

Original Research Article

Modeling and simulation of lower limb dynamics using simmechanics for potential applications in bilateral prosthesis control

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Abstract

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Bilateral transfemoral amputees face many vital problems in normal walking that prevent them from being full-time prosthesis users such as unbalance, instability and increased energy expenditure. They use two prosthetic legs where each leg has a separate control system. In human body, both legs are related to each other and are related to human body while in motion. Consequently, the two prosthetic legs should be controlled using one system. In this work, a SimMechanics model that relates human legs dynamics to each other and to the body in normal walking was developed for further use in bilateral prosthesis control and development. Model inputs are motion data of ankle, knee, and hip joints and the outputs are joints forces and torques. The model showed a good performance when comparing its outputs to the literature bio-mechanics data.

Keywords: Human Body Motion, Modeling and Simulation, Simulink, SimMechanics, Human Biomechanics Data, Human Body Metrics, Reverse Dynamics, Lower Limb Dynamics, Bilateral Prosthesis Control

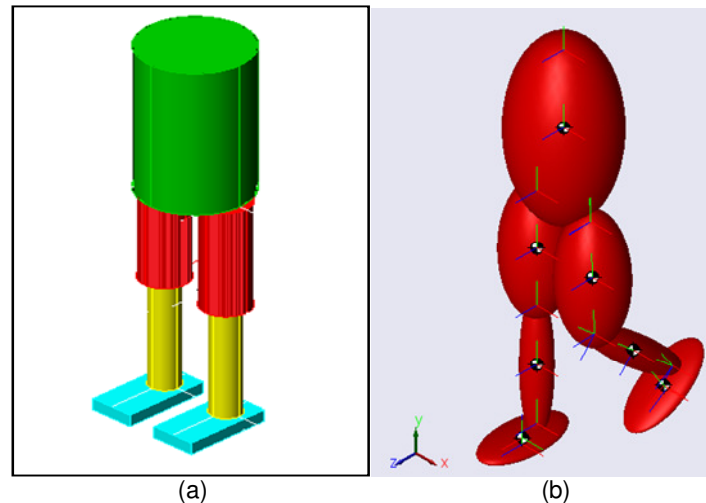
INTRODUCTION

Bilateral transfemoral amputees present a real challenge to scientists because amputees at the beginning of using prosthesis fail to use full length prosthesis as they experienced a severe unbalance and instability (Carroll and Richardson, 2009). In order to have a positive prosthesis consequences there are important factors that can result in favorable outcomes for patients (Carroll and Richardson, 2009), these include; getting a peer-to-peer care, having a good physical shape using contented sockets, well residual limbs, ongoing physical therapy, positive family support/involvement, and unwavering commitment from the prosthetic and rehabilitation team. In addition, to have a full motion using two transfemoral prosthesis they are advised to follow a stepped methodology to prosthesis fitting where patients start with short legs and increase the length of the prosthesis gradually to reach the full-length legs. A microprocessor-controlled swing-and-stance hydraulic knees and energy-storing prosthetic feet are used. These knees automatically adapt to the speed, length, and frequency of each step, ensuring the best natural gait pattern probable. This helps amputees in walking on slopes, uneven surfaces, curbs, or stairs (Hafner et al., 2007),

but these knees are controlled separately as each prosthetic leg has its own control system and patients can't use them directly at the beginning as explained earlier. Hence, a model that can relate the motion dynamics of both legs is needed for robust control of bilateral amputee prosthesis to synchronize their motion and supply patients with better balance, stability and decreasing the rehabilitation period while using different prosthesis lengths to train and balance themselves on full length prosthetic legs. Using one model to present the relation between both legs and their relation to the human body will simplify the control process that targets the amputee safety. Further, in human gait all of hip joint, knee joint and ankle joint are related together. Hence, each prosthetic leg of them shouldn't be controlled separately from the other. Moreover, energy requirements are increased for bilateral transfemoral amputee as the body expends 280 percent more energy than that required from a person who is not an amputee (Huang et al., 1979). Hence, this should be taken into consideration while designing the prosthesis in order to provide the required energy through the amputee gait and make the patients have a natural gait without that

Table 1. Body segments' data

Body segment	Shape	Mass [Kg]	Inertia [Kgm ²]
Upper body	Cylinder	37.968	[1.3856 0 0; 0 0.6358 0; 0 0 1.3858]
Upper leg	Cylinder	5.768	[0.1072 0 0; 0 0.0397 0; 0 0 0.1072]
Lower leg	Cylinder	2.4	[0.0466 0 0; 0 0.0033 0; 0 0 0.0466]
Foot	Rectangular	0.84	[0.0018 0 0; 0 0.0065 0; 0 0 0.0079]

**Figure 1.** (a) Body segments drawn by AUTOCAD, (b) Body segments in SimMechanics while simulation.

increase in the energy required.

