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An Assistive Passive Pelvic Device for Gait Training and Rehabilitation Using Locomotion Dynamic Model

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Abstract

In this paper, we propose the design of a novel gait rehabilitation passive device to training the people who lost their walking ability due to neurological injuries, such as spinal cord injury and stroke. As known, the movements of pelvis play an important role in balance and propulsion during a gait cycle. We simulate a passive orthosis to assist the motion of pelvis of gait patients. We expect that the proposed orthosis, connected to pelvis, assists the patients to regain initial gait pattern by creating the rotations of pelvis close to reference ones during ambulation on treadmill. This optimized pelvic device is un-motorized and contains only optimum passive elements. However, with this device, we consider a Body Weight Support (BWS) to unload a portion of human body weight during treadmill training. The design is accomplished based on a proper dynamic model of human body by focusing on motions of lower leg body segments. The structural configuration of device is determined during an optimization process defined on dynamic model.

The simulation results show the time trajectories of pelvic rotations are close to reference trajectories and the optimum passive device can create the desired motion of pelvis without any effort of subject. Also, the results of sensitivity analysis of device by varying the optimum design parameters by 1–3 percent in addition to variation in anthropometry parameters of subject by 1–5 percent show the robust and decent performance of pelvis orthosis during treadmill gait.

Keywords: Pelvic Rotations, Gait Analysis, Passive Device, Dynamic Model, Optimization.

1. Introduction

There are a large number of people who are suffer from disability in walking because of occurring some injuries such as spinal cord injury and stroke. As known, to traditional BWS treadmill training of these patients, a team of four therapists are required. Two of them derive the legs; one assists the motion of pelvis to provide the balance and stability, thereby possibility of motion of swing leg. The other one is operator of treadmill and BWS system. During such manual gait therapy assistance, intensive effort is applied to trainers which can lead to reduction in effectiveness of therapy process and also probable injuries. These motivate the design

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of suitable devices that can assist gait patients in correct movement during walking. However, by development of robotic gait rehabilitation, the problems of manual traditional rehabilitation methods are reduced.

Pelvis has an important role during locomotion. It transfers forces from the lower extremity to the trunk and helps in forward propulsion during ambulation [1]. Pelvis also places the Center of Mass (COM) of the trunk above the support foot to allow the motion of swing leg [2, 3]. It also modulates the vertical motion of the COM of the body to reduce energy consumption during walking [3]. Hence, attention to natural pelvic motion is essential in gait rehabilitation as a therapist is required to control the motion of pelvis during gait training. Providing of desired movements of pelvis led to improving of walking performance for patients with gait impairments.

Development of pelvic orthoses is a relatively new topic in the field of robotic rehabilitation to providing correct locomotion performance. Aoyagi et al. developed a robotic device, PAM (Pelvic Assist Manipulator) that assists the human pelvic motion during treadmill training. PAM allows natural motion of the pelvis through six actuated pneumatic cylinders [4]. Stauffer et al. [5] introduced a new device for paraplegics, the WalkTrainer. This device is composed of a leg and pelvic orthosis, an active BWS and closed loop muscle stimulation. Pelvic trajectories are measured and programmed on the WalkTrainer and applied to subjects by mean of the pelvic orthosis. Pietrusinski et al. [6] designed a Robotic Gait Rehabilitation (RGR) trainer, which apply corrective force fields to the pelvis depending on patient's efforts based on deviations from normal pelvic motion. This device applies the forces only in vertical direction via an orthopedic pelvic brace to create pelvic obliquity. Utilizing two linear slides, the motion of pelvis in horizontal plane is left free. Trieu Phat Luu et al. [7] presented an adaptive BWS apparatus which supports the pelvic motion during the gait training process. This mechanism is designed to control pelvic motion based on a planned reference trajectory with capability of active body weight support.

All gait rehabilitation devices to assist pelvic motion already presented are derived by one or more actuators. *To the best knowledge of the authors, purely passive devices that improve the motion of pelvis during walking have not been explored*. As known, the passive devices are constructed from only passive elements without employing any actuator, sensor and controller. Furthermore, the passive devices can be safer and economical, compared to active devices. In this paper, we present an innovative application of optimized passive elements (such as masses and springs) in design of a pelvic assistive device to assist and/or improve the human gait. This device would be connected to the pelvis and move it in an appropriate trajectory during the gait cycle. In contrary to design of resent pelvic rehabilitation robots which are kinematic based, the proposed passive design is dynamic model based. We present an appropriate dynamic model for human body focusing on lower leg movements. In this work, we close the motion of the pelvis to a desired motion by optimization of design parameters of device. This philosophy is similar to that used in the design of the swing-assist exoskeleton presented in [8]. In comparison to our initial theoretical work presented in [9], the dynamic model of lower legs is more complete and closer to reality by considering the motion of foot as a rigid body (not a point mass in end of shank) with 3D motion and also determination of location of hip joint centers in pelvis. Whiles, in former research works, the desired trajectory of motion of lower legs had been estimated for subject, in present simulation work, we provide real desired trajectories using gait pattern of subject obtained in a gait analysis laboratory. Furthermore, here, we consider and study the double support phase of gait in our problem. Also, to contemplate the fabrication issues, we consider some feasible bounds for the device parameters during optimization.

The organization of this paper is as follows: In Section 2, we summarize the mechanical construction of the proposed device having passive elements. In Section 3, the dynamic model of lower legs and upper body is discussed. We describe the kinematic relationship of body segments and apply Newton-Euler approach to presented model in Section 4. Next, we explain the details of gait analysis; outline the optimization problem for several cases and present simulation results in Section 5. We study the effect of variation of parameters as sensitivity analysis in section 6. Finally, discussion of results and conclusion are presented.

2. System Description

We aim to present a passive device creating the desired pelvic motion by fixing to pelvis during gait on treadmill. This proposed device comprises a set of springs and masses connected to pelvis through relevant rods and a pelvis corset, as shown in Figure 1. Each spring is connected at one end to a rod and it's another end is fixed. In addition, we predict a BWS system to unload a portion of subject's weight during walking. As shown in Figure 1, we consider a reference right-hand coordinate system *XYZ* on inertial frame as the *X*-axis is defined along direction of the treadmill with its positive direction anterior and the *Y*-axis is vertically upwards, against the direction of gravity. The origin point of global reference frame is determined at heel-strike point of right leg of selected subject for simulation. Also, a local frame of *xyz* in coincidence to pelvis anatomical axes (explained in next sections) is defined on pelvis as shown in Figure 1.

