DEVELOPEMENT OF GAIT GENERATION SYSTEM FOR A LOWER LIMB PROSTHESIS USING MEASURED GROUND REACTION FORCES

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ABSTRACT

Lower limb dynamic models are useful to investigate the biomechanics of the knee and ankle joints, but many systems have several limitations which includes simplified forces, non physiologic kinematics and the complicated interactions between the foot and the ground. Many approaches in control of prosthetic devices use prediction algorithms to estimate ground reaction forces (GRF), which can degrade the performance and the efficiency of the devices due to calculation errors. In this study, the variation of the GRF during different gait cycles was investigated in the design of an adaptive fuzzy controller for a dynamic model of an active ankle-knee prosthesis, the efficiency of the controller was tested for walking gait, stair ascent and descent. Real experimental kinematics of the lower limb and GRF measured by forces platforms were selected as the controller inputs and fuzzy reasoning was used to determine the adequate torques to actuate the prosthetic device model. The capacity of the active prosthesis and the designed controller to provide walking, stair ascent and descent cycles was tested by comparing the gait kinematics to those provided by a healthy subject.

KEYWORDS

Knee, Active Prosthesis, Fuzzy Takagi-Sugeno Control, PID, Reaction Forces.

1. INTRODUCTION

During gait, the ground provides reaction forces responsible for maintaining gait balance. These forces depend on different factors like ground contact surface condition, slope, elevation of the terrain and gait types [1]. These forces have been recently used in gait analysis and classification: an approach to movement recognition, using the vertical component of a person's GRF was presented in [2], typical movements such as taking a step, jumping, drop-landing, sitting down, rising to stand and crouching were decomposed and recognized using the vertical component of the GRF signal measured by a weight sensitive floor. Another approach in [3] presented a comparison of gait phases detected from the data recorded by force sensing resistors mounted in the shoe insoles. A new method based on GRF measurements of human gait was used for subject recognition [4]; the classification task is accomplished by means of a kernel-based support vector machine. A study in [5] used also GRF in feature recognition based on fuzzy reasoning. GRF measurements were also a helpful tool in flat foot problem diagnosis and were used as a criterion to discriminate between normal and flat foot subjects [6]. Other techniques were also used to provide lower limb assistance like MRI which used collected clinical human data to estimate joints kinematics [7] [8]. In another field, many researchers considered GRF variations in designing lower limb prosthetic devices given its great effect on the stability, balance, and energy consumption of the prosthesis and consequently the user's stability, so they have to be considered

in the control strategy. Accelerometers and gyroscopes have been used to detect an elevation change of the ground in [9] [10] [11]. A method using gyroscopes and infrared sensors was used in [12] to estimate the ground slope and elevation of the foot above the ground. A Terrain Recognition System was presented in [13] to estimate the height and slope of the terrain using laser distance sensors. In some few studies, complete hierarchical controllers were established. These controllers perceive the users locomotive intent based on signals from the user, environment as a first step, then translate this information to a desired output state for the device used as a reference input in a specific control loop that executes the desired movement [14] [15] [16] [17] [18] [19] [20] [21][22].

In this paper, experimental data of GRF combined with healthy limb data is used to generate the adequate gaits for lower limb prosthesis. The collected data is fed to a Fuzzy Inference System (FIS) that actuates a dynamic model of the prosthetic device. The proposed strategy is evaluated for different types of gaits such as walking, stair ascent and descent and validated by simulation results. Mathematical equations of human lower limb model, consisting of two degrees of freedom (Knee and ankle joints), are presented in a first section. In the second section gait analysis is provided for three different gait activities: walking, stairs ascent and descent. Then the fuzzy controller is evaluated for these different cycles and the results are presented and discussed finally.

2. Dynamics of the Lower Limb Active Prosthesis

2.1. Link-Segment Diagram

In this section, the dynamics of the ankle-knee prosthesis is modeled by a link-segment diagram. Since total human control is assumed at the biological hip joint of the residual limb, this paper focuses on the dynamics and control of the ankle-knee prosthesis.

Fig.1 shows the link-segment representation of the leg on the amputated side in the sagittal plane. The segment lengths are L_1 and L_2. Locations of the segment centers of gravity (cog) are represented by r_1 and r_2. Horizontal feet disturbances are assumed as a fixed acceleration [redacted] of the ground contact point. τ_1 and τ_2 are the joint torques corresponding to the ankle and the knee respectively. θ_1 and θ_2 depict respectively ankle and knee angles. F_1 and F_2 represent the horizontal and vertical components of ground reaction force applied to the prosthetic ankle joint. These forces, caused by the interaction of the foot with the terrain, influence highly the gait cycle and have a critical role in supporting the body weight, ensuring stability, and providing the necessary propulsion for the gait. The only inputs to the model are externally applied GRF and ankle and knee joint torques.

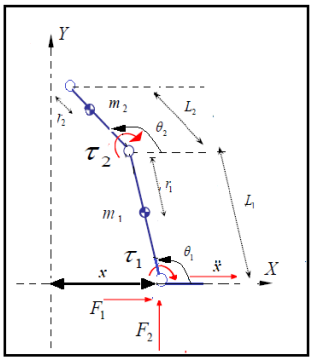


Figure 1. Knee-Ankle prosthesis model.