Human body modelling can be used in the prosthesis and orthotics design and development (Oçgüder et al., 2012; Jamshidi et al., 2009), humanoid robots design and control (Daumas et al., 2005; Winder and Esposito, 2008), analysis of sportive motion in different sports for sake of optimization (Hubbard et al., 1989; Cavanagh and Lafortune, 1980). A few researchers used external devices to measure body segments' forces and torque by fixing these devices on the segments, this limits the motion and affects the accuracy of the data (Serbest et al., 2015), so more accurate modeling methods should be found and applied. In (Kutilek and Hajny, 2010) the human body motion was measured as angles between body segments and a SimMechanics model is created such that it handles these angles as inputs and outputs the required forces and torques to create these movements by body segments to evaluate the quality of human walking. Segments' weight and moment of inertia can be adjusted according to human body measurements. In (Serbest et al., 2015) a SimMechanics model of human body is developed also, the aims of the model were to calculate the joint torques on the lower limb while daily body movements of sit-to-stand, bend down-stand up, and stair ascending deprived of external help. A link segment model was prepared with respect to

human anthropometric measurements.

The aim of this study is to create a SimMechanics model that can simulate human body motion in normal walking for bilateral prosthesis control by synchronizing the motion of both legs and help in delivering the required energy for amputee gait. Then, model validation using human biomechanical data in normal walking will be done with literature data. SimMechanics represents a cost saving modeling method that suits the researchers in the developing country where they can't afford buying the open source software, e.g., OpenSim, Anybody, BoB, etc. that permit realistic geometric representation of the body segments and allow for the insertion of prosthetic components and custom control systems.

Modeling

Human body segments and joints (Mechanical model).

In the model developed in this work the human body was segmented into seven segments including; the upper body segment, two upper leg segments, two lower leg segments, and two foot segments. Segments' masses and inertia are shown in Table 1. The segments are

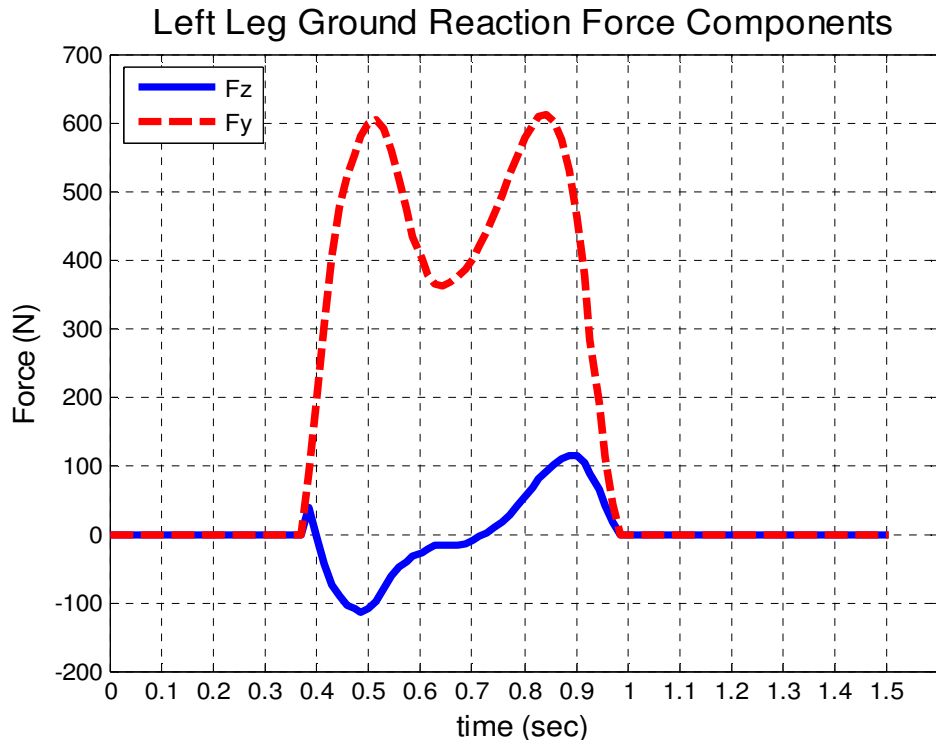


Figure 3. Ground reaction force components of left leg.