In our proposed passive structure, the details of design parameters are listed below for i = 1…n, where, n is the number of rods:

β_i: Angle between the *z*-axis and i^th rod fixed to pelvis.
k_i: Stiffness constant of i^th spring.
δ_i: Initial stretch of i^th spring.
$x_{o_i}, y_{o_i}, z_{o_i}$: Coordinates of the fixed point of i^th spring in the inertia coordinate frame.
m_i: Mass of i^th point mass.
d_{mi}: Position of i^th point mass on the i^th rod.
d_{si}: Location of fixed point of i^th spring on the i^th rod.
f_g: Vertical body support weight.

Considering above parameters, the number of design parameters of related optimization problem is 9n+1. The limitations of fabrication process of device and interface issue of user and device led to bounding of design parameters to lower and upper values.

Considering these constraints on our passive design, a reasonable question to ask is: 'How many springs, point masses and rods are needed to use in the design?' Using 'trial and error' approach, we obtain n = 6. We studied the optimization problem with smaller number of springs, point masses and rods; however, we could not obtain adequate performance for device during simulation for n < 6. Consequently, we have 55 design parameters in our optimization problem. The device with optimized selected parameters can create the motion of pelvis close to desired configuration during ambulation on treadmill, passively.

3. Dynamic Model

In this work, we present a proper dynamic model which has an acceptable consistency with the human body motion. The structure of pelvis and lower extremities

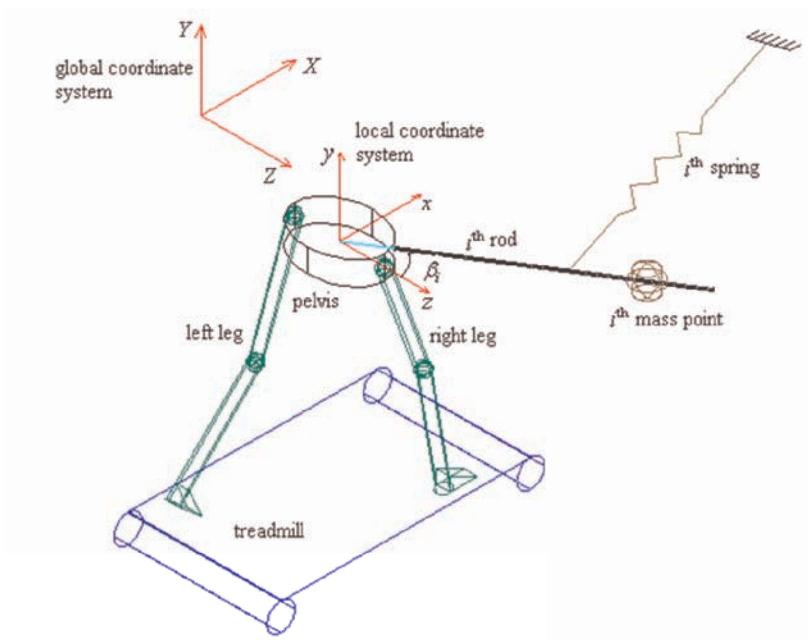

Figure 1. The 3D schematic of the passive device connected to pelvis.

in addition to their joints is observed in sample skeleton of lower legs shown in Figure 2. The pelvis and each of thigh, shank and foot are considered as rigid bodies which are connected together via assumed spherical joints. By these assignments, the actual kinematics of lower legs is justified. We have tried to have a real model of pelvis, swing and stance legs.

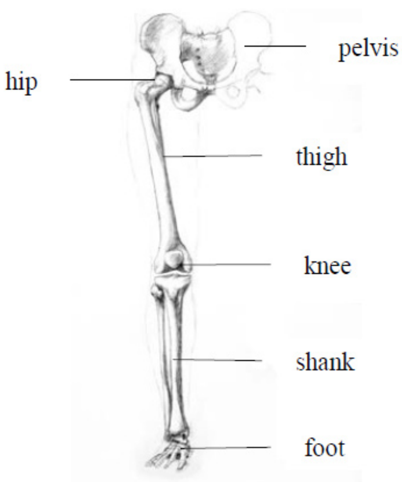

Figure 2. The skeleton of pelvis and lower extremities.

In our model, we consider the pelvis as a rigid body with three orientation DOFs, i.e. rotation, tilt and obliquity; and three translational movements, i.e. medial-lateral, fore-aft and vertical displacements. Also, we assume the pelvis to be a cylinder with ellipsoid section and a uniform mass distribution. In addition, we consider inertial, physical and geometrical features of the other lower limbs i.e. the thighs, shanks and foots as rigid bodies as presented in [10] for lower legs.

As known, from a 3D kinematic study of gait cycle, each hip joint has three angles of flexion/extension, abduction/adduction and rotation, each knee joint has three angles of flexion, varus/valgus and rotation and each ankle has dorsi/plantarflexion, progression and rotation angles [11]. Therefore, as shown in Figure 3, in our model, all joints of segments consisting hip, knee and ankle of swing and stance legs as well as the contact point of stance leg to ground or treadmill are considered spherical. By having this complete model, it is possible to have three orientation angles for any segment resulting in feasibility of three-dimensional gait analysis of human body. It may be noted that the spherical joint between stance foot and treadmill is considered as a moving joint under foot in coincidence with the point