2.2. Euler-Lagrange Equations

The dynamics of the prosthesis are derived using the Euler-Lagrange approach [20] for a nominal gait cycle. To utilize the Lagrangian method, Cartesian coordinates of the cog for each link, *(x_1, y_1)* and *(x_2, y_2)*, are defined as:

$$\begin{cases} x_1 = x + r_1 \cos\theta_1 \\ y_1 = r_1 \sin\theta_1 \end{cases} \tag{1}$$

$$\begin{cases} x_2 = x + r_2 \cos\theta_2 + L_1 \cos\theta_1 \\ y_2 = r_2 \sin\theta_2 + L_1 \sin\theta_1 \end{cases} \tag{2}$$

The time derivative of displacement of the cog for each link is calculated according to (1) and (2):

$$\begin{cases} \dot{x}_1 = \dot{x} - r_1 \dot{\theta}_1 \sin\theta_1 \\ \dot{y}_1 = r_1 \dot{\theta}_1 \cos\theta_1 \end{cases} \tag{3}$$

$$\begin{cases} \dot{x}_2 = \dot{x} - L_1 \dot{\theta}_1 \sin\theta_1 - r_2 \dot{\theta}_2 \sin\theta_2 \\ \dot{y}_2 = L_1 \dot{\theta}_1 \cos\theta_1 + r_2 \dot{\theta}_2 \cos\theta_2 \end{cases} \tag{4}$$

The kinetic energy of the whole system, T, is the sum of kinetic energy of individual links, and can be written as:

$$T = \frac{1}{2} m_1 \left(\dot{x}_1^2 + \dot{y}_1^2\right) + \frac{1}{2} I_1 \dot{\theta}_1^2 + \frac{1}{2} m_2 \left(\dot{x}_2^2 + \dot{y}_2^2\right) + \frac{1}{2} I_2 \dot{\theta}_2^2 \tag{5}$$

The total potential energy of system, U, can be obtained by:

$$T = \left(m_1 y_1 + m_2 y_2\right) g \tag{6}$$

The Lagrangian, L, is a scalar function that is defined as the difference between kinetic and potential energies of the mechanical system:

$$L = T - U \tag{7}$$

The equations of motion for the prosthesis are derived using the Lagrangian in equation (7) and the following equations:

$$Q_{tot} = \frac{d}{dt}\left(\frac{\partial L}{\partial \dot{q}}\right) - \frac{\partial L}{\partial q}, \ \mathrm{q} = \theta_1, \theta_2, x \tag{8}$$

From (7) and (8), the equations of motion can be written as:

$$\begin{aligned}\tau_1 &= \ddot{\theta}_1\left(m_1 r_1^2 + I_1 + m_2 L_1^2\right) + \ddot{\theta}_2\left(m_2 r_2^2 + I_2\right) - (m_1 r_1 + m_2 L_2)\ddot{x}\sin\theta_1 - m_2 r_2 \ddot{x}\sin\theta_2 \\ &+ m_2 r_2 L_1 (\ddot{\theta}_1 + \ddot{\theta}_2)\cos\left(\theta_2 - \theta_1\right) + m_1 r_1 g\cos\theta_1 + m_2 L_1 g\cos\theta_1 + m_2 r_2 g\cos\theta_2 + F_2 L_1 \sin\theta_1 \\ &+ F_2 L_2 \sin\theta_2 - F_1 L_1 \cos\theta_1 - F_1 L_2 \cos\theta_2\end{aligned} \tag{9}$$

$$\begin{aligned}T_2 &= \ddot{\theta}_2\left(m_2 r_2^2 + I_2\right) - m_2 r_2 \ddot{x}\sin\theta_2 + m_2 r_2 L_1 \ddot{\theta}_1 \cos\left(\theta_2 - \theta_1\right) + m_2 r_2 L_1 \dot{\theta}_1^2 \sin\left(\theta_2 - \theta_1\right) \\ &+ m_2 r_2 g\cos\theta_2 - F_1 L_2 \cos\theta_2 + F_2 L_2 \sin\theta_2\end{aligned} \tag{10}$$

$$F_1 = \left(m_1 + m_2\right)\ddot{x} - \left(\ddot{\theta}_1 \sin\theta_1 + \dot{\theta}_1 \cos\theta_1\right)\left(m_1 r_1 + m_2 L_1\right) - m_2 r_2\left(\ddot{\theta}_2 \sin\theta_2 + \dot{\theta}_1 \cos\theta_2\right) \tag{11}$$

3. GAIT ANALYSIS

To ensure better control of the prosthesis, it is necessary, first, to study the biomechanical activity of the lower limb for different types of activities. In this paper, three types of gait will be studied: walking, stair ascent and descent. These activities are cyclical and can be broken down into several phases depending on the relationship between the foot and its contact point with the ground.

3.1. Walking Cycle

During the walking cycle, the considered lower limb alternates a support phase (foot in contact with the ground) and an oscillating phase (foot without contact on the ground). A walking cycle is thus composed of a support phase (approximately 60% of the cycle) and an oscillating phase (approximately 40% of the cycle) of the lower right and left limbs (Fig.2).

