

A 3DOF Transtibial Robotic Prosthetic Limb

¹D.G.K. Madusanka,²L.N.S. Wijayasingha,³K. Sanjeevan,⁴M.A.R. Ahamed,⁵J.C.W. Edirisooriya,⁶R.A.R.C. Gopura

Dept. of Mechanical Engineering

University of Moratuwa

Katubedda, Sri Lanka 10400

¹kanishka@mech.mrt.ac.lk, ⁶gopura@mech.mrt.ac.lk

Abstract— A robotic prosthesis is a device which is used to replace a missing body part. These devices are intended to return the amputees to their pre-amputation functional status. Below knee (Transtibial) amputation is the most common amputation occurred in the lower limb. That is caused by reasons such as diseases, injury due to explosions of anti-personnel land mines and accidents. The existing prostheses for transtibial amputees have yet to be improved to reinstate the biomechanical functions normally provided by the ankle joint. Therefore, the purpose of this research is to develop a transtibial robotic prosthesis which would provide functions usually provided by the ankle. The proposed design consists of 3 Degrees of Freedom (DOF) to generate similar biomechanical motions at the ankle joint. Further it includes a passive regenerative system to reduce motor power requirement of dorsiflexion/plantarflexion. Experiments are carried out to verify the effectiveness of the proposed prosthetic limb and to verify the possibility of using electromyographic (EMG) signal based control of the prosthesis.

Keywords— *prosthesis; transtibial; biomechanical; ankle joint;*

I. INTRODUCTION

Amputations are caused by diseases, injury due to explosions of anti-personal land mines and accidents which in turn affect negatively on activities of daily living. It can be seen that a significant number of amputees are with lower limb amputations and the majority of them are transtibial [1]. In transtibial amputation the lower limb is amputated between knee and ankle (below knee).

In order to help these amputees, transtibial prostheses have been developed [2]-[5]. These prostheses are supposed to enhance capabilities of amputee making him ready once again to a normal life. A comprehensive prosthesis must be functional, cosmetic and comfortable for the user. It is appealing to have prostheses with movements similar to the lost body part. Furthermore the prosthesis should match the motion ranges of joints, weight, applied forces, torques, velocities and accelerations.

Existing transtibial prostheses can be classified in to three: cosmetic prosthesis, body powered prosthesis and externally powered prosthesis. Furthermore externally powered prosthesis can have passive joints and active joints. Prosthetic device named ‘C-walk’ [2] is containing passive regenerative system with an elastic elements which stores energy during a part of the gait cycle where the foot is in contact with the ground. The energy is released during other parts of the gait cycle. This passive prosthetic device can be used to walk on uneven terrains [3]. Intelligent prosthetic ankle joint with magneto rheological (MR) brake is been actuated passively [4]. In this system active braking enhances the functionality of the device. Although low

weight and reliability are appealing features of the passive devices, muscle fatigue and low torque hinders their usability.

Massachusetts Institute of Technology (MIT) has developed an active transtibial prostheses, ‘I walk’ which is commercially available and contains an active ankle joint [5], [6]. SPARKy [7] and PROPRIO [8] have actuated ankles with single DOF. An active prosthesis with only ankle actuation was developed by the MIT which was designed specifically for sprinting [3]. PANTOE [9] is equipped with both ankle and toe actuation. The actuation methods and Degrees of Freedom (DOF) of available lower limb prostheses are given in Table I.

The lack of DOF in the transtibial prosthesis causes gait asymmetries and lack of human alike motion generation. In order to overcome this issue a novel transtibial prosthesis with 3DOF ankle joint and a passive regenerative system has been proposed in this paper. Biomechanics of the lower limb is described in section II. Section III explains the design of the prosthesis and section IV presents experimental results of verification of motion generation and possibility of applying electromyographic (EMG) signal based control.

II. BIOMECHANICS OF LOWER LIMB

Human lower limb generally consists of 3 major joints: hip, knee and ankle. It can be assumed as a 7 DOF structure, with three DOF at hip, one at knee and three at ankle. Thirty two bones can be found in the human lower limb including hip bone, femur, patella and tibia. Since the ankle joint and the foot are the main focus of this research the biomechanics of the ankle joint and foot is discussed.