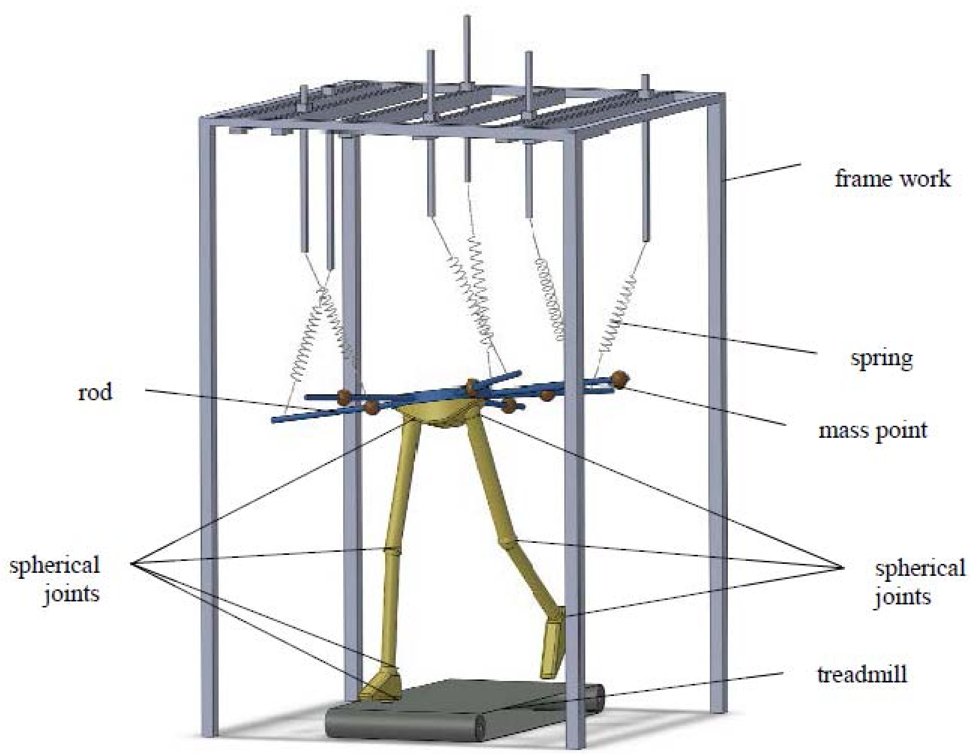

Figure 3. Dynamic model of legs and pelvis connected to passive device during walking on treadmill.

of application of treadmill reaction force vector on foot during stance phase. This unknown force vector, which can be determined during the solution of related dynamic problem, starts at the lateral border of heel at initial contact and moves along the center of the foot and finishes under hallux at toe-off.

In this work, we model the effect of upper body as a centralized force on pelvis center due to its vertical inertia force and its weight. The vertical acceleration of upper body is considered equal to vertical acceleration of pelvis center. Although the range of motion of upper body is very small during walking on treadmill but we should consider the dynamic effect of upper body particularly in vertical direction because of its considerable mass. However, we don't consider dynamic effect for the hands since they are assumed to be attached to walking frame during the gait cycle. Furthermore, we use a variable BWS to overcome an optimized portion of body weight. The effect of BWS can be modeled as an upward vertical force on pelvis due to its weight and an inertia force due to vertical acceleration which has the same magnitude of vertical acceleration of the center of mass of pelvis but in opposite direction during gait on treadmill.

As known, the double support phase is occurred twice during a gait cycle and includes nearly 20 percent of cycle. Thus, elimination of this phase of gait causes an incorrect dynamic model of the gait. In presented work, we didn't model the double support phase of gait cycle as in reality when two legs are in stance phase and fixed to treadmill. However, we maintain the double support phase using a single support model considering the variation of reaction force vector when the stance leg enters in double support interval. By study of diagram of reaction forces measured by the force platform in clinical gait analysis labs, it can be observed that the ground reaction forces on stance foot (particularly vertical force) in double support time interval is decreased approximately linearly (until to get zero at the beginning of swing phase relevant to toe-off time) [10, 11]. We calculate the ground reaction forces of stance leg in hill-strike time of the other leg and then put the linear time function of reaction force vector on former stance leg assuming swing state for it in the time interval of double support. So, we encounter with a single support structure of lower legs during double support phase of gait. This simplifying assumption is considered because of complexity of its dynamics due to the closed loop configuration of system during double support phase. In addition, such dynamics leads to appearance of several singular points in numerical solution of governing differential equations during optimization process. Although, the proposed model for double support gait is not in coincidence to real closed loop structure perfectly, it facilitates dynamic analysis of model of system as well as optimization process of relevant optimizing problem. During each phase of gait, one of two legs is in contact with treadmill. This results in switching in the dynamic equations during optimization when the support leg changes during the gait cycle.

4. Dynamic Analysis

In this research work, we try to close the motion of pelvis to natural trajectories using the proposed passive device without applying any effort by subject. This is performed during an optimization process to reach an optimum pelvic orthosis. In each step, to compare the actual state with the desired state, we should obtain the actual motion of pelvis via dynamic analysis of presented model of body and device.

In this work, we aim to assist the motion of pelvis to get the improved gait pattern. Therefore, the natural motion of the lower extremities i.e. thighs, shanks and feet are considered to be provided by helping of two physical therapists or utilizing the other passive or active assistive devices such as Lokomat and/or LOPES[12] connected to legs during gait. So, in our dynamic model, we consider the prescribed motion for lower extremities by applying the desired trajectories during ambulation on a treadmill with fixed constant speed. By this restriction on dynamic model, the system only has three degrees of freedom relevant to the rotations of pelvis. Then, the generalized coordinates can be considered as the angular orientations of the pelvis, i.e., rotation (ψ), obliquity (θ) and tilt (ϕ). Rotation of the pelvis is about y-axis of global coordinate system in vertical direction, obliquity is the rotation of pelvis about x-axis of current coordinate system and tilt is rotation of pelvis about z-axis of current frame [13]. We assign the pelvic anatomical coordinate system based on current international standards as shown in Figure 4.

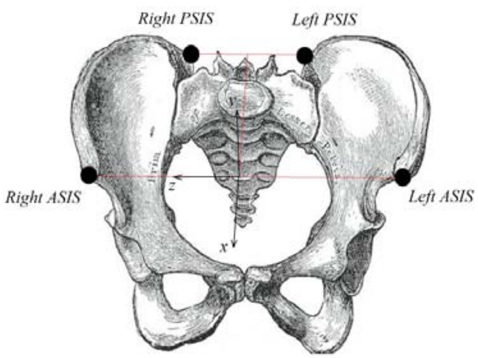

Figure 4. Pelvic anatomical frame: the midpoint between the anterior superior iliac spines (ASISs) is the origin point of frame. The *z*-axis is defined along the line passing through the ASISs with its positive direction from left ASIS to right one; the x-axis is defined in pelvic quasi-transvers plane specified by ASISs and midpoint between the posterior superior iliac spines (PSISs) and passes the midpoint between ASISs and is perpendicular to z-axis, with its positive direction anterior, the *y*-axis is perpendicular to both the x-axis and the z-axis and its positive direction is upward.