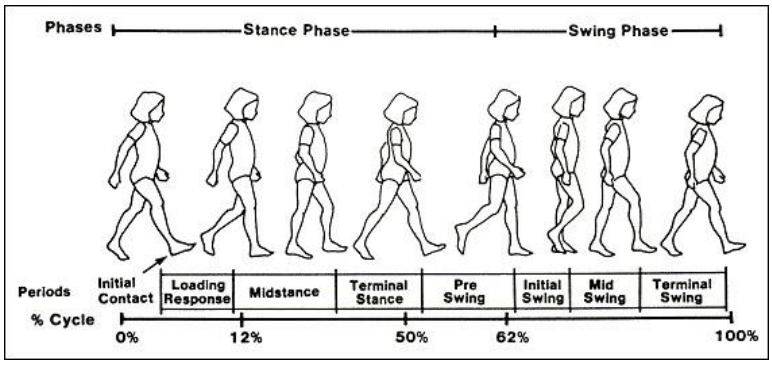


Figure 2. Sub-phases of the healthy leg during walking gait cycle.

3.2. Stair Descent

The descent cycle begins with body weight transfer on the lower limb of support, until the single limb support phase. The descent is performed by a flexion of hip and knee. The center of gravity begins to advance, simultaneously, the ankle controls the progressive advance of the tibia. Then, it moves vertically down. At this moment, the ankle is in extension and has a damping role. The weight is then transferred to the contra lateral lower limb to begin the oscillation phase (Fig.3).

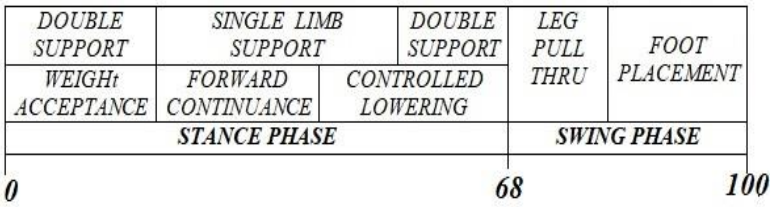


Figure 3. Sub-phases of the healthy leg during stair decent.

3.3. Stair Ascent

The cycle begins with the foot contact with the ground. Then, the weight of the body is transferred to the anterior leg, which extends through a concentric muscle contraction of the quadriceps. This causes a vertical displacement of the center of gravity. The unipodal support phase continues with anterior displacement of the center of mass associated with anterior flexion of the trunk and ends with the contact of the second foot with the ground. During the swing phase, the foot is ascended and placed on the upper step (Fig.4).

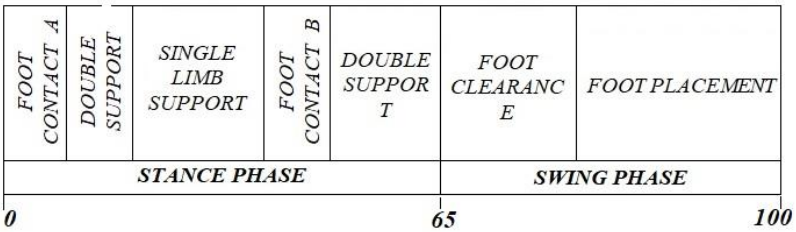


Figure 4. Sub-phases of the healthy leg during stair ascent.

4. CONTROL OF THE ACTIVE PROSTHESIS

In the approach proposed herein (Fig.5), measured GRF's combined with femur and tibia inclination angles are used as a training set for a FIS that enables the prosthesis model to follow a displacement profile similar to that of a natural leg during three types of gait.

The FIS can decide the gait mode to execute from the GRF pattern and the real data of the healthy leg. In this work, experimental data available in [23] and [24] are used to validate the approach. Two other PID controllers are used as secondary controllers for knee and ankle joints to correct the prosthetic position discrepancies from the desired values due to disturbances like uneven terrain [25].