The ankle joint can be modelled as a hinge [10]. However it is not a simple hinge joint. The ankle is actually made up of several important structures. The unique design of the ankle makes it a very stable joint. According to [11], ground reaction force on foot when walking had a peak value ranging from 1.0 to 1.5 body weights (BW) and while running, this ranged from 2.0 to 3.5 BW. However, internal structures of the foot encountered much greater forces. Predicted peak talocalcaneal and calcaneocuboid joint loads for walking were 5.4 and 4.2 BW respectively. This is increased to 11.1 and 7.9 BW during running [11]. A transtibial prosthetic device should be able to withstand forces of these magnitudes.

The ankle joint is formed by three major bones called the tibia, fibula and talus. Tibia is the largest bone and it forms the inside part of the ankle. The fibula is the smallest calf bone that forms the outside part of the ankle and the talus is on the upper part of the foot that connects the ankle to the foot. This joint allows the foot to move up and down. The subtalar joint is the primary joint of the foot located just below the ankle joint.

TABLE I. COMPARISON OF AVAILABLE PROSTHESIS

Model	Actuation method	DOF
SPARKy [7]	Ball screw driven by motor	1
PROPRIO [8]	Ball screw driven by motor	1
MIT Prosthesis 1 [3]	Ball screw driven by motor	1
MIT Prosthesis 2 [6]	Ball screw driven by motor	1
PANTOE [9]	Ball screw driven by motor	2

It is formed by two major bones called the talus and calcaneus. The talus is on the upper part of the foot and the calcaneus is the heel bone. This joint allows the foot to move side to side. Articular cartilage at the joint end bones helps to reduce the friction and allows the bones to move smoothly against one another while ambulation and other lower limb activities.

The ankle joint provides 3 DOF. Dorsiflexion/plantarflexion, abduction/adduction and inversion/eversion [5]-[10]. Fig. 1 and Table II illustrate these motions and their movable ranges respectively.

III. DESIGN OF THE PROSTHESIS

This section describes the conceptual modelling and design of the proposed prosthetic limb. The challenges that should overcome when designing a lower limb prosthesis are lightweight of the device and generation of human alike ankle motion. The aim is to find a simple solution with a great autonomy and lower energy consumption while providing the amputee an improved ambulation compared to a passive prosthesis.

Proposed prosthesis is capable of actuating 3 DOF. Out of 3 DOF two are active and the other one is passive. Where dorsiflexion/plantarflexion and abduction/adduction are actively achieved using 2 DC motors and the inversion/eversion is passively achieved using two springs.

The design is carried out in 3 stages while considering the feasibility of design at each stage. In the first stage, the prosthetic limb is designed with two active and one passive DOF. The dorsiflexion/plantarflexion is achieved using a ball screw where a DC motor is coupled to the screw for actuation (see Fig. 2). The next active DOF, abduction/adduction is designed with a worm and wheel drive. The passive DOF, inversion/eversion is designed with two springs. Fig. 2 shows the 3D model of the initial design. Drawbacks of the initial design are less reliability of the design, reduced speed due to higher gear ratio of worm and wheel, and less maintainability. The passive DOF, inversion/eversion is designed with two springs.

In the second design, abduction/adduction mechanism is moved up and spur gear pair is used instead of the worm and wheel drive (see Fig. 3). The ball screw driver of the dorsiflexion/plantarflexion is replaced with a linear actuator. The drawbacks of this design is that the required power of the linear actuator is high and also the weight of the prosthetic limb goes high with the added weights of gear wheels and linear actuator.

The final design is to be addressed the drawbacks of first two designs. In this design, the gear drive of the

TABLE II. MOVEABLE RANGES OF NATURAL LIMB

Motion	Maximum Range (Degrees)
Dorsiflexion/ Plantarflexion	20 / 50
Abduction/ Adduction	20 / 25
Inversion/ Eversion	35 / 25