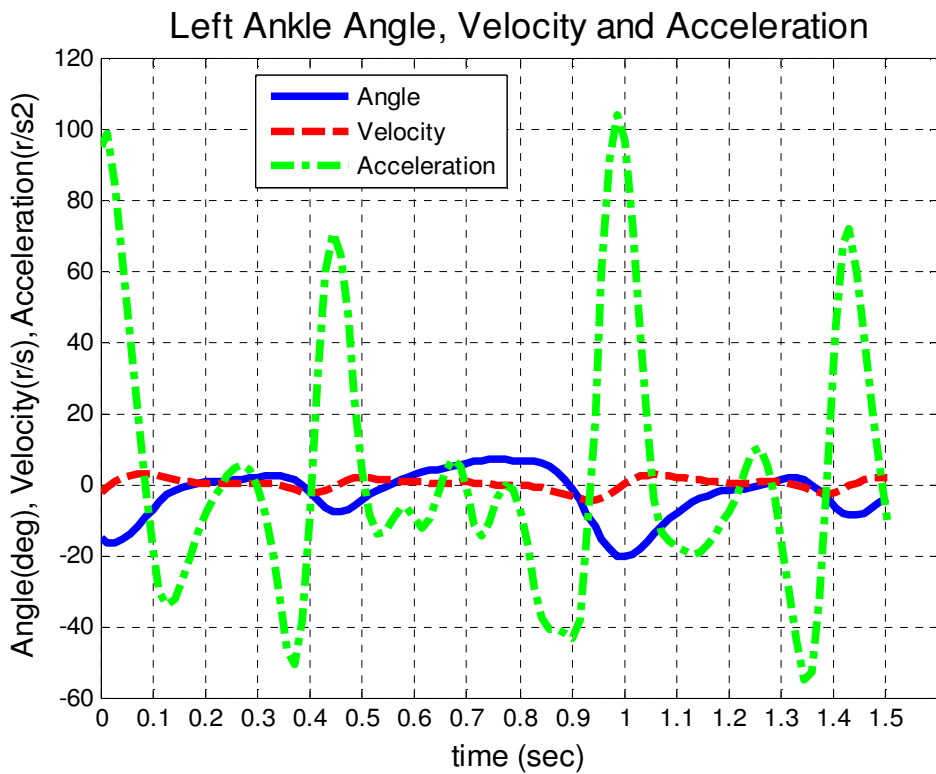


Figure 4. Position data of left ankle joint.

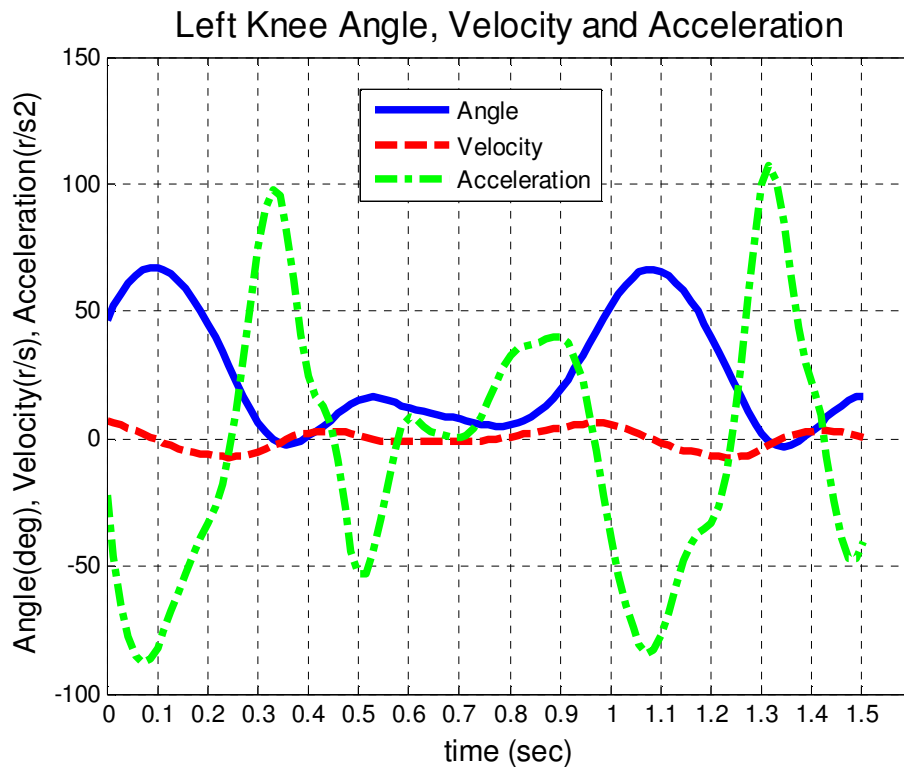


Figure 5. Position data of left knee joint.