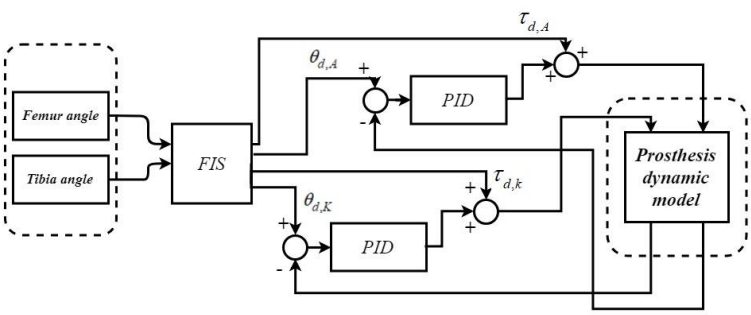


Figure 5. Control diagram for the Knee-Ankle prosthesis

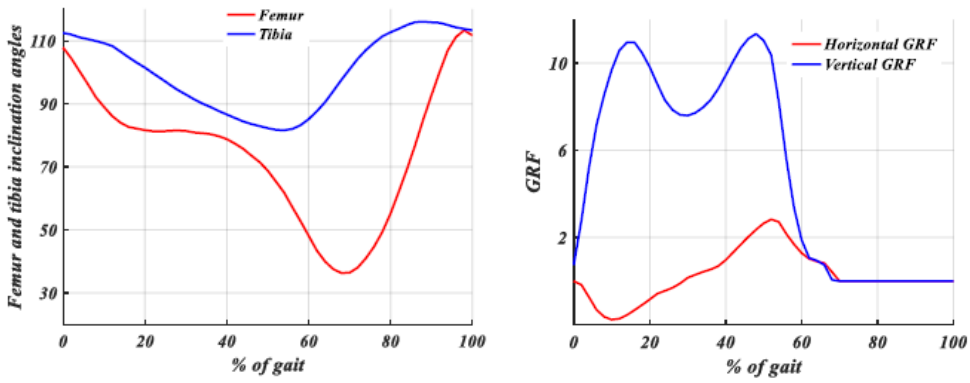
Based on the real human healthy limb data and GRF measurements, the fuzzy controller must predict which phase of the gait cycle the prosthesis is in and generate nominal torques $\tau_{d,k}$ and $\tau_{d,A}$ and reference trajectories for knee and ankle joints $\theta_{d,k}$ and $\theta_{d,A}$. The FIS incorporates the human-like reasoning style of fuzzy systems through the use of fuzzy sets and a linguistic model consisting of a set of IF-THEN fuzzy rules. The rules of the Takagi-Sugeno FIS are expressed as:

$$\begin{aligned}
&\mathrm{R}_1 : \text{IF } \mathrm{x}_1 \text{ is } \mathrm{Ph}_1^1 \text{ AND } \mathrm{x}_2 \text{ is } \mathrm{Ph}_2^1 \\
&\text{AND } \mathrm{x}_3 \text{ is } \mathrm{Ph}_3^1 \text{ AND } \mathrm{x}_4 \text{ is } \mathrm{Ph}_4^1 \\
&\text{THEN} \\
&\begin{cases}
\mathrm{y}_1^l(x) = c_{1,0}^l + c_{1,1}^l \mathrm{x}_1 + c_{1,2}^l \mathrm{x}_2 \\
\mathrm{y}_2^l(x) = c_{2,0}^l + c_{2,1}^l \mathrm{x}_1 + c_{2,2}^l \mathrm{x}_2 \\
\mathrm{y}_3^l(x) = c_{3,0}^l + c_{3,1}^l \mathrm{x}_1 + c_{3,2}^l \mathrm{x}_2 \\
\mathrm{y}_4^l(x) = c_{4,0}^l + c_{4,1}^l \mathrm{x}_1 + c_{4,2}^l \mathrm{x}_2
\end{cases}
\end{aligned} \tag{12}$$

where x_1, x_2, x_3 and x_4 are the FIS inputs (femur and tibia angles, vertical and horizontal ground reaction forces), $y_1^l(x), y_2^l(x)$, $y_3^l(x)$ and $y_4^l(x)$ represent first, second, third and fourth outputs of the ith rule corresponding to the torques and the angular positions of the prosthetic knee and ankle joints, $c_{1,0}^l, c_{1,1}^l, c_{1,2}^l, c_{2,0}^l, c_{2,1}^l, c_{2,2}^l, c_{3,0}^l, c_{3,1}^l, c_{3,2}^l, c_{4,0}^l, c_{4,1}^l, c_{4,2}^l$ are consequent parameters of the first and second outputs respectively, for the lth rule; and phi_j^i is defined as the phase fuzzy state i for the jth input.

Using the training data to tune the controller, the output parameters can be achieved readily by the Least Square (LS) optimization method assuming the shapes and parameters of all the input fuzzy sets fixed ahead of time.

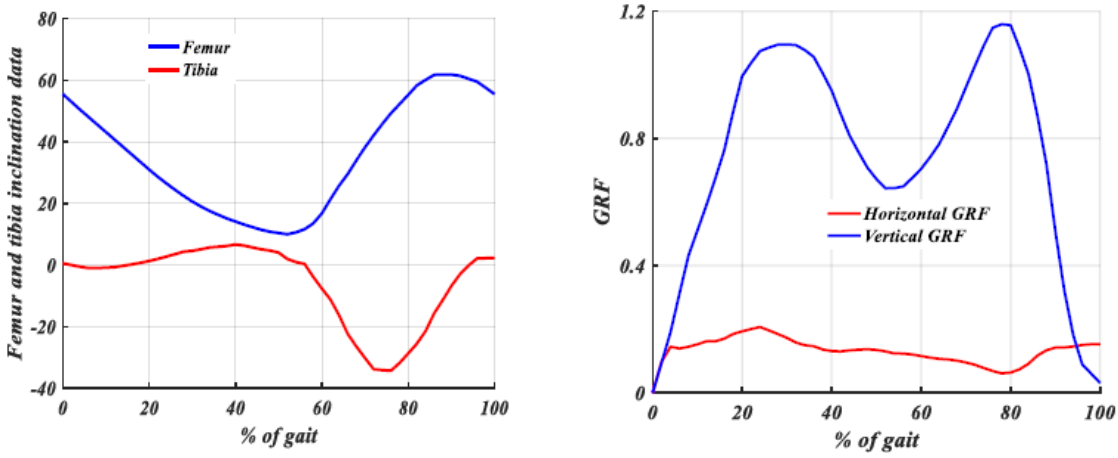
The inputs of the Takagi-Sugeno FIS are the inclination angles of the tibia and femur and the horizontal and vertical GRF's. Training data were selected from [19] for the walking gait cycle and from [20] for the stairs ascent and descent cycles which are partitioned based on sub-phases of the gait cycles of a healthy leg as shown in Fig.6, Fig.7 and Fig.8.