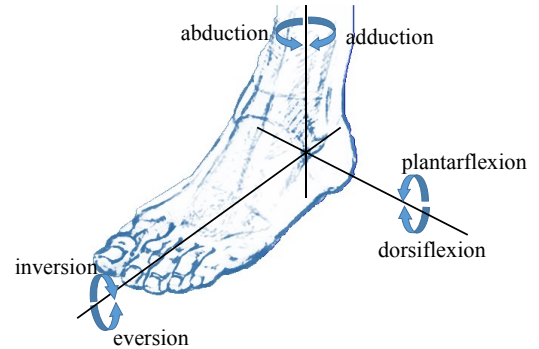


Fig. 1. DOF of lower limb

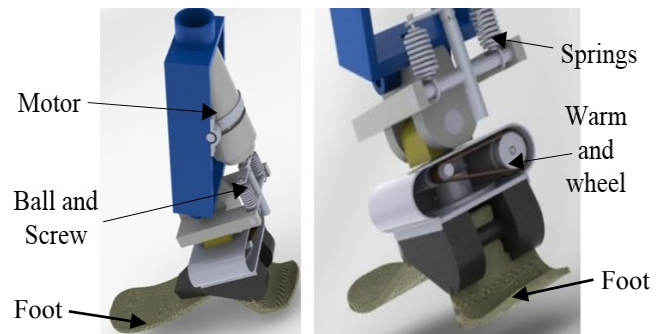


Fig. 2. Initial Design

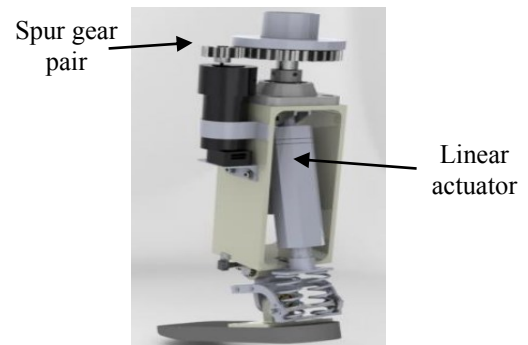


Fig. 3. Second Design

abduction/adduction mechanism is replaced with a belt drive to reduce the weight. The mechanism of dorsiflexion/plantarflexion was kept as the first design with a ball screw drive. The weight of the frames are reduced by reducing materials of the frames based on the finite element analysis (FEA) results. Fig. 4 and Fig. 5 show the 3D model of the final design.

In the final design, inversion/eversion is implemented passively. This DOF is rarely available in most of the state-of-

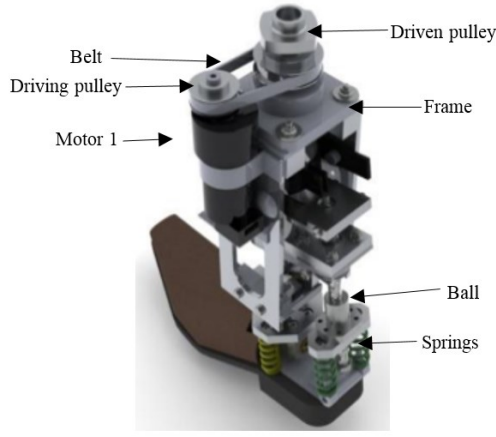


Fig. 4. Solid model of the final design

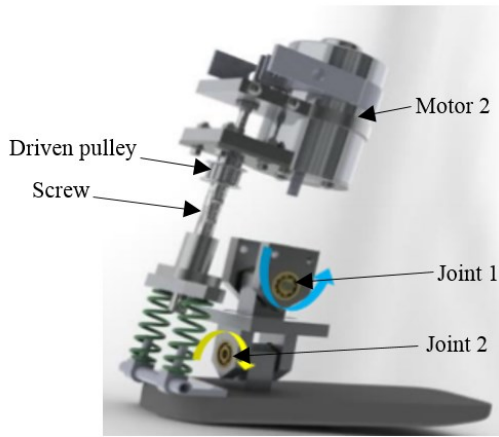


Fig. 5. Solid model of the final design without the frame

the-art prostheses. This passive DOF is being implemented with two springs attached. The importance of this DOF is apparent since inversion/eversion contributes to 10% of the ankle movement during the gait cycle.