The rotation matrix of the pelvic anatomical coordinate system with respect to reference coordinate system is given by:

$$\mathbf{R}_p = \mathbf{R}_{y(\psi)}\mathbf{R}_{x(\theta)}\mathbf{R}_{z(\phi)} = \begin{bmatrix} c(\psi)c(\theta) & -s(\psi)c(\phi)+c(\psi)s(\theta)s(\phi) & s(\psi)s(\phi)+c(\psi)s(\theta)c(\phi) \\ s(\psi)c(\theta) & c(\psi)c(\phi)+s(\psi)s(\theta)s(\phi) & -c(\psi)s(\phi)+s(\psi)s(\theta)c(\phi) \\ -s(\theta) & c(\theta)s(\phi) & c(\theta)c(\phi) \end{bmatrix} \quad (1)$$

where, c and s stand for cos and sin, respectively. The relation between coordinates of a vector in current frame $[x, y, z]$ and inertial reference $[X, Y, Z]$ is given by the rotation matrix:

$$[X, Y, Z]^T = \mathbf{R}_p [x, y, z]^T \quad (2)$$

Also, using rotation matrix and its time derivative, the angular velocity vector of pelvis is expressed in pelvis coordinate system as:

$$\omega = \text{vect}(\mathbf{R}_p^T \dot{\mathbf{R}}_p) \quad (3)$$

Using the time derivative of Equation (3), one can obtain the angular acceleration of pelvis. Also, in gait analysis, using time trajectories of the optical markers attached to bony landmarks of segments of subject during walking on a treadmill, we could derive the rotation matrix, Euler angles, angular speed and angular acceleration of the segments, which can be considered as rigid bodies, versus time during walking on treadmill. As one would expect, the motion of the origin of pelvis frame depends on the posture of the stance leg and pelvis angles. To get the dynamic equations of model, first, the kinematic relation between the motion of pelvis and stance leg should be derived. To this end, we express the position, velocity and acceleration of COM of pelvis in terms of Euler angles and corresponding angular derivatives of motion of thigh, shank and foot of stance leg in addition to generalized coordinates of system i.e., ψ, θ, ϕ and their derivatives considering constant speed for treadmill. As mentioned, in this work, we consider desired prescribed time trajectories for 3D-motion of each segment of swing and stance leg during gait on treadmill. To create the movements of thigh, shank and foot of legs, we need sufficient number of force or torque actuators in presented dynamic model such that the system is not under-actuated. Therefore, in dynamic model, we assume the torque actuators in spherical joints of hip, knee and ankle of swing leg in addition to knee, ankle and ground reaction force point of stance leg. Thus, corresponding to any limb of each leg we have an actuator in system. We suppose each actuator creates a three-dimensional torque vector corresponding to three Euler angles of motion of relevant segment during walking. The torque of these assumed actuators in dynamic model is considered instead of force effect of probable physical therapists creating the motion of legs or any assistive orthosis (such as Lokomat) connected to legs of patient during rehabilitation process. However, even for a normal subject the force effects of muscles of limbs providing the movements of swing and stance legs can be replaced by assumed joint torques; Since the lever arms of most muscles are small in compared to segment lengths, they can be represented as torque motors[11].

As known, the angles of stance hip joint are corresponded to pelvis angles during gait. For a healthy subject, the muscles connecting pelvis to stance thigh are major muscles which create the pelvis motion during gait. In this paper, we expect that the proposed passive assistive device provides the motion of pelvis of a subject without muscle gait activity during gait on treadmill. Therefore, in our model, we release the hip joint of stance leg without any assumed torque actuator. *We try to present a passive device which compensates the lack of hip joint torques of patient by exerting proper forces to pelvis.*

The dynamic model of the system comprises rigid bodies consisting of the pelvis fixed to device, the thighs, the shanks and the feet of swing and stance legs. Here, Newton-Euler method is used to derive the equations of motion of the system. In Newton-Euler approach, we separate these rigid bodies and consider the force diagram for each one of them. We apply the Newton equation to each rigid body, presented in reference coordinate system of *XYZ*, written as

$$\sum \mathrm{F} = m\mathrm{a}_C = m(\ddot{X}_C\hat{i}_0 + \ddot{Y}_C\hat{j}_0 + \ddot{Z}_C\hat{k}_0) \quad (4)$$

where, $\hat{i}_0$, $\hat{j}_0$, $\hat{k}_0$ are unit vectors of reference coordinate system. Also, X_c, Y_c and Z_c are coordinates of COM of body in global frame. $\sum_{\mathrm{F}}$ is summation of the external forces on the rigid body consisting of joint reaction forces, weight, spring forces, and body support weight.

Since the equation of angular momentum of any rigid body is usually written in fixed local coordinate system, we prefer to state the Euler equation of the body about its COM in relevant anatomical coordinate system as follows:

$$\begin{aligned}
&\sum \mathbf{M}_C = \dot{\mathbf{H}}_C = \\
&= (\dot{H}_x - H_y\omega_z + H_z\omega_y)\hat{i} \\
&\quad + (\dot{H}_y - H_z\omega_x + H_x\omega_z)\hat{j} \\
&\quad + (\dot{H}_z - H_x\omega_y + H_y\omega_x)\hat{k} \qquad (5)
\end{aligned}$$

where, H_c is the vector of angular momentum of rigid body about its COM and ω_x, ω_y, ω_z are components of angular velocity in related local frame. The components of H_c along axes of local coordinate frame are expressed as

$$\begin{aligned}
H_x &= I_{xx}\omega_x + I_{xy}\omega_y + I_{xz}\omega_z \\
H_y &= I_{xy}\omega_x + I_{yy}\omega_y + I_{yz}\omega_z \\
H_z &= I_{xz}\omega_x + I_{zy}\omega_y + I_{zz}\omega_z \qquad (6)
\end{aligned}$$

Also, $\sum \mathrm{M}_c$ is the summation of moment of external forces about the COM of body as well as its joint torques.

We combine Newton-Euler equations of all rigid bodies to eliminate all joint reaction forces and torques and derive three dynamic equations corresponding to three DOFs of the system. The dynamic equations are presented in terms of generalized coordinates and prescribed movements of lower extremities as well as their time derivatives. It may be noted that the design variables of passive device are geometric and inertia parameters of links and springs properties.

