system, central nerves system (CNS), somatosensory system (includes, proprioceptive and sensory system), biomechanical constrains and movement strategies work together to maintain postural balance in individuals during ADLs (Horak, 2006). Many of these important structures and links in amputated leg are missing and this loss leaves the prosthetic leg the only source to provide a substitutive sensory system through external sensors and haptic feedback. Hence, the efficient prosthetic leg should bring back the experience the user had received before amputation was performed. This can be provided with many sensory feedback embedded into prosthetic leg.

Figure 5 shows the efficient system configuration of TFA prosthetic leg including the interaction with amputee residual limb and the environment. The change in the environment and amputee intent can be either sensed directly or estimated indirectly using variety of biopotential and/or mechanical sensors. The hierarchical control system is recommended for controlling lower limb prostheses and exoskeletons to provide the information required to perform activities of daily living. The intent and activity recognition is associated with estimation and detecting the amputee activity and movement based on the measured information provided from the amputee and environment to the control unit through sensors while the low-level control associated with the torque control of the actuator. The torque control is used to provide safe interaction between amputee, prosthesis and the environment. In order to compensate the missing of proprioceptive and sensory feedback, a haptic feedback can be used to increase the amputee stability and provide information about the prosthetic leg position and whether it touches the ground or not. This layout provides the big picture for development of a smart prosthetic leg.

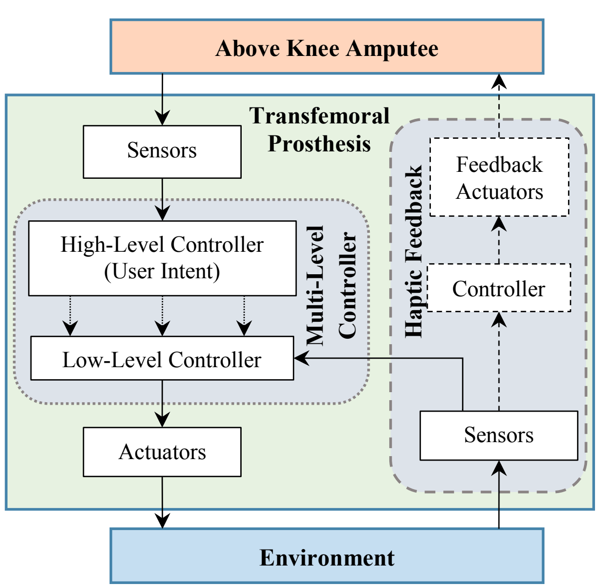

Fig. 5. System configuration and layout of the efficient lower limb prosthesis system.

4. PROSTHETIC KNEE MECHATRONICS SYSTEM DESIGN

Based on the biomechanics considerations for joints energy which were discussed in section 2, the prosthetic knee mechanism should be able to deliver the required positive energy when needed, and recover the negative energy produced during the gait cycle. Hence, a mechanical design and analytical analysis for a semi-active single axis prosthetic knee that can be easily back-driven in the passive mode was presented by the authors based on their previous work (Awad et al., 2012, Awad et al., 2011, Lui et al., 2015).

The Leeds prosthetic knee (LPK) system is equipped with a variety of sensors to monitor the prosthetic status and estimate the user intent and changes in the environment and terrains. Figure 6 shows the mechatronics system design of the developed prosthetic knee by presenting the number of sensors and actuators were used.

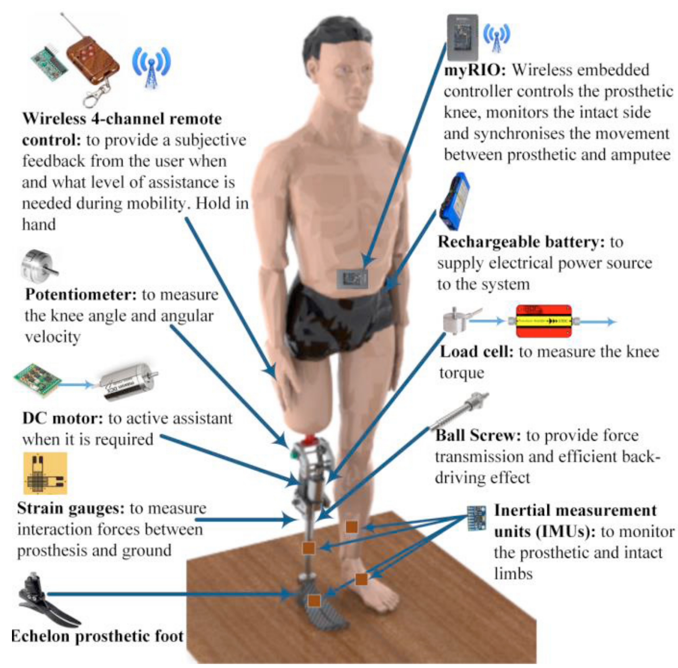

Fig. 6. Leeds prosthetic knee (LPK) mechatronics system design.

