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A Novel Passive Pelvic Device for Assistance during Locomotion

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***Abstract*—A large number of people suffer from impairments, such as injury to joints or the spinal cord, that limit motion of the pelvis. This motion plays an important role in balance and propulsion during a gait cycle. In this work, we present a method to design a passive device that assists the pelvis to move close to a reference trajectory during walking with partial body support. This device is un-motorized and contains only passive elements.**

In this paper, we model subjects with different ability levels and body weight support and determine optimum design parameters for the device. The simulation results show the configuration of the optimum device and time trajectories of pelvic rotations in comparison to the reference.

I. Introduction

Pelvis plays a central role in gait. It transfers forces from the lower extremity to the trunk and helps in forward propulsion during ambulation [1]. Pelvis also plays a role in placing the center of mass (COM) of the trunk above the support leg to enable swing of the other leg [2], [3]. It also modulates the vertical motion of the COM of the body to reduce energy consumption during walking [2]. Hence, attention to pelvic motion is essential in assistance and/or rehabilitation of gait.

These motivate design of suitable devices that can assist in correct movement during walking. Development of pelvic orthoses is a relatively new topic in the field of rehabilitation, particularly if it is passive. Surdilovic and Bernhardt presented a robotic prototype, STRING-MAN, for advanced gait rehabilitation. This system is a wired robot for gait rehabilitation and restoration of motor functions by combining the advantages of partial body-weight bearing with a number of robotic and humanoid control functions [4]. However, STRING-MAN improves the human gait by control of zero-moment-point (ZMP) of the body but does not control the motion of pelvis explicitly. 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 [5]. Matjacic presented a passive apparatus for dynamic balance training during treadmill walking, which was tested on a subject with incomplete spinal cord injury. This device connects to the pelvis and facilitates dynamic balancing and training during treadmill walking [6]. Stauffer, et al. introduced a new device for paraplegics, the WalkTrainer. This device is composed of a leg and pelvic orthosis, an active bodyweight support and closed loop muscle stimulation. Pelvic trajectories are measured and programmed on the WalkTrainer and applied to healthy subjects by mean of the pelvic orthosis [7].

In this paper, we present a method to design a passive 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 by exerting forces on it during the gait. *To the best knowledge of the authors, passive devices that improve the motion of pelvis during walking have not been explored*. Furthermore, a passive device can be safer and economical, compared to an active device. In this work, we match the motion of the pelvis to a desired motion by optimization of design parameters in the device. This philosophy is similar to that used in the design of swing-assist exoskeleton [8].

The organization of this paper is as follows: In Section II, we summarize the mechanical model of the device with passive elements. In Section III, the dynamic model of pelvis and its interaction with the leg and upper body are discussed. We present the dynamic equations using Newton-Euler approach in Section IV. Next, we outline the optimization problem and present simulation results in Section V. Finally, discussion of results and conclusion are presented.

II. System Description

Our goal is to design a passive device, connected to the pelvis of a user during treadmill walking. A set of springs and masses are assumed to be connected to the pelvis, as shown in Fig. 1. The rods in the device are considered mass less. Each spring is connected at one end to the rod and the other end is fixed in the inertial frame, say to a walker around the treadmill [9]. Also, in this design, we expect to use a body support to unload a portion of subject's weight during walking.

To model the proposed design, we consider a reference coordinate system *XYZ* in the inertial frame and a body local coordinate system *xyz* attached to the pelvis. The origin of the local frame is located at the COM of pelvis. The X axis is perpendicular to the direction of the treadmill motion and the Z axis is vertically upwards, against the direction of gravity, as shown in Fig. 1.

We assume that the pelvis COM follows the prescribed motion but its orientation needs to be assisted during the motion.

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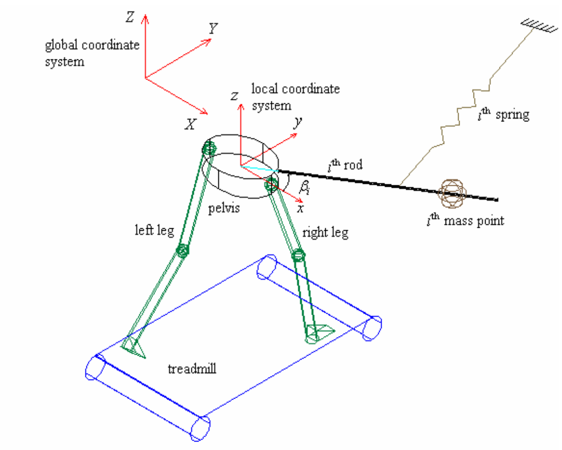

Fig. 1. The 3D schematic of the passive device connected to pelvis

In our problem formulation, the design parameters are listed below for i = 1…n, where n is the number of rods:

β_i : Angle between the x axis and i^{th} rod fixed to pelvis.