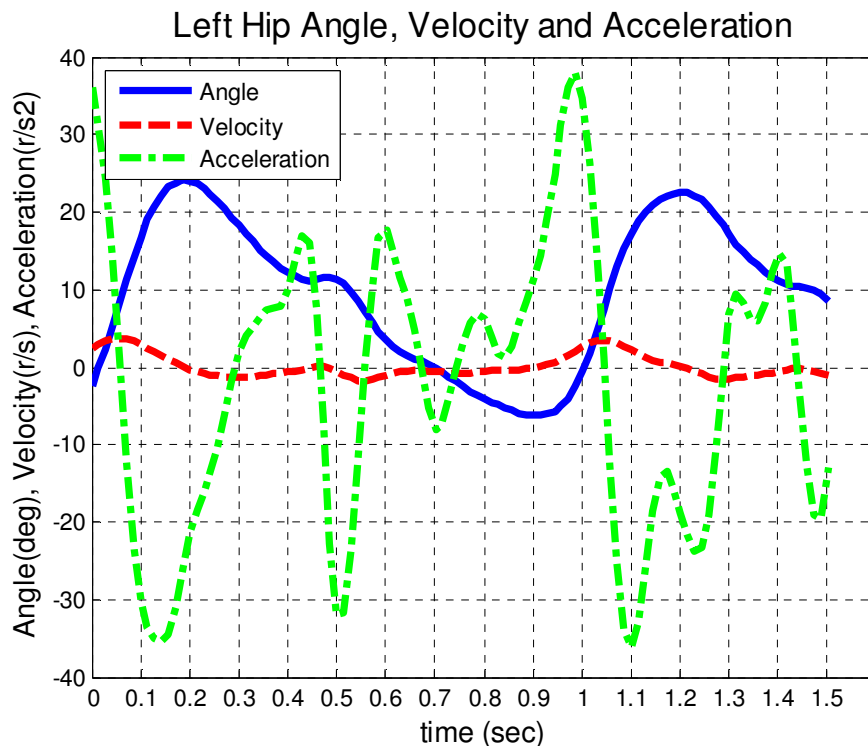


Figure 6. Position data of left hip joint.