5. Optimization and Simulation Results

To simulate the problem, we use the gait pattern of a healthy subject having 74.84 *kg* weight and 1.854*m* height obtained by tracking the trajectories of optical markers attached on bony landmarks of lower legs based on the Modified Helen Hayes (MHH) model [11]. This gait data collection had been done in Neuromuscular Biomechanics Lab in Department of Mechanical Engineering at University of Delaware. For each segment, using time trajectories of fixed three markers which are not on a straight line, tracking of its motion as well as deriving of Euler angles can be performed kinematically. For each individual segment, using these markers, a marker axes system can be defined that is a local frame. Then, the anatomical calibration process is accomplished to acquire the relation between marker axes and anatomical axes. Here, extra calibration markers may be placed temporarily on the segment to specify the anatomical axes. Then, we can easily calculate the constant rotation matrix from anatomical frame to the marker frame. Also, we drive the time varying rotation matrix that rotates the global reference axes to the tracking marker axes. The combination of these two rotation matrices gives us the rotation matrix from global to anatomical frames [10]. However, because of very few easily identified bony landmarks on thigh, not only the derivation of rotation matrix of this segment is difficult, but also the location of hip joint center (HJC) is not known to be utilized in dynamic model. This problem is still a challenge in gait analysis and several strategies have been developed to find the exact location of HJC. In MHH model, regression equations estimate the location of HJC based on location of two ASIS markers and Sacrum (S2) marker and the height of the subject [14, 15]. In this research work, the approach presented by Bell et al. [14] is applied to define HJC as follows:

$$x_h = -0.19 \text{ PW}, y_h = -0.30 \text{ PW}, z_h = 0.36 \text{ PW} \quad (7)$$

where, x_h, y_h and z_h are coordinates of right HJC in pelvic anatomical frame defined in Figure 4. Also, PW is the distance between ASISs as pelvis width. Respectively, the left HJC is defined in symmetry with respect to sagittal plane passing pelvis origin point.

The mass and inertia properties of the segments are derived as stated in [10]. Furthermore, their geometrical properties are calculated using calculation of distance between markers on bony markers of the segments. Also, the speed of treadmill is considered to be 0.6 *m/s* during walking.

We define an optimization problem as follows: To reach optimum values of design parameters so that the actual motion of pelvis becomes close to desired one during gait cycle. Using evaluated design parameters during optimization process, the forward dynamic equations of system are solved numerically and the difference between the actual rotations of pelvis and desired angles are determined subjected to a proper objective function. This function is chosen as:

$$f = \int_0^T \Big[\left| ((\psi(t) - \psi_d(t))/d_\psi)^{n_\psi} \right| + \left| ((\theta(t) - \theta_d(t))/d_\theta)^{n_\theta} \right| + \left| ((\phi(t) - \phi_d(t))/d_\phi)^{n_\phi} \right| \Big] \, dt \quad (8)$$

where, T is period time of gait cycle. ψ, θ and ϕ, the generalized coordinates of dynamic system, representing the actual time trajectories of pelvic rotations, i.e., rotation, obliquity, tilt derived by forward dynamic solution of problem. Also, ψ_d, θ_d and ϕ_d express the desired time trajectories. Using such an objective function results in the actual trajectories to be placed in a tunnel around the desired ones [16]. The parameters n_ψ, n_θ, n_ϕ in the objective function capture the slope of walls of assumed tunnels around the desired trajectories of ψ_d, θ_d, ϕ_d and the parameters d_ψ, d_θ, d_ϕ express the width, respectively. Higher the values of n_ψ, n_θ, n_ϕ, the walls of tunnels become steeper and the widths get closer to values of d_ψ, d_θ, d_ϕ. In this research work, these parameters are chosen as we reach to appropriate optimum pelvic rotations around desired trajectories at the end of the optimization process applying selected bounds of variation of design parameters. As said, the design parameters are bounded based on the practical considerations of fabrication. Also, in this design, by applying suitable limits on the parameters, we prevent physical interface between subject and passive device during motion. Therefore, we deal with a constrained optimization problem subject to restrictions on design variables in addition to dynamic equations of the presented model. It is noted that we allow the device to be adjustable corresponding to optimized variables relevant to any subject with different geometric and inertia parameters. However, this issue will be considered during the fabrication of device performed in near future.

In this work, we present the simulation results of problem for different selected values of d_ψ, d_θ, d_ϕ in objective function of optimization. The vicinity of any pelvis rotation angle trajectory to desired one is related to assumed width of tunnel around it. The more width of each tunnel results in more difference of corresponding actual and desired trajectories. Although our aim is to coincide all three pelvis rotations to corresponding desired ones but it is nearly impossible for our optimization problem considering the constraints of dynamic equations and bounds of design parameters. It means we may not get to the ideal state where, $f = 0$. However, we can vary the shape of graphs of pelvis angles with respect to desired ones and obtain the suitable results by appropriate selection of the parameters of objective function. This is accomplished by considering the frequency amplitude of desired pelvis angles and employing trial and error approach. Also, if we can disregard the importance of proximity of one (or two) of pelvis angles to desired one, by increasing the width of its related assumed tunnel, we obtain more acceptable results for the other angles.

Here, the optimization problem is performed for three groups of values of d_ψ, d_θ, d_ϕ in objective function. First, we consider the best case where we assume equal importance for proximity of all angles to corresponding desired ones. Next, we concentrate our attention on two of pelvis rotations by increasing the width of the tunnel for the least important pelvic rotation in optimization problem. Finally, we increase the proximity of actual and desired trajectories of one of pelvis rotations in contrast to other rotation angles. However, for all simulations, the values of n_ψ, n_θ, n_ϕ stating the slope of walls of assumed tunnels around desired pelvis angles are considered to be one. This selection results in having smooth and uniform actual pelvis rotational trajectories around desired ones at the end of optimization process.

Note that the related optimization problems are solved using an interesting approach by combination of a written program of Genetic Algorithm and *fmincon* function of MATLAB software. The function *fmincon* is utilized to optimizing of general constrained optimization problem and is programmed based on Modified Constrained Steepest Descent Algorithm that uses a potential set strategy as *sequential quadratic programming* (SQP) methods [17]. The simulation results for the above-mentioned cases are presented next.

A) $d_\psi = 0.2\ rad,\ d_\theta = 0.1\ rad,\ d_\phi = 0.1\ rad$

This selection of values of tunnel width which is performed by considering the frequency amplitude of three pelvis rotation angles during gait cycle creates an equality state in vicinity of all of actual trajectories to related desired ones. This case leads to the most proximity of actual movement of pelvis to desired one during walking which is the main purpose of the research. However, we are forced to assume a large interval of variation of design parameters in optimization problem to reach to an acceptable result; which may not be suitable for fabrication of the device. The appropriate lower and upper bounds of design variables of passive device are given in Table 1.

Here, X_{0p}, Y_{0p} and Z_{0p} are the initial coordinates of pelvis COM in reference coordinate system. These values are related to location of global reference frame on heel-strike point of right leg and are calculated using kinematic condition of model as:

$$X_{0p} = -0.1397\ \text{m},\ Y_{0p} = -0.0844\ \text{m},\ Z_{0p} = 0.9607\ \text{m} \quad (9)$$

Table 2 shows the results of optimum design parameters for the case (A).

Also, the proper value of body support weight is $f_g = 842.4808\ N$. As observed, by bounding the variables, feasible design parameters are obtained for fabrication process in future. Also, the simulation results corresponding to optimum device namely pelvic optimum rotations versus time of gait cycle are shown in Figure 5.