(a) Healthy femur and tibia inclination angles

(b) Horizontal and vertical ground reaction forces

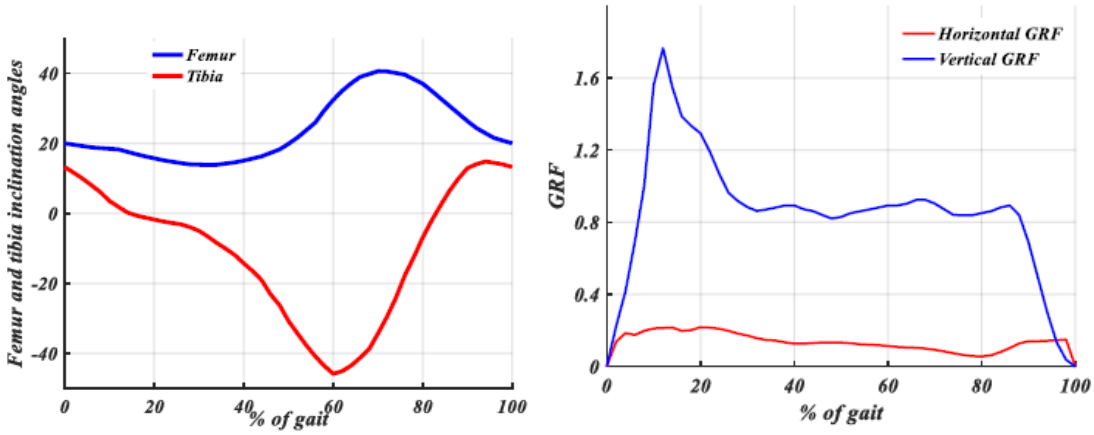
Figure. 6: FIS inputs for walking gait.



(a) Healthy femur and tibia inclination angles

(b) Horizontal and vertical ground reaction forces

Figure. 7: FIS inputs for stair ascent.



(a) Healthy femur and tibia inclination angles

(b) Horizontal and vertical ground reaction forces

Figure. 8: FIS inputs for stair descent.

4. SIMULATION RESULTS

The anthropometric parameters of the human segments which are function of the mass and the length of the body were adopted from the anthropometric data given by [23] and shown in Fig.9 and Table.1. The PID parameters are achieved by online adjustments to enhance the process performance.

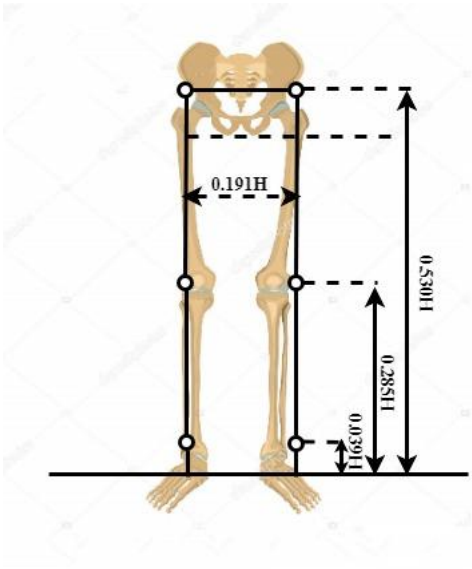
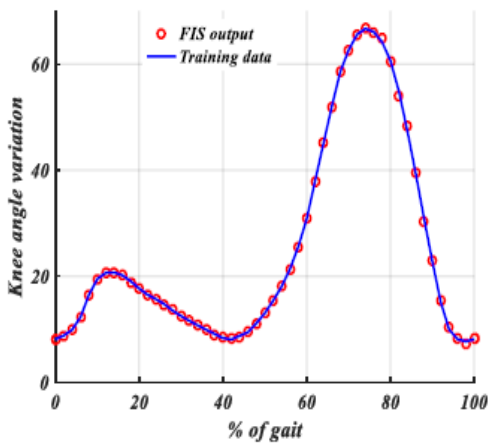


Figure. 9: Human lower limb segments lengths

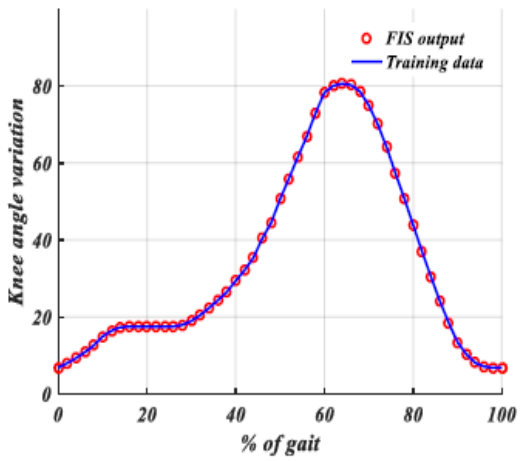
Measure	Shank	thigh
Mass	m1 = 0.0485 *m	m2 = 0.1 *m
Moment of inertia	I1 = m1 _ $(0.302 * L1)^2$	I2 = m2 _ $(0.302 * L2)^2$
Center of mass	r1 = 0.433 *L1	r2 = 0.433 *L2

Table.1: Anthropometric data for the human lower body.