As illustrated in Fig. 4 Motor 1 is connected to the frame using Aluminum attachments and the driven pulley is attached to the shaft of Motor 1, which is then connected to the driving pulley mounted on the frame. It can rotate relative to the frame. These two pulleys are connected using a toothed belt drive. The driven pulley is mounted on an attachment which is the connection to the stump limb of the amputee. As Motor 1 rotates, the frame can rotate relative to the stump attachment. This motion generates the abduction/adduction motion of the prosthetic limb. Motor 2 is connected to the frame as shown in Fig. 5. It is connected from inside the frame. The shaft of Motor 2 is connected to a driving pulley and the pulley is connected to a screw through a toothed belt drive and a driven pulley. The screw slides through the ball (nut) as it rotates moving the foot relative to the frame. This motion generates the dorsiflexion/plantarflexion of the prosthetic limb. The Motor 2 rotates screw through the belt drive. Two joints which are shown in Fig. 5 are used to connect frame to the foot. Joint 1 is to generate dorsiflexion/plantarflexion as screw slides through the ball. Joint 2 is to accommodate for the passive DOF of

inversion/eversion. Two springs which are connected to the upper surface of the heel of the foot and to the ball (see Fig. 4 and Fig. 5) bring the limb back to its original position after inversion or eversion. Also these two springs can compress when the foot performs plantarflexion allowing them to store energy. The stored energy is released when the foot performs dorsiflexion. Therefore stored energy in the plantarflexion can be used for the dorsiflexion. This process reduces the power requirement for the dorsiflexion and which can be identified as the passive regeneration.

A. Finite Element Analysis (FEA)

FEA is performed for the critical components of the prosthesis to verify whether they can withstand for the applied loads and to optimize the dimensions of the frames to reduce the weight. The analysis has proven that the design can bear the applied loads. Fig. 6 shows stress distribution of joint 2. It can be seen that maximum stress is $7.5e^{+007} \text{ Nm}^{-2}$. Since Al used for the fabrication has a yield stress of $2.75e^{+008} \text{ Nm}^{-2}$, the minimum factor of safety is 3.65.

B. Kinematic analysis

In order to determine the characteristics of the final mechanism, a kinematic analysis is carried out (see Fig. 7).

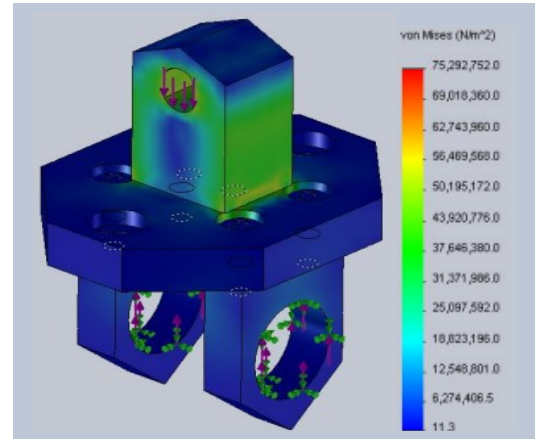


Fig. 6. Results of FEA of joint 2 – load distribution

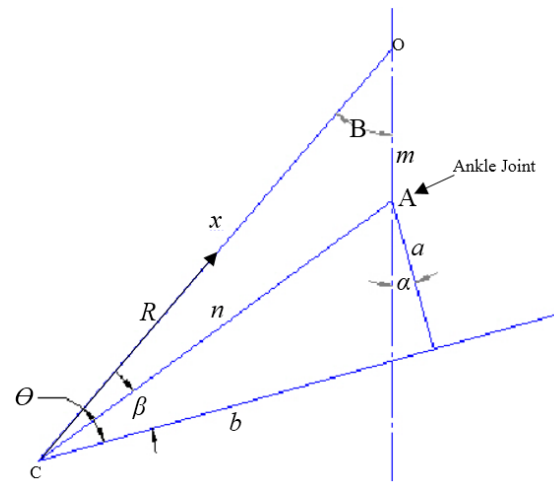


Fig. 7. Schematic of ankle joint

In the Fig. 7, A, C, OC, R, θ and α are ankle joint, heel, series linear elastic actuator (SEA), resultant force on joint C, instantaneous angle between the foot and the SEA and ankle angle respectively. Note that SEA is the combination of linear actuator and the springs as shown in Fig. 5.

From Fig. 7, it can be shown that

$$T = R \cos \theta . a - R \sin \theta . b \quad (1)$$

where T is the ankle torque given by actuator.