$x_{o_i}, y_{o_i}, z_{o_i}$: Coordinates of the fixed point of i^{th} spring in the inertia coordinate frame.

f_g : Vertical body support weight.

k_i : Stiffness constant of i^{th} spring.

δ_i : Initial stretch of i^{th} spring.

m_i : Mass of i^{th} point mass.

d_{m_i} : Position of i^{th} point mass on the i^{th} rod.

d_{s_i} : Location of fixed point of i^{th} spring on the i^{th} rod.

In total, we have 9n+1 design parameters in this optimization problem which will have specified lower and upper bounds. Considering that the design is passive and has limited design parameters, a reasonable question to ask is how many springs 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 n < 6. This results in 55 design parameters, which need to be optimized so that the pelvis follows a prescribed orientation time history during gait on the treadmill.

III. Dynamic Model

In this work, we present a proper dynamic model which has an acceptable consistency with the human body motion. As an example, the structure of skeleton of pelvis and lower extremities is shown in Fig. 2. We try to have a real model of pelvis, swing and stance legs.

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.

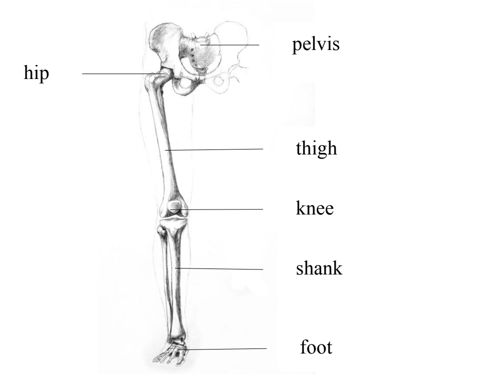

Fig. 2. The skeleton of pelvis and lower extremities

As one would expect, the motion of the origin of pelvis frame depends on the posture of the stance leg and pelvis angles. With our assumption, the pelvis COM follows a prescribed motion, while the orientation of pelvis is assisted by the passive assist device. In the model, we assume the pelvis to be a cylinder with ellipsoid section and a uniform mass distribution. We model the effect of upper body and body support as a vertical force on COM of pelvis.

As shown in Fig. 3, in dynamics model, the pelvis is assumed to be connected to leg at hip with a spherical joint corresponding to its three angular movements i.e., rotation, obliquity and tilt during walking. Furthermore, we consider the swing leg is attached to pelvis through the other hip using a universal joint such that the hip has two DOFs related to flexion/extension and abduction/adduction of swing leg.

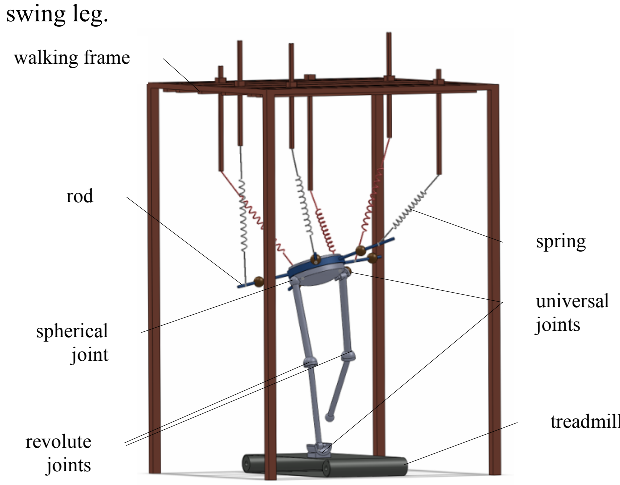

Fig. 3. Dynamic model of legs and pelvis connected to passive device during walking on treadmill

Also, we consider the foot as a mass point at the end of the shank of the leg. Each leg is considered as two rigid links that represent the thigh and shank. These links on the stance leg support the pelvis. Moreover, the contact between the stance leg and the treadmill is assumed to be a universal joint. The knees are modeled by revolute joints. We consider that the swing leg is attached to the pelvis at the other hip and has a structure similar to the stance leg.

During gait, at least one foot is assumed to be in contact with the treadmill and the reaction force is applied to the stance leg during gait. This results in switching in the dynamic equations when the support leg changes during the gait cycle. Therefore, the optimization needs to switch the dynamic equations with change in the support leg.