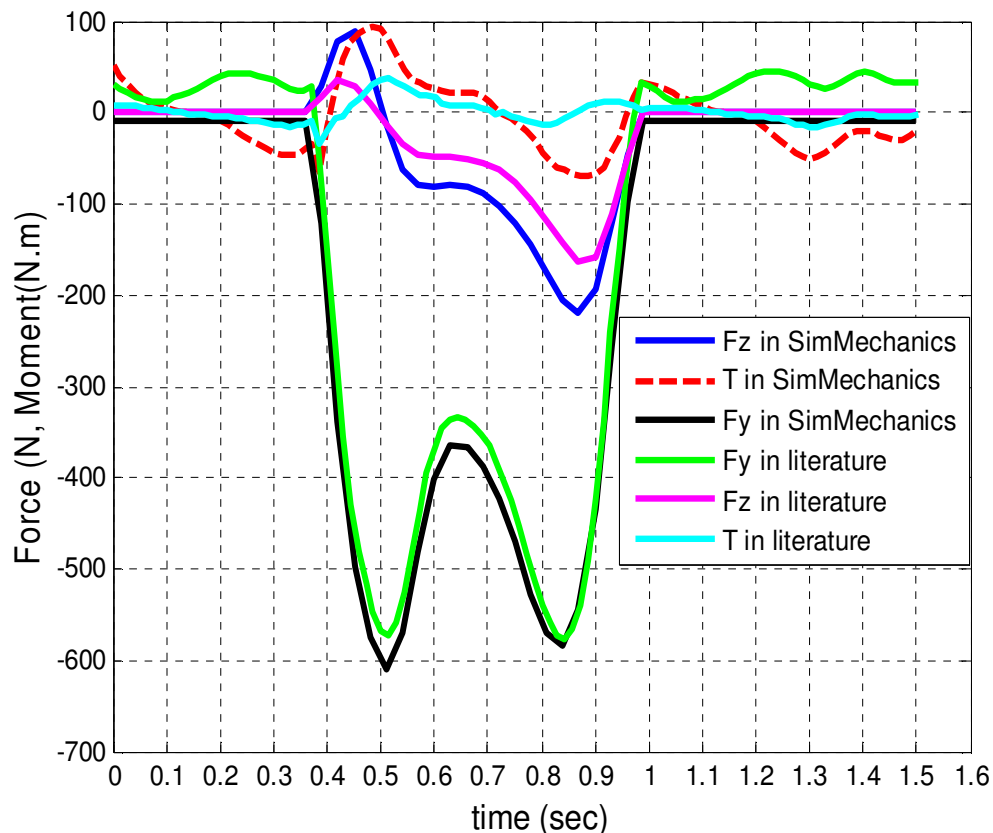


Figure 7. Left ankle reaction forces and moment in both literature and simulation.

Modeling technique

SimMechanics software is used in system modeling, SimMechanics is using a block diagram modeling technique in designing and simulating engineering systems composed of rigid multi-body machines and represents their motions with Newton equations for forces and torques.

The developed SimMechanics model

In the SimMechanics model each segment is represented by a rigid body block that is defined by its mass, inertia, and coordinate origins and axes for center of gravity. The body initial position and orientation is set, each body is connected to a joint, each joint is actuated by a joint actuator giving the motion data and the resulted force and torque of the joint are tracked using joint sensors (Kinematic mode in simulation). Figure 1 (b) presents body segments in SimMechanics simulation and Figure 2 illustrates the SimMechanics developed block diagram.

Simulation

Input data preparation

Input data for measurements of normal walking in human used in validating the model through simulation are obtained and prepared for the simulation from (Brand, 1992; Winter, 1990), model inputs are the position data for each joint which are angle, velocity and acceleration in addition to the ground reaction force components for each foot, while the output data are force and torque components at these joints. The ground reaction force components are shown in Figure 3. Further, the position data of the ankle, knee and hip joints are shown in Figure 4, Figure 5 and Figure 6, respectively.

The motion of both left and right legs are simulated. The developed system is simulated and the force and torque data of each joint are recorded and stored in data files. The cycle time is about 1 second for each step. The step is divided into two major areas the *swing* phase where the foot is not touching the ground, it is about 40% of cycle time, and the *stance* phase where the foot is

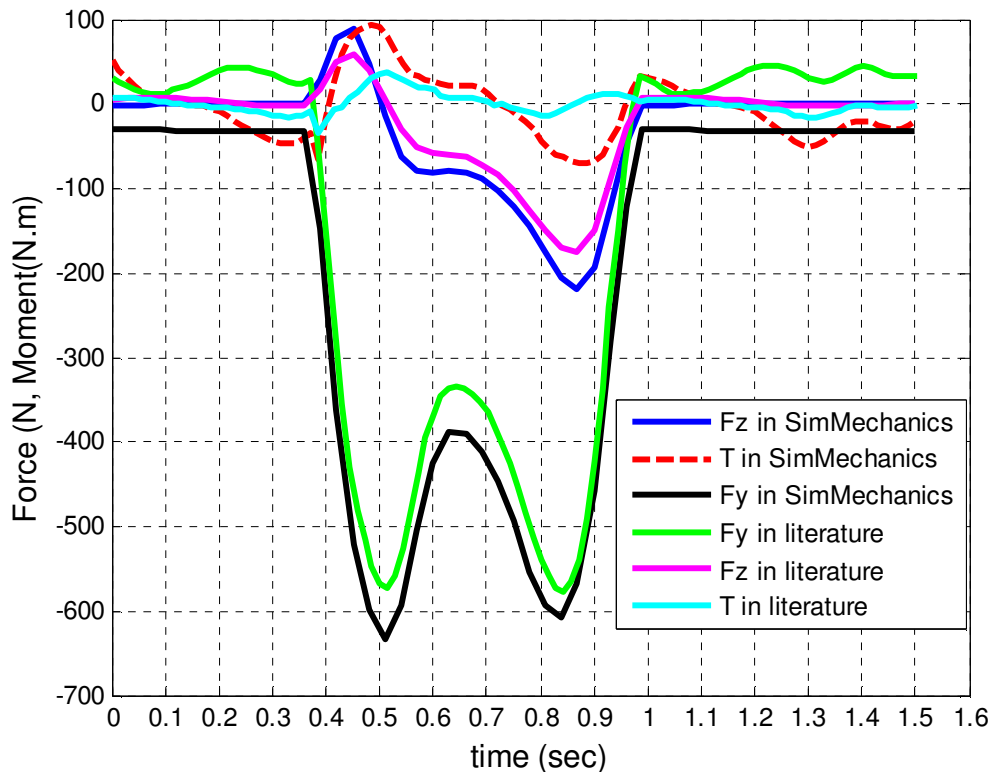


Figure 8. Left Knee reaction forces and moment in both literature and simulation.