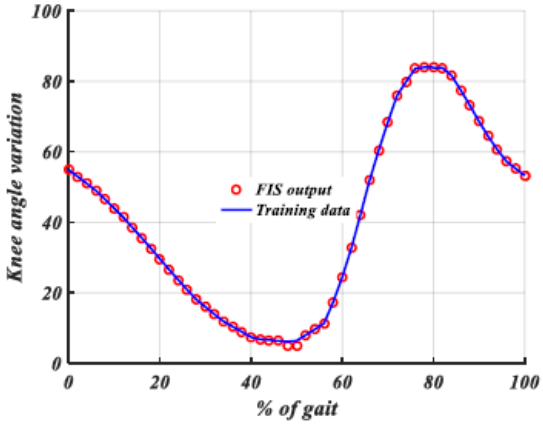
Comparison between FIS outputs and training data are illustrated in Fig.10, Fig.11, Fig.12 and Fig.13. The FIS outputs achieved very closely the desired mapping data. The results proved that adding GRF as inputs for the controller has enhanced the performance of the FIS.



(a) Walking

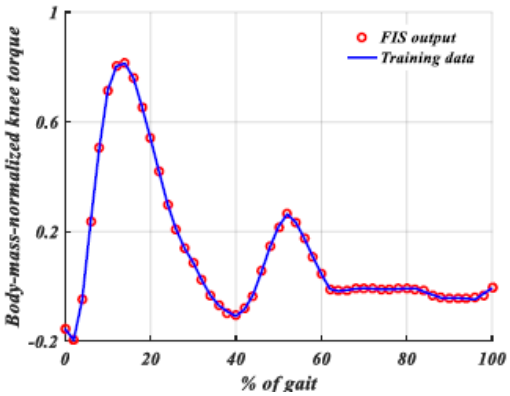


(b) Stair descent

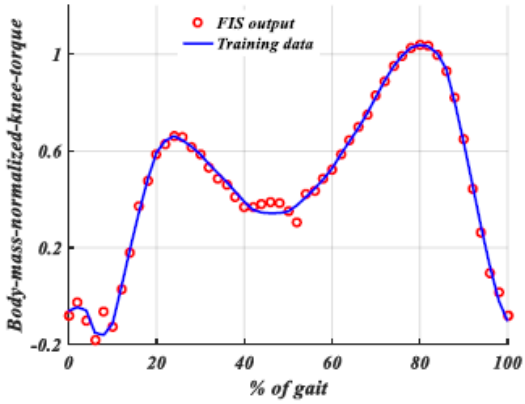


(c) Stair ascent

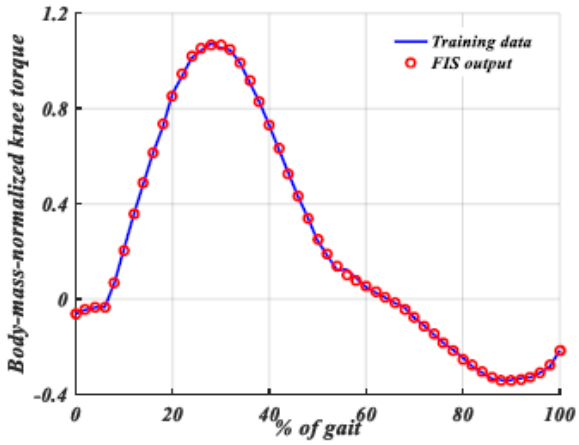
Figure. 10: Takagi-Sugeno FIS first output vs. desired outputs for the different gait cycles.