In the instrumented prosthetic leg, a single-turn potentiometer was attached to the prosthetic knee centre of rotation to monitor the angle and angular velocity. A miniature single axis load cell was also used to measure the driving knee torque in the active mode and the resistance torque in the passive mode. Two limit switches are connected in series with the motor terminals and were attached to the extreme positions of the prosthetic knee to act as a failsafe. A full bridge strain gauges and force sensors were used to measure anterior and posterior bending moment and axial load in the prosthetic shank. A six degrees of freedom inertial measurement unit (IMU) consisting of three axis accelerometer and three axis gyroscope (MPU 6050, InvenSense Inc) was placed on both shank and foot of prosthetic and intact side. These measurements will be used to detect the gait phases, events and estimate user intent and then switch between the controller states to provide right control assistant during ambulation. The motor supplies assistive power in active mode and harvest energy during regenerative braking mode. A wireless 4-channel remote control is used by the TFA to provide online subjective feedback if the level of the assistant power was not enough or it was not on the right moment. myRIO wireless controller (National Instruments, USA) was used to control and monitor the prosthesis status and transfer the information wireless to the computer.

Figure 7 shows the Leeds prosthetic knee (LPK) system attached to Echelon prosthetic foot (Chas. A. Blatchford & Sons, UK) and a custom-made prosthetic socket. The Echelon foot is a hydraulic dynamic passive prosthetic foot/ankle. At the moment, this passive prosthetic foot was used to evaluate the performance of the prosthetic knee. In the future, it is planned to consider the whole system as shown in Figure 5.

(a) LPK attached to a commercial prosthetic foot and custom-made socket (b) LPK system attached to TFA

Fig. 7. LPK connected to a TFA.

5. PRELIMINARY TESTING IN PASSIVE MODE

The semi-active LPK was testing by one male transfemoral amputee (age: 53 years old; height: 166.1 cm; weight: 66.8 kg). He lost his leg in 2009 due to chronic infection on the knee. The transfemoral amputee had no other neurological or orthopedic disorder apart from his amputation and performed all the experiments without the use of an ambulation aid apart from using parallel bars for safety reason. Information sheet and letter of consent were handed over to the participant containing information about the research background, consequences of participating and description of the experimental activities and a consent form was signed by the participant. All experimental procedures carried out in this research were approved by the Leeds Ethical Review Board.

Gait analysis experiments were carried out on able-bodied subjects and the TFA during level ground walking using a real time 3D motion capture system Qualisys ProReflex MCU240, Track Manager (QTM) (Gothenburg, Sweden) and C-Motion Visual3D V4 (Germantown, MD, U.S.A) to obtain knee angle joint data. The amputee was using his own commercial prosthetic leg for daily use which was a passive prosthetic leg, shown in Figure 8. The motion data were used to compare with the performance of LPK at his self-selected normal walking speed along a straight walking path. The mean angles are shown in Figure 9 where the dashed line indicates the knee angle of the able-bodied subjects.

The LPK was fitted to the TFA participant and he was asked to perform level ground walking within parallel bar structure for safety at his normal pace self-selected speed in passive mode as shown in Figure 10. During the experiment, the signals were acquired from the sensors, transferred wireless and then recorded to the computer. The LPK angle mean and maximum variations were calculated for a total of 33 strides as shown in Figure 9. Shaded area in the figure shows the LPK variations during the experiment in the passive mode. This variations in the collected data from amputee gait was highly influenced by the length of the walking path inside the parallel bar frame (about 3.5m) and transitions at the end of walking path.

Fig. 8. TFA subject with his daily use prosthetic Leg in gait analysis lab.

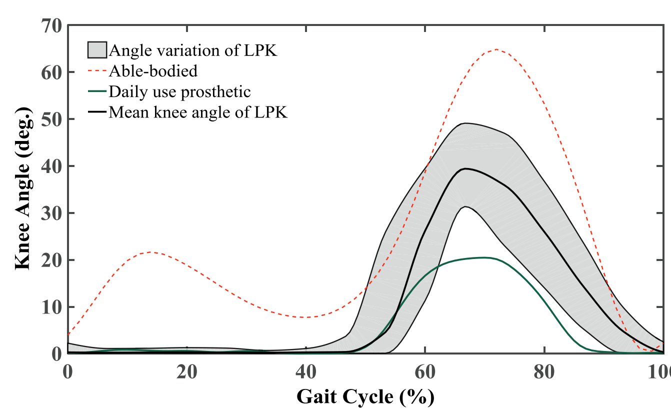

Fig. 9. LPK angle performance compare to able-bodied subject and TFA daily use prosthetic leg.

The results showed that although the amputee can drive the prosthetic leg in passive mode without assistance from the actuator, he cannot produce knee flexion/extension in stance phase (Figure 9). This lack of knee flexion/extension control during stance may be due to the fact that the user was not generating adequate extension torque about the knee joint to prevent buckling during gait cycle. So, he tried to keep his prosthetic leg straight during stance phase. This will affect the gait asymmetry between prosthetic and intact sides and may produce abnormal walking which requires compensatory strategy by the user. The proper control of knee flexion/extension assistance and resistance during stance can be achieved by switching between passive and active mode at the right time. This shall improve the performance in swing phase to mimic the natural human knee angle trajectory.

Fig. 10. TFA subject performs level ground walking with LPK in passive mode.

6. CONCLUSIONS

This paper presented a mechatronic system design of an instrumented lower limb prosthetic leg in addition to preliminary testing in unactuated mode. This semi-active prosthetic knee is a back-driveable mechanism to restore the knee negative energy during ADLs and provide assistance when the positive power is required. The results of the preliminary testing showed promising results of using the developed prosthetic leg in passive mode which required no assistance from external power source in swing phase to produce more natural knee flexion/extension. Future work will focus on further developing of the control algorithm for both active and passive modes for LPK in addition to testing on more amputees. The prosthetic leg system will have the capability of auto-settings and tuning based on the amputees requirements.

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