As observed in Figure 5, the actual rotations of pelvis are close to desired ones during a gait cycle on treadmill. Actual rotations are created using optimum passive device connected to pelvis of patient without gait ability. An appropriate gait configuration during walking on treadmill can be obtained by applying these results to pelvis.

B) $d_\psi = 0.2\ rad,\ d_\theta = 0.1\ rad,\ d_\phi = 1\ rad$

Next, we increase the width of tunnel around desired tilt angle (ϕ) of pelvis considering its frequency amplitude during gait cycle. In other words, the proximity of rotation and obliquity angles of pelvis to their desired ones are assumed more important than tilt angle. This results in that the actual time trajectory of tilt angle is not obtained very close to desired one. Instead, as expected, the other two pelvis angles (ψ, θ) become very close to corresponding desired ones. However, the trajectory of tilt angle is still acceptable in comparison to desired one. In this case, the interval of design parameters of device can be determined closer than previous case. Consequently, utilizing of related optimum values of parameters facilitates the fabrication process and leads to have a more practical device. Table 3 shows the values of upper and lower bounds of variation of design parameters.

Table 1. The lower and upper bounds of design parameters of passive device

Design parameter	Lower bound	Upper bound
β_i(rad)	$-\pi$	π
k_i(N/m)	0	100
δ_i(m)	0	0.3
X_{oi}(m)	$X_{op} - 2$	$X_{op} + 2$
Y_{oi}(m)	$Y_{op} - 2$	$Y_{op} + 2$
Z_{oi}(m)	0	$Z_{op} + 1.25$
m_i(kg)	0	2
d_{mi}(m)	0.3	2
d_{si}(m)	0.3	2
f_g(N)	0	851.9

Table 2. The optimum design parameters of passive device for case (A)

	i = 1	i = 2	i = 3	i = 4	i = 5	i = 6
β_i(rad)	0.5083	–1.9614	1.8456	–0.9974	–0.5896	3.0085
k_i(N/m)	69.7585	23.3601	23.7139	57.3657	94.0816	31.0047
δ_i(m)	0.0812	0.1278	0.0420	0.0826	0.0626	0.1738
X_{oi}(m)	1.5563	0.5120	–0.5288	0.0665	1.1922	0.4446
Y_{oi}(m)	0.7377	–0.5719	–0.5474	0.3713	–1.0655	1.1847
Z_{oi}(m)	1.7510	0.7193	1.3051	1.4293	0.9317	1.3811
m_i(kg)	1.7807	1.0503	1.7748	1.3120	0.3520	1.7154
d_{mi}(m)	1.8882	1.8691	1.6907	1.4022	0.5209	1.7867
d_{si}(m)	0.8483	1.7313	1.2465	1.6094	1.6416	0.9319

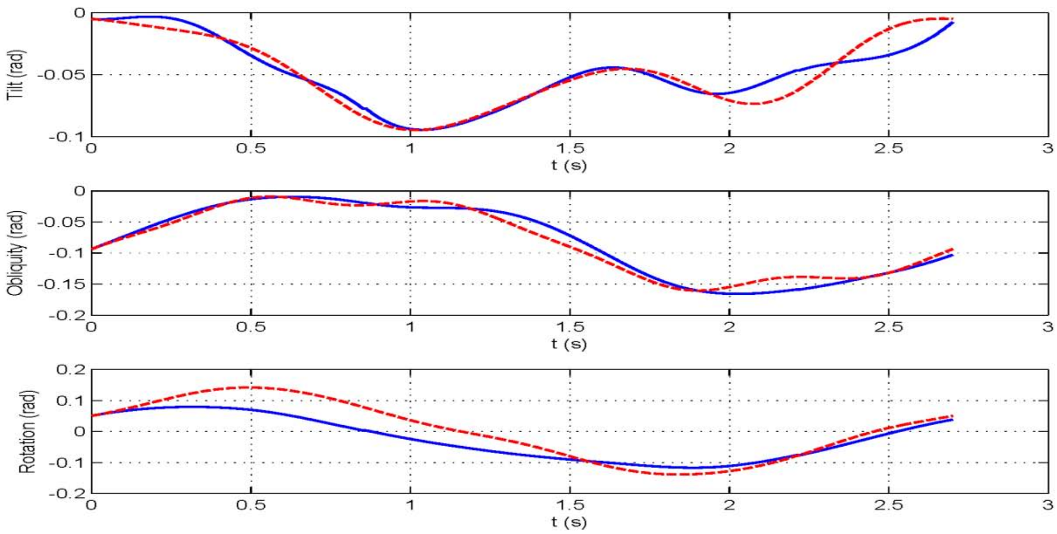

Figure 5. Tilt, Obliquity and Rotation trajectories of pelvis of subject during gait cycle for case (A). Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories.

Table 3. The lower and upper bounds of design parameters of passive device for case (B)

Design parameter	Lower bound	Upper bound
β_i(rad)	$-\pi$	π
k_i(N/m)	0	100
δ_i(m)	0	0.3
X_{oi}(m)	$X_{op} - 1.5$	$X_{op} + 1.5$
Y_{oi}(m)	$Y_{op} - 1.5$	$Y_{op} + 1.5$
Z_{oi}(m)	Z_{op}	$Z_{op} + 1.25$
m_i(kg)	0	1.5
d_{mi}(m)	0.3	1.5
d_{si}(m)	0.3	1.5
f_g(N)	0	851.9

Table 4. The lower and upper bounds of design parameters of passive device for case (C)

Design parameter	Lower bound	Upper bound
β_i(rad)	$-\pi$	π
k_i(N/m)	0	100
δ_i(m)	0	0.3
X_{oi}(m)	$X_{op} - 1$	$X_{op} +1.5$
Y_{oi}(m)	$Y_{op} - 1.5$	$Y_{op} +1.5$
Z_{oi}(m)	Z_{op}	$Z_{op} + 1.25$
m_i(kg)	0	1
d_{mi}(m)	0.3	1
d_{si}(m)	0.3	1
f_g(N)	0	851.9

As shown in Figure 6, optimum time trajectories for rotation and obliquity of pelvis during walking on treadmill are in good nearness to related desired ones.

C) $d_\psi = 0.1\, rad,\ d_\theta = 1\, rad,\ d_\phi = 1\, rad$

Using this selection of tunnel widths, we have focus on rotation angle (ψ) of pelvis more than pelvic tilt (θ) and obliquity (ϕ). Here, we try to get a better optimum result for rotation angle of pelvis in comparison to cases A and B. In this case, more limitation on lower and upper bounds of some of design parameters of device leads to more suitable design parameters for a practical device. The bounds of parameters are specified in Table 4.