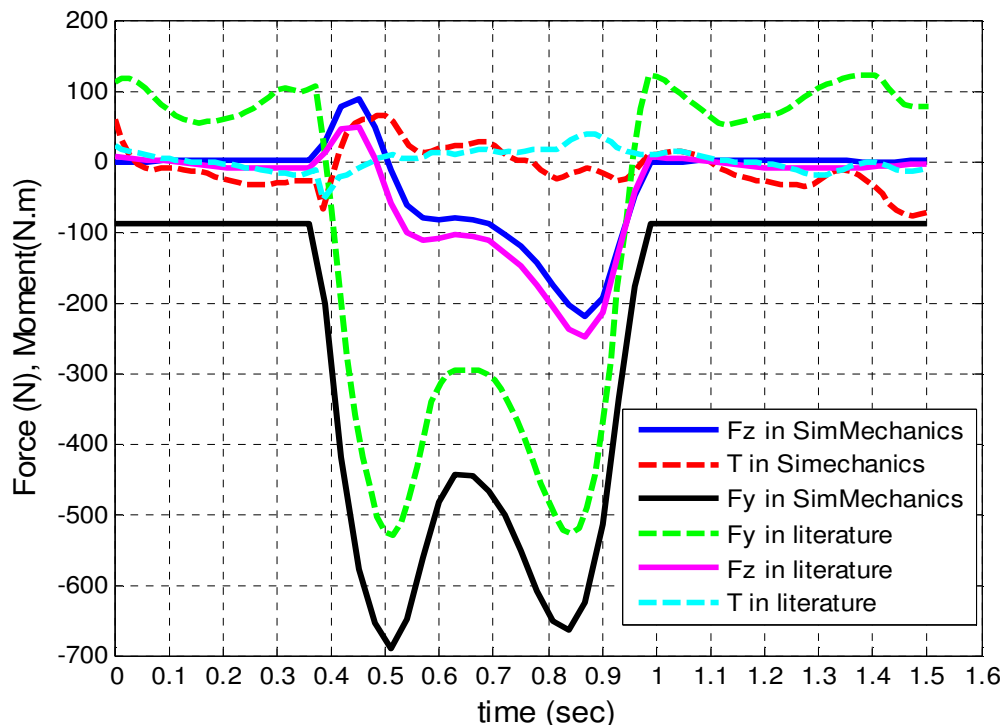


Figure 9. Left hip reaction forces and moment in both literature and simulation.

Table 2. RMSE for force and torque components in left joints.

	Fz (RMSE)	Fy (RMSE)	T (RMSE)
Left ankle joint	7.3127	26.8431	5.2415
Left knee joint	7.9879	50.2108	5.2415
Left hip joint	13.7837	163.6531	10.7717