Using (1), R can be calculated for a required T and θ .

$$\theta = 90 - B - \alpha \quad (2)$$

By applying sine rule to triangle CAO,

$$\frac{\sin(\theta-\beta)}{m} = \frac{\sin B}{n} \quad (3)$$

$$\tan \beta = \frac{a}{b} \quad (4)$$

By applying cosine law for triangle CAO,

$$x^2 = m^2 + n^2 + 2mn . \sin(\alpha + \beta) \quad (5)$$

α and $\dot{\alpha}$ can be found by gait analysis data. m, n, a and b are known. Therefore, x and $\dot{x}$ can be obtained. These values are used for analysis of power requirements.

C. Analysis of Passive Regenerative System

Robotic tendon approach is applied for the systems to calculate the power requirements of the motor when the system has regenerative passive elements. Here, it is assumed that the ball and screw act as a linear actuator. Furthermore, both springs are being modelled as a single spring (refer Fig. 8).

$$X_g = X_m + X_s \quad (6)$$

In Fig. 8, X_m, X_g, X_s , and F are distance from motor mounting to the point where spring attaches to the screw, distance from motor mounting to the end of spring, length of spring and resultant force acting on spring respectively.

F is calculated by converting the moment needed in gait to equivalent force.

Let natural length of spring be d_0 ,

$$\Delta X_t = X_s - d_0 = \frac{F}{K} \quad (7)$$

$$F = K . \Delta X_t \quad (8)$$

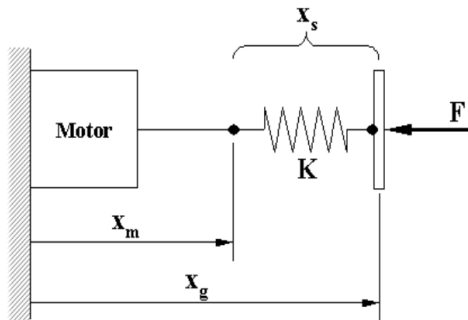


Fig. 8. Robotic tendon approach

where ΔX_t is the deformed length of the spring. It can be shown that,

$$\dot{X}_m = \dot{X}_g - \frac{\dot{F}}{K} \quad (9)$$

Required motor power can be taken by multiplying $\dot{X}_m$ by F . Therefore,

$$P_m = F . \dot{X}_g - \frac{F . \dot{F}}{K} \quad (10)$$

where, P_m and K are required motor power and stiffness of the spring respectively.

$F, \dot{F}$ and $\dot{X}_g$ can be found by analyzing human gait data. Therefore, only design parameter which has to be dealt with is K . The stiffness value which minimizes both energy consumption and maximum power is to be found.

From the analysis (refer Fig. 9), it can be seen that the power requirement of motor is less than that of human ankle joint power requirement. Fig. 9 shows positive values for the power required to carry out the dorsiflexion and negative values to carry out the plantarflexion of the prosthetic limb. The analysis is performed for various values of K ranging from 1kN/m to 50 kN/m with a step of 2kN/m. It was found that minimum power level could be obtained when $K = 30$ kN/m for a 70kg human. Maximum power at this K was found to be 83 W.

Shafts are manufactured using stainless steel and other metallic parts are from Aluminum. Rubber and Nylon are also used to manufacture some parts. It is desirable for the prosthesis to have a weight less than that of an actual human limb. The prototype weighed slightly greater than a typical human limb, 3.8 Kg. Fig. 10 shows the developed prosthetic limb.