(a) Walking

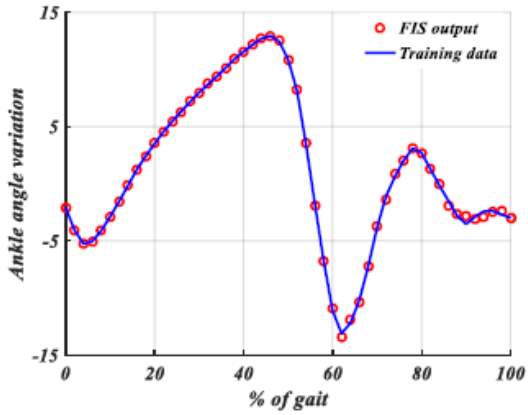


(b) Stair descent

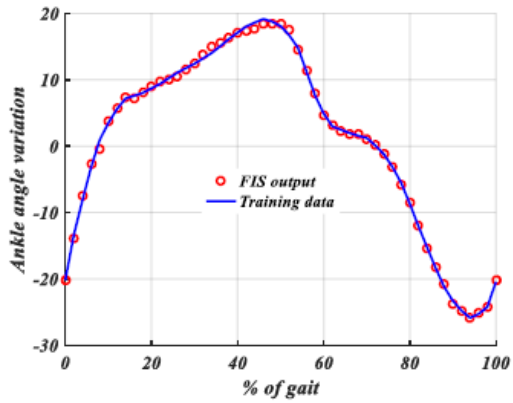


(c) Stair ascent

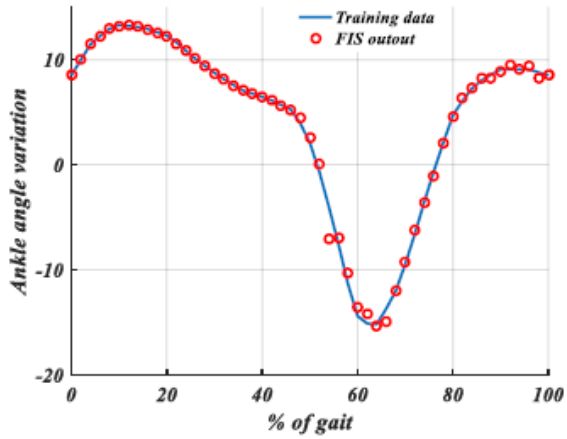
Figure. 11: Takagi-Sugeno FIS second output vs. desired outputs for the different gait cycles.



(a) Walking

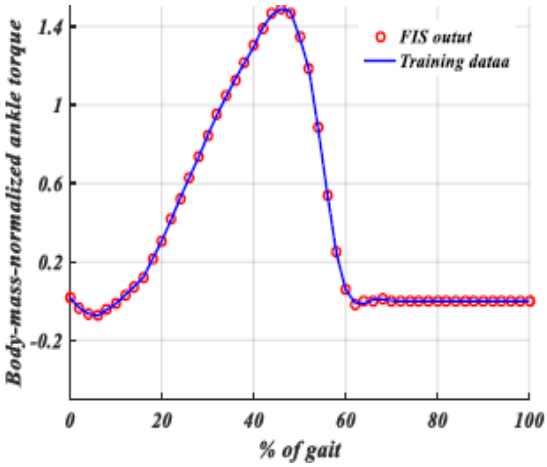


(b) Stair descent

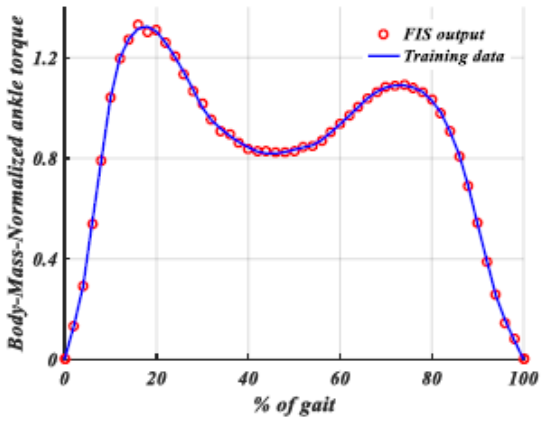


(c) Stair ascent

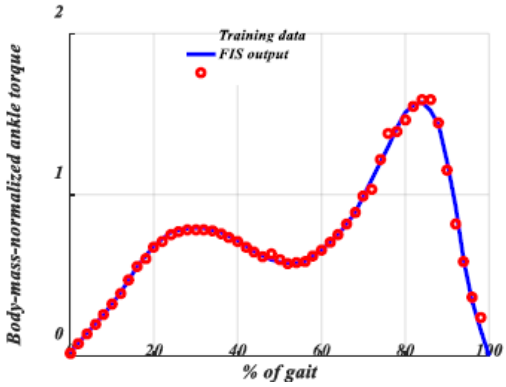
Figure. 12: Takagi-Sugeno FIS third output vs. desired outputs for the different gait cycles.



(a) Walking



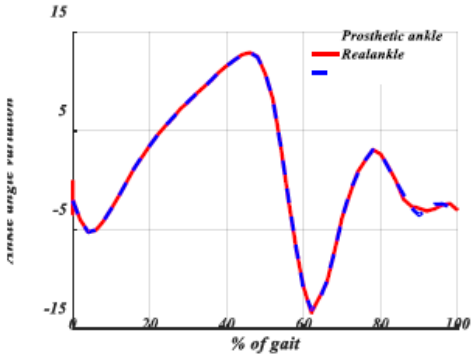
(b) Stair descent



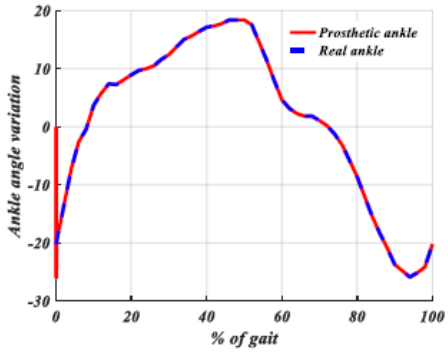
(c) Stair ascent

Figure. 13: Takagi-Sugeno FIS fourth output vs. desired outputs for the different gait cycles.