As shown in Figure 7, actual time trajectory of rotation is in perfect coincidence to desired one.

In this case, the importance of tilt and obliquity of pelvis movement is assumed less than the rotation. However, the actual time trajectories of tilt and obliquity are not derived very far from related desired ones.

We hope that applying these rotational movements to pelvis of patient by our passive device leads to increasing gait ability.

6. Sensitivity Analysis

Here, to consider the non-structural uncertainties of dynamic model, we study the performance of passive device by varying the design parameters of pelvic orthosis from their optimum values in addition to variation of anthropometry parameters of the subject

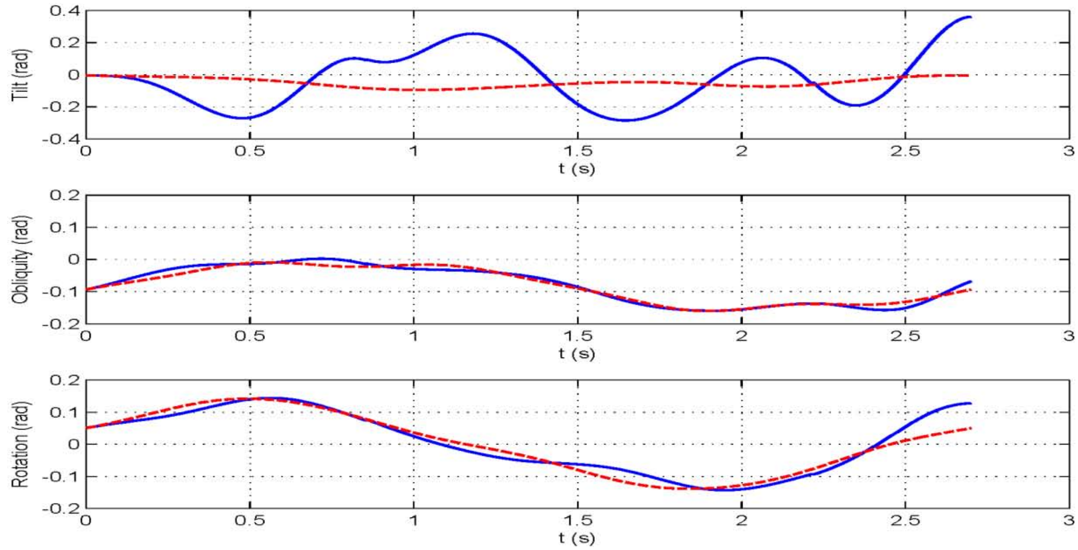

Figure 6. Obliquity and Rotation trajectories of pelvis of subject during gait cycle for case (B).
Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories.

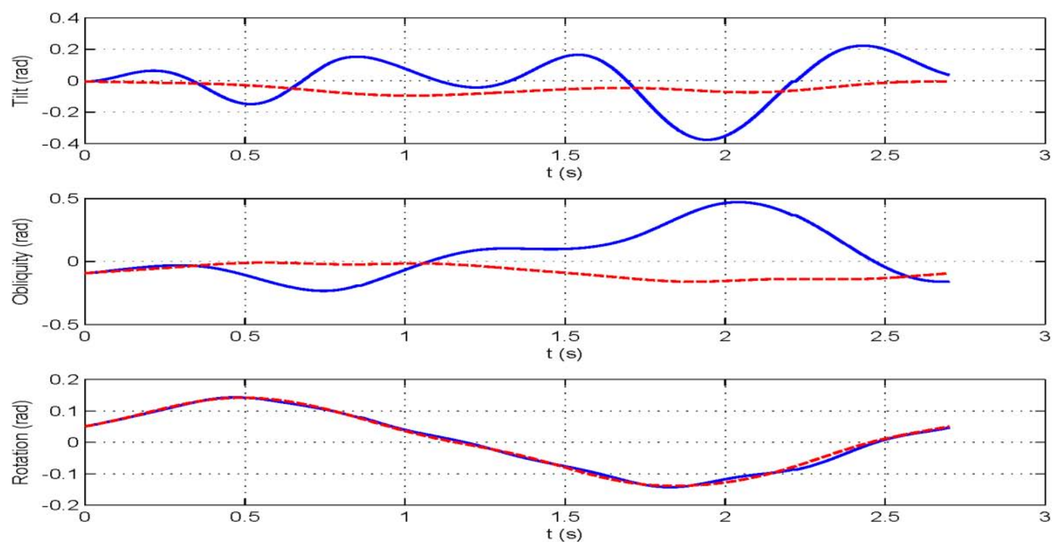

Figure 7. Rotation trajectory of pelvis of subject during gait cycle for case (C).
Solid blue line: Actual Trajectory, Dashed red line: Desired Trajectory.

from calculated and/or predicted values. This problem is more important for the parameters which can not be measured exactly such as the mass of segments of subject. We estimate their values using statistical data and simple formulations given in literature. Therefore, these estimated values may be different from the actual values. We study the effect of variation of parameters of model as well as device, separately. It is noted that since all parameters of device are measurable, the probability of error in these parameters is little in comparison to subject anthropometry ones. However, sensitivity analysis is performed considering simulation results of case (A) stated in previous section with corresponding values of assumed tunnel widths around desired pelvis rotations as $d_\psi = 0.2\ rad$, $d_\theta = 0.1\ rad$, $d_\phi = 0.1\ rad$.

We consider the variation of optimum design parameters of device corresponding to i = 1 (Relevant to the first rod: see Figure 1 and Table 2) by the percentages selected randomly as in Table 5.