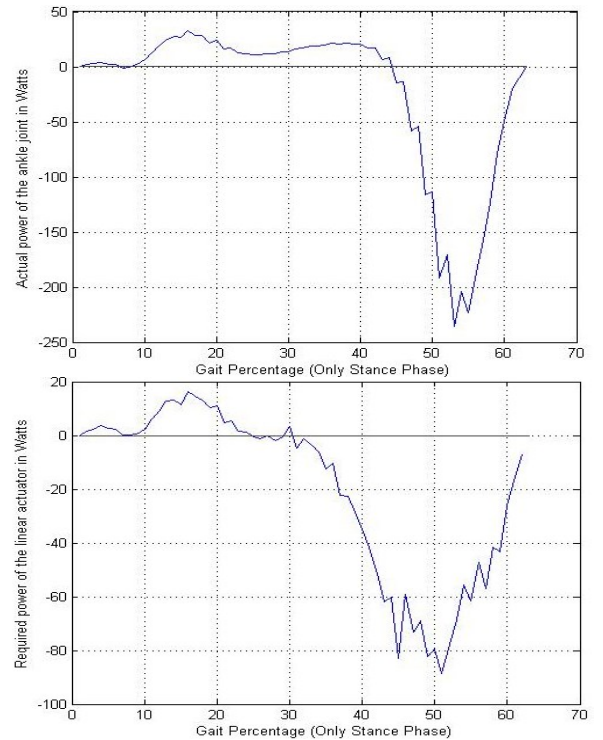


Fig. 9. Results of passive regenerative system

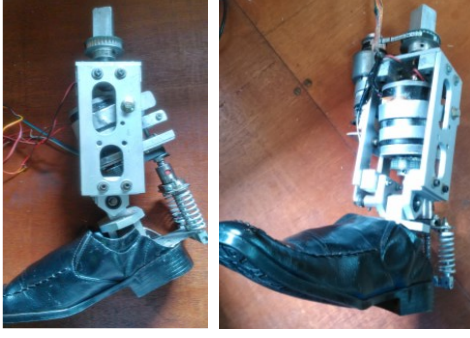


Fig. 10. 3DOF prosthetic limb

IV. EXPERIMENTS AND RESULTS

A. Experimental Setup

Prototype of the prosthetic limb is manufactured (see Fig. 10) and evaluated its effectiveness. For the evaluation purpose an attachment (see Fig. 11) which can be worn onto a healthy limb is fabricated. The attachment consist of a 50k Ω potentiometer. It is used to extract the natural motions of the human lower limb. These motions are regenerated by the prosthetic limb. Furthermore, EMG signals extracted from the remaining muscles can be used to control prosthesis [12].

The experimental setup is shown in Fig. 11. It consists of 3DOF prosthetic limb, attachment for the healthy limb, microcontroller (Arduino Uno), H-bridge (SX8847, SONXUN), H-bridge (L298), data acquisition (DAQ) Card (6220, NI), EMG extraction system (16 channel Bagnoli™ desktop EMG system, DELSYS) and a personnel computer with LabVIEW software. L298 is used for Motor 1 and SX8847 for Motor 2. In the Fig. 11 arrows show the direction of signal flow.

B. Experiments

In the first experiment, proportional-derivative (PD) controller is implemented in the microcontroller for both motions to verify the motion ranges of the prosthetic limb. Three human subjects are used to carry out this experiment. The PD controller is implemented as of (11).

$$M_s = K_p \cdot E_p + K_d \cdot E_d \quad (11)$$

where, M_s , K_p , K_d , E_p and E_d are input signal for motor, proportional gain, derivative gain, proportional error and derivative error respectively.

$$E_p = P_d - P_a \quad (12)$$

where, P_a and P_d are actual angular position and desired angular position of a prosthetic joint respectively.

$$E_d = \frac{E_p(t) - E_p(t-1)}{t} \quad (13)$$

where, $E_p(t)$, $E_p(t-1)$ and t are current angular position error, previous angular position error and sample time respectively.

Second experiment is carried out to verify the possibility of controlling the prosthetic limb using EMG signals. In the experiment surface EMG signals are used since they can be easily extracted from the skin surface of muscles. EMG signals

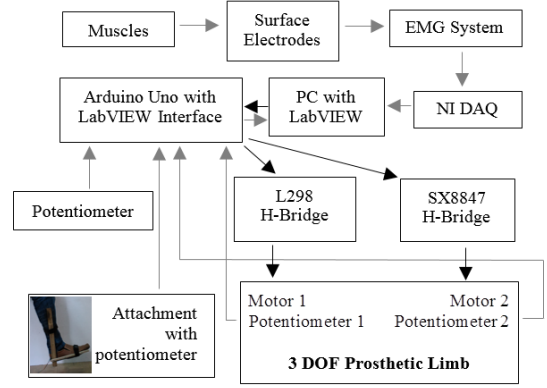


Fig. 11. Experimental Setup

are sampled at 2000Hz and 100 samples are used to obtain a single root mean square (rms) value.

Two EMG signals are extracted from the tibialis anterior (ch:1) and gastrocnemius (ch:2). Amplification gain of 1k and 10k are used for ch:1 and ch:2 respectively. Then two EMG signals were taken into the personal computer through the 'National Instruments' (NI) DAQ card. These two signals are read from LabVIEW software and the same software is used to process data. The rms value of two signals are calculated separately.