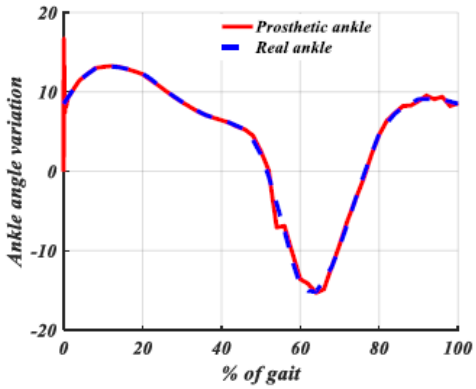
The performance of the controller is demonstrated through the closed-loop simulation and depicted in Fig.14 and Fig.15 illustrating the evolution of the prosthetic knee and ankle angles variations compared to real healthy leg joints angles for the different gait cycles. The developped system is able to reproduce the real limb joint angles for the three types of gait assuring stability and good mapping for reference trajectories.



(a) Walking

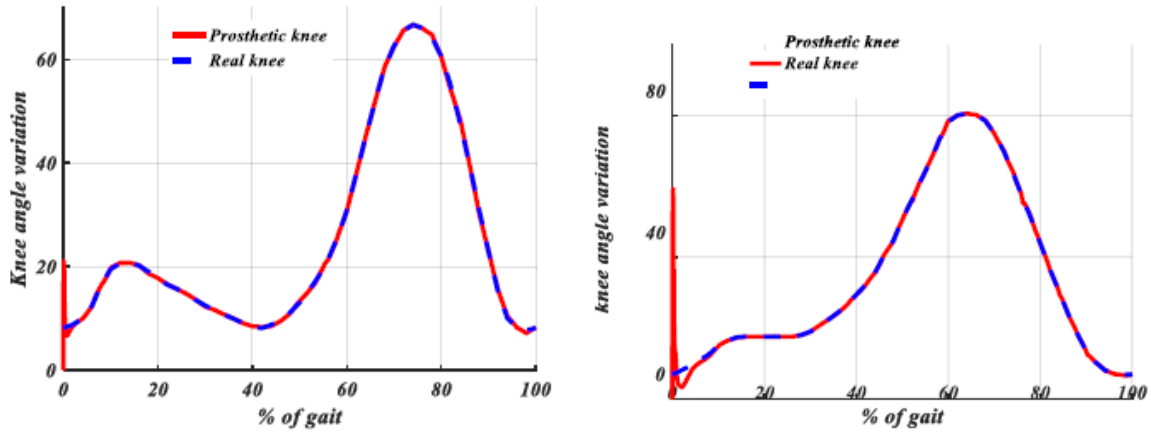


(b) Stair descent

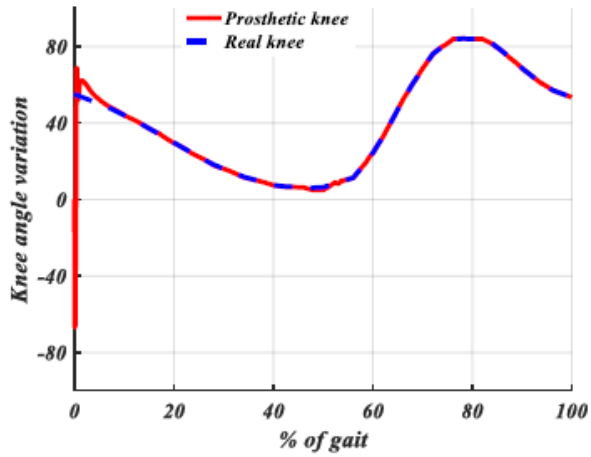


(c) Stair ascent

Figure. 14: Ankle angle variation for the different gait cycles.



(a) Walking (b) Stair descent



(c) Stair ascent

Figure. 15: Knee angle variation for the different gait cycles.

The two secondary controllers corrected the prosthetic position discrepancies from the desired values due to environmental disturbances.

5. CONCLUSION

In this paper, the variation of the GRF during gait cycle was used for the design of an adaptive fuzzy controller for a dynamic model of active ankle-knee prosthesis. The main idea of the proposed strategy was to use collected GRF's from force platforms to generate the adequate gait mode for the active prosthesis. A dynamical model for the human lower limb considering GRF's was established, using euler-lagrange equations, in order to simulate the human motion dynamics. The model parameters were extracted from the anthropometric data of a healthy human body. From the variation of GRF's and healthy leg data, the controller can decide in which gait phase is in and actuate the prosthetic joints with the appropriate torques.

Simulation results indicate that using GRF's to train the FIS combined with femur and tibia angles has enhanced the efficiency of the controller in tracking the real human lower limb dynamic motions and generated prosthetic joint angles mimicking the real joint angles variations. This behaviour was guaranteed for the three different activities (walking, stairs ascent and

descent). The mechanical design of the prosthetic device and its clinical validation was not investigated in this work and can be topic of further researches

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