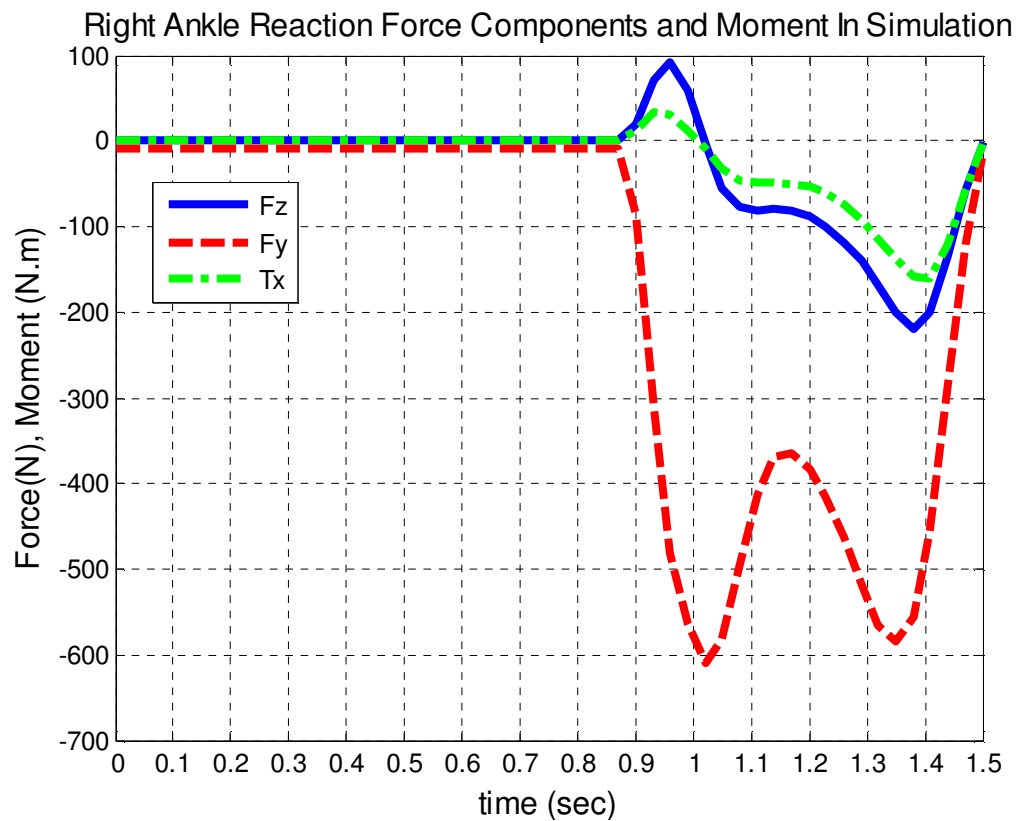


Figure 10. Right ankle Forces and moment in simulation.

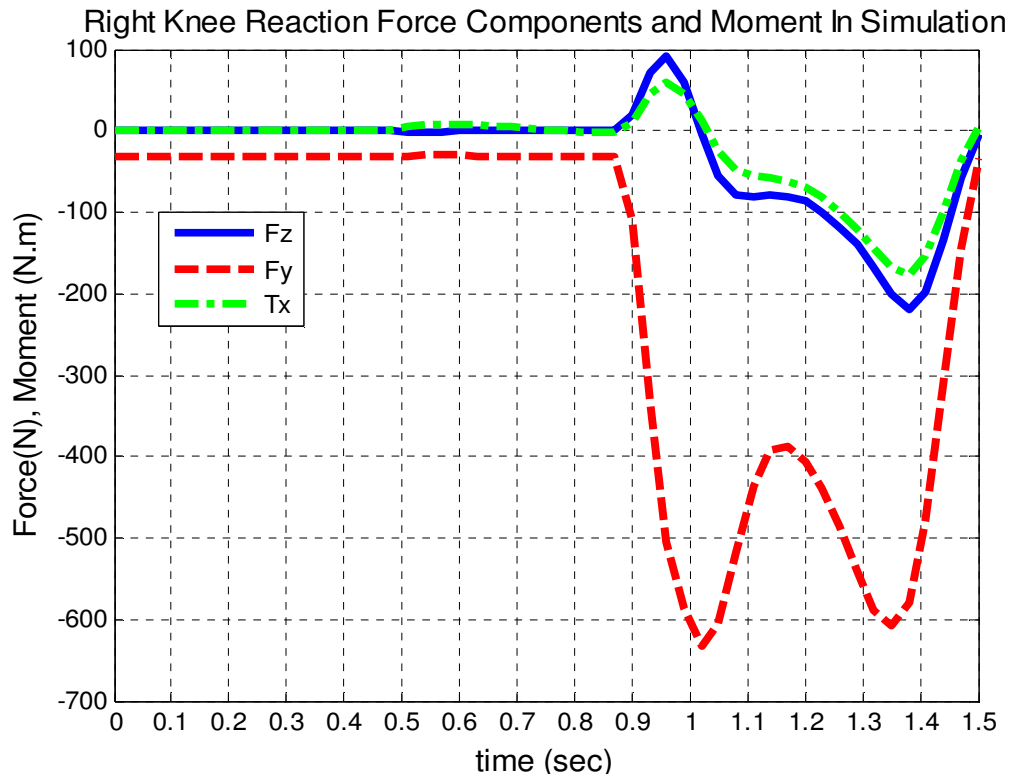


Figure 11. Right knee forces and moment in simulation.