Table 5. The percentage of design parameters variations

	β_1	κ_1	δ_1	x_{01}	Y_{01}	Z_{01}	m_1	d_{m_1}	d_{s_1}
percentage of variation	3	2	2	3	2	2	1	3	2

Table 6. Tthe percentage of subject parameters variations

	m_{foot}	m_{shank}	m_{thigh}	m_{pelvis}	l_{foot}	l_{shank}	l_{thigh}	h_{pelvis}
percentage of variation	4	4	5	6	3	3	3	5

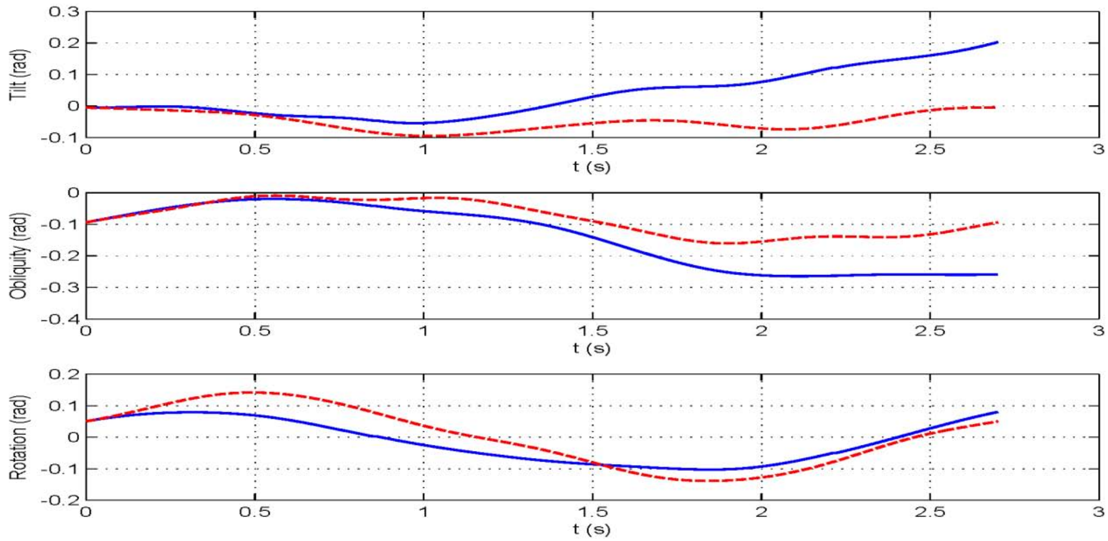

Figure 8. Tilt, Obliquity and Rotation trajectories of pelvis of subject during gait cycle by variation of design parameters. Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories.

Figure 8 shows the deviance of actual pelvic angles with respect to desired ones due to variation of the design parameters from their optimum values. As shown, although we change the optimum values of design variables by 1-3 percent, considering the small amplitude of periodic rotational movements of pelvis, we still get rather good performance of device particularly about rotation angle.

Also, we study the effect of device on optimum pelvic rotations by varying the determined values of some geometrical and inertia parameters of subject employing the orthosis during gait on treadmill. Table 6 shows the random percentages of variation of anthropometry parameters of the valid user.

Where, m, l and h stand for mass, length and height, respectively. The deviance of actual pelvic rotations with respect to desired ones due to variation of the determined parameters of subject is presented in Figure 9.

As observed, we still have a good performance of device by changing the several subject parameters consisting of mass and length of lower legs. The results of sensitivity analysis can validate the proposed passive design in assisting pelvis motion before fabrication process of a prototype.

7. Conclusion

In this research, we studied the design of an assistive passive device to help the subject having gait impairments to move his/her pelvis in the vicinity of its desired motion during gait on treadmill. The intent is to create pelvis rotations in assumed tunnels around the desired trajectories without using any actuators. Using an optimization algorithm and iterative solution of forward dynamic equations, we achieved the optimum design parameters of passive device. We considered a healthy subject with given sampled gait pattern and studied his anthropometry and calculated 3D motion of lower legs to apply for simulation of dynamic problem. In this work, we assumed the prescribed motion for the thighs, shanks and feet of user which can be provided by cooperation of therapists or a robotic gait orthosis.

The proposed design is limited to lower and upper bounds of parameters to have practical feasibility for the fabrication and to be prevented from any interface

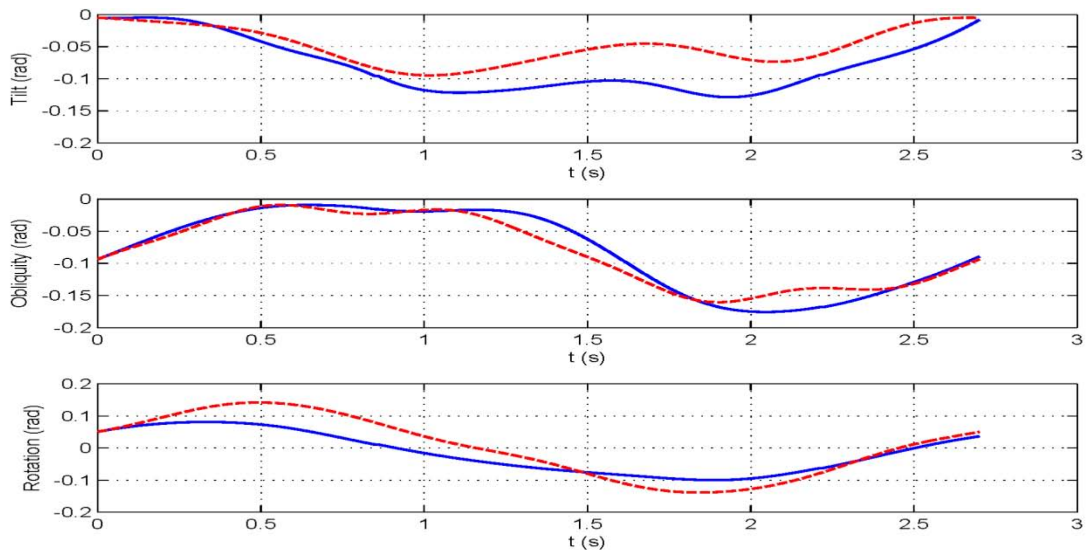

Figure 9. Tilt, Obliquity and Rotation trajectories of pelvis of subject during gait cycle by variation of his anthropometry parameters.

Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories.

between user and device during gait on treadmill. However, we allowed the design of device to be adjustable for any subjects with different geometric and inertia parameters.

The simulation results are presented in three cases considering selected values for the width of assumed tunnels around desired pelvis angles. We concentrate our attention to proximity of three, two and one of pelvis angles to corresponding desired ones in cases of (A), (B) and (C), respectively. The most nearness of actual and desired angles of pelvis is occurred in case (C); also, we have obtained the most feasible design parameters of device in this case (attending to fabricating limitations). Furthermore, we studied the issue of sensitivity analysis of passive device by varying the design by 1-5 percent around the optimum parameters in addition to variation in geometry and inertia parameters of the subject. As observed, the results are still appropriate to use in training a patient to recover his/her normal pelvis motion during ambulation. This can verify the performance of device before carrying out the experiment on device performance. However, we will validate our studies by fabrication of a prototype of device with optimum mechanical parameters and present the experimental results in near future.

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