Equation (14) is used to control the dorsiflexion/plantarflexion motion of the prosthetic limb.

$$M_E = K_1 \cdot RMS_1 - K_2 \cdot RMS_2 \quad (14)$$

where M_E , RMS_1 , RMS_2 , K_1 and K_2 are input signal for motor, rms of ch:1, rms of ch:2, muscle dependent parameter for RMS_1 and muscle dependent parameter for RMS_2 respectively.

K_1 and K_2 are varying from person to person and can be found experimentally. In this case K_1 and K_2 are set to 1. Sign of M_E gives the motor rotation direction. Then this signal is send to the motor controller through microcontroller using LabVIEW software.

C. Experiment Results

Summary of the first experiment results shown in Table III. It gives experimentally achieved range of angular motion of dorsiflexion/plantarflexion, abduction/adduction and inversion/eversion. From the results, it can be seen that only dorsiflexion/plantarflexion has lesser movable range than that of the natural limb. However, achieved range is sufficient to carry out the activities of daily living.

Results of the PD control for are dorsiflexion/plantarflexion and abduction/adduction are shown in Fig. 12 and Fig. 13 respectively. Movable ranges shown in Table III are extracted from the results shown in Fig. 12 and Fig. 13.

TABLE III. ACHIEVED MOVABLE RANGES OF THE PROSTHETIC LIMB

Motion	Angular range (Degrees)
Dorsiflexion/ Plantarflexion	12 / 32
Abduction/ Adduction	20 / 25
Inversion/ Eversion	30 / 25

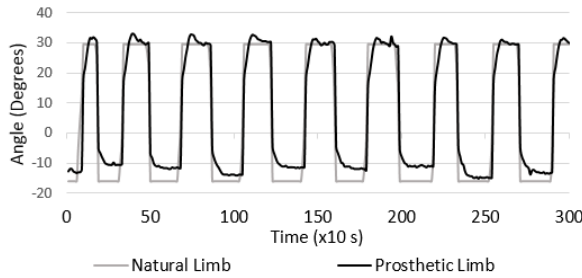


Fig. 12. Results of Dorsiflexion/plantarflexion

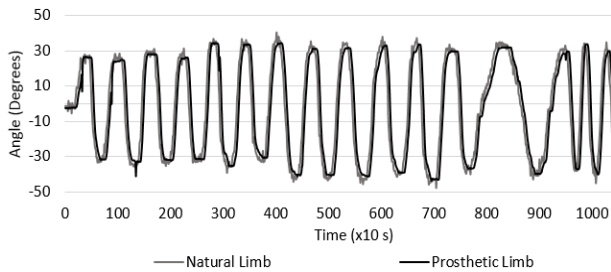


Fig. 13. Results of Abduction/Adduction

As per the results, prosthetic limb can follow the natural limb motion given through the attachment for the healthy limb. However dorsiflexion/plantarflexion have some distortions at lower maximum, when it is compared to the input signal (see Fig. 12). This is because of the minimum value that the prosthetic limb can achieve is little lesser than that of the natural limb angle.

Fig. 14 shows the rms signals of ch:1 and ch:2 of EMG signals. Fig. 15 shows the prosthetic limb angle and the natural limb angle for the EMG based control. Even though the prosthetic limb has a little lag in motion generation due to communication delays, it follows the natural limb motion.

V. CONCLUSION

This paper proposed a robotic prosthetic limb for transtibial amputees. The proposed prosthesis has 3 DOF at the ankle joint with a passive regenerative system. The experimental results verified the effectiveness of motion generation and the possibility of use of EMG based control method for controlling. Simulation results shows the effectiveness of passive regenerative system of the prosthetic device.

ACKNOWLEDGMENT

The authors would like to acknowledge the support given by Senate Research Council (SRC) grant (no. SRC/LT/2013/07) and National Research Council (NRC) grant (no. 11-067).

REFERENCES

[1] N. S. Gunawardena, R. A. Senevirathne and T. Athauda, "Prosthetic outcome of unilateral lower limb amputee soldiers in two districts of Sri Lanka," *J. Prosthetics and Orthotics*, vol. 16, no. 4, pp. 123-129, Oct. 2004.