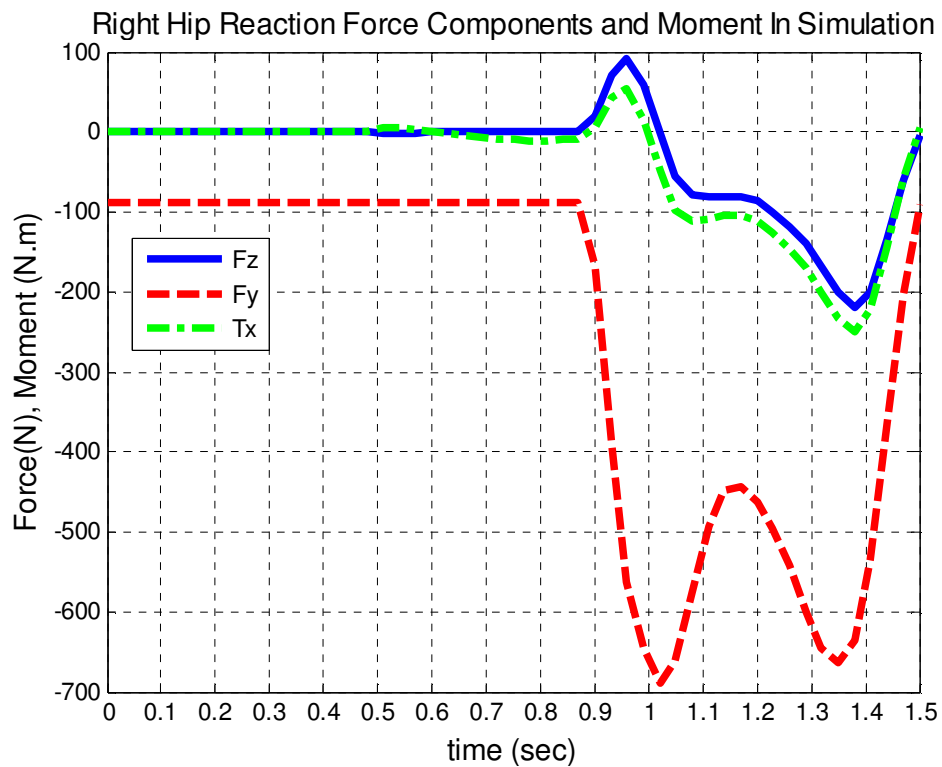


Figure 12. Right hip forces and moment in simulation.

touching the ground, which is about 60% of the cycle time.

RESULTS

The force and torque components of left ankle, knee and hip joints were recorded from simulation, then they were plotted against the real measured values taken from literature in Figure 7, Figure 8 and Figure 9 respectively for further comparison. Table 2 showed the Root Mean Squared Error (RMSE) for force and torque components Fz, Fy and T for the left ankle joint, left knee joint and left hip joint.

Moreover, the output data for right ankle, right knee and right hip joints were recorded and presented in Figure 10, Figure 11 and Figure 12, respectively.

DISCUSSION

Figure 7, Figure 8 and Figure 9 represent a comparison between the force and torque components resulted from the model simulation compared to the same data acquired from literature for the left ankle, left knee and left hip joints.

In these Figures; Figure 7, Figure 8 and Figure 9 characterize the dynamics of the left ankle, left knee and left hip joints; the vertical force component Fy shows a good tracking to the literature values in stance phase, while a little deviation in swing phase. The horizontal force component Fz shows a good tracking to the literature values in swing phase, while in stance phase it shows a little deviation for the ankle joint. With respect to the torque components, they follow the literature values with a small difference.

Table 2 shows the RMSE for the force and torque components related to the three joints. Overall, the Fy component has the higher RMSE and T has the lower RMSE for the same joint. Further, the hip joint has the higher RMSE for the Fz, Fy and T components.

These errors may be due to approximating body segments to simple engineering shapes, treating the upper parts of the body as one segment and neglecting the third ground reaction force components as it is too small. The model needs some enhancement to catch all the detailed characteristics of human normal walking.

Figure 10, Figure 11 and Figure 12 represent the force and torque components resulted from the model simulation for the right ankle joint, right knee joint and right hip joint. These figures show that the right leg has the same performance such as the left leg but with a time shift. This is reasonable as there is a 50% shift between human legs in motion.

CONCLUSION

The model developed in this work was built to simulate the human lower limbs' dynamics in normal walking for the purpose of controlling the human bilateral prosthesis. The model shows the same overall performance of human legs in normal walking. Model inputs are position data and the outputs are the motion dynamics at the lower legs' three joints. Body motion could be animated using the model. The model can be modified by changing the weight and moment of inertia of the body segments. The model could be used to simulate normal walking, running, stair ascending and descending and any human sportive activity depending on the measured biomechanical data. In the next step of this research the model will be applied to the control of human bilateral prosthesis.

Conflict of Interest

The author has no conflicts of Interest to declare concerning the contents of this manuscript.

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