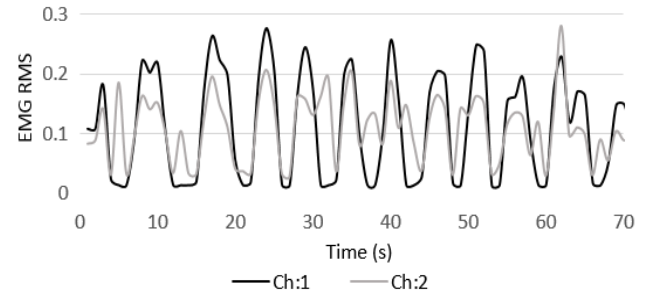


Fig. 14. EMG Signals used for the experiment

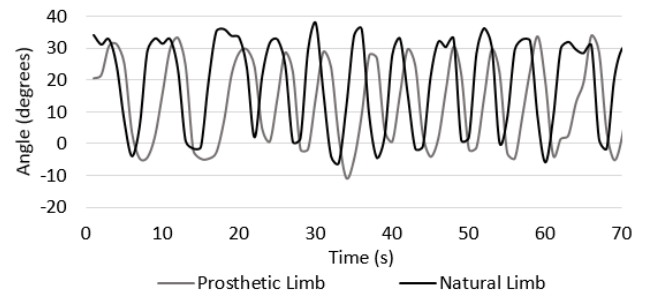


Fig. 15. Results related to EMG experiment

[2] Ottobock, "C-Walk 1C40," ottobock, 2014. [Online]. Available: http://www.ottobock.com/cps/rde/xchg/ob_com_en/hs.xsl/2016.html. [Accessed 21 08 2014].

[3] S. Au and H. Herr, "Powered ankle-foot prosthesis," *IEEE Robot. Automat. Mag.*, vol. 15, no. 3, pp. 52-59, Sep. 2008.

[4] T. Arman and V. Maxime, "Design of a below-knee prosthesis powered by electric drives," Master's thesis, Dept. of Mech. Eng., Vrije Universiteit, Brussel, Belgium, 2008.

[5] S. K. Au, H. Herr, J. Weber and E. C., "Powered Ankle-Foot Prosthesis for the Improvement of Amputee Ambulation," in *Eng. in Medicine and Biology Soc. 29th Annu. Int. Conf. of IEEE*, Lyon, 2007, pp. 3020-3026.

[6] M. F. Eilenberg, H. Geyer and H. Herr, "Control of a Powered Ankle-Foot Prosthesis Based on a Neuromuscular Model," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 18, no. 2, pp. 164-173, April. 2010.

[7] R.D. Bellman, M.A. Holgate and T.G. Sugar, "SPARKy 3: Design of an active robotic ankle prosthesis with two actuated degrees of freedom using regenerative kinetics," in *Biomedical Robotics and Biomechanics 2nd IEEE RAS & EMBS Int. Conf.*, Scottsdale, 2008, pp.511-516.

[8] Ö. Americas, "OSSUR," OSSUR, 2014. [Online]. Available: <http://www.ossur.com/prosthetic-solutions/products/feet/bionic-feet/proprio-foot>. [Accessed 20 08 2014].

[9] J. Zhu, Q. Wang and L. Wang, "PANTOE 1: Biomechanical design of powered ankle-foot prosthesis with compliant joints and segmented foot," in *Advanced Intelligent Mechatronics. IEEE/ASME Int. Conf.*, Montreal, 2010, pp.31-36.

[10] N. Wrabaski and J. J. Dowling, "An investigation into the deformable characteristics of the human foot using fluoroscopic imaging," *Clinical Biomechanics*, vol. 22, no. 2, pp. 230-238, Feb. 2007.

[11] V. L. Giddings, G. S. Beaupre, R. T. Whalen and D. R. Cater, "Calcaneal loading during walking and running," *Medicine & Sci. In Sports & Exercise*, vol. 32, no. 3, pp. 627-634, Mar. 2000.

[12] T.Lenzi, S. D. Rossi, N. Vitiello and M.C. Carrozza, "Proportional EMG control for upper-limb powered exoskeletons," *Eng. in Medicine and Biology Soc. Annu. Int. Conf. of IEEE*, Boston, 2007, pp. 628-631.