IV. Dynamic Equations

In this work, Newton-Euler method is used to derive the equations of motion of the system. For design, we describe the motion of pelvis for given design parameters. The generalized coordinates are the angular orientations of the pelvis, i.e., rotation (ψ), obliquity (θ) and tilt (ϕ). Rotation of the pelvis is about Z axis of global coordinate system in vertical direction, obliquity is the rotation of pelvis about y axis of current coordinate system and tilt is rotation of pelvis about x axis of current frame [10]. So, the rotation matrix of the current coordinate system with respect to reference coordinate system is given by:

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

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:

$$[X, Y, Z]^T = \mathbf{R}[x, y, z]^T \tag{2}$$

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

$$\boldsymbol{\omega} = \mathrm{vect}(\mathbf{R}^T\dot{\mathbf{R}}) \tag{3}$$

Using the time derivative of (3), one can obtain the angular acceleration of pelvis. In this model, we consider the prescribed trajectories for flexion/extension angle of knee and hip joints and hip abduction/adduction angle of swing leg during gait. Therefore, assuming desired time trajectories for swing leg and origin of coordinate frame on pelvis, the system has only three DOFs of pelvis rotations. To derive the dynamic equations for the model, first, the kinematic consistency between the motion of pelvis and stance leg must be ensured. The joint angles of the stance leg are dependent on rotational and translational movements of pelvis. Thus, we express the joint angles of stance leg corresponding to knee and hip flexion/extension as well as hip abduction/adduction in terms of three generalized coordinates of system i.e., ψ, θ, ϕ and prescribed motion $X_p(t)$, $Y_p(t)$ and $Z_p(t)$ of pelvis COM. We are able to express the angular velocity and acceleration of each rigid body of the model as well as COM acceleration in terms of pelvis rotations and motion of pelvis COM and their time derivatives. Furthermore, in this model, we consider sufficient number of torques in each joint to provide the movements of pelvis, swing and stance legs.

The dynamic model of the system consists of rigid bodies including the pelvis with the device, links of thigh and shanks of swing and stance legs. In Newton-Euler approach, we separate these rigid bodies and draw 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

$$\sum \vec{\mathbf{F}} = m\vec{\mathbf{a}}_G = m(\ddot{X}_G\hat{\mathbf{i}}_0 + \ddot{Y}_G\hat{\mathbf{j}}_0 + \ddot{Z}_G\hat{\mathbf{k}}_0) \tag{4}$$

where $\hat{\mathbf{i}}_0$, $\hat{\mathbf{j}}_0$, $\hat{\mathbf{k}}_0$ are unit vectors of reference coordinate system. Also, X_G, Y_G and Z_G are coordinates of COM of body in global frame. $\sum \vec{\mathbf{F}}$ are the external forces on the rigid body consisting of joint reaction forces, link weight, spring forces, and body support.

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

$$\sum \vec{\mathbf{M}}_G = \dot{\vec{\mathbf{H}}}_G = \\ = (\dot{H}_x - H_y\omega_z + H_z\omega_y)\hat{\mathbf{i}} + (\dot{H}_y - H_z\omega_x + H_x\omega_z)\hat{\mathbf{j}} + (\dot{H}_z - H_x\omega_y + H_y\omega_x)\hat{\mathbf{k}} \tag{5}$$

where $\vec{\mathbf{H}}_G$ 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 $\vec{\mathbf{H}}_G$ along axes of local coordinate frame are expressed as

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

$\sum \vec{\mathbf{M}}_G$ 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 pelvis origin as well as their time derivatives. It may be noted that the design variables of passive device which should be optimized appear in mass, length parameters and spring forces in dynamic equations.

In this work, we intend to consider the interaction between subject and device during gait via consideration of level of ability of patient to move his/her pelvis. To study this issue, we consider that the hip joint torques of stance leg in our dynamic model creates the rotational movements of pelvis during walking for a healthy subject. If we consider the desired motion of pelvis using the dynamic model, we can obtain the perfect values of hip joints torques by solving the inverse dynamic equations for a healthy subject without using assistive device. Therefore, we can define the level of ability of any patient to move pelvis based on the percentage value of desired hip joint torques which he/she can provide during walking. So, we can design the passive device considering the level of interaction between user and device. We try to present a passive device which compensates the lack of hip joint torques of patient by exerting proper forces to pelvis.

Using three algebraic equations of motion i.e., Newton equations and three Euler equations, we solve forward dynamic equations during optimization process by numerical approach using MATLAB software.

V. Optimization and Simulation Results

We consider the physical and geometrical properties of a healthy subject having 70 *kg* weight and 1.74 *m* height for simulation. 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:

$$f=\int_0^T\left[((\psi(t)-\psi_d(t))/d_\psi)^{n_\psi}+((\theta(t)-\theta_d(t))/d_\theta)^{n_\theta}+((\phi(t)-\phi_d(t))/d_\phi)^{n_\phi}\right]dt \quad (7)$$

where ψ, θ, ϕ, the generalized coordinates of dynamic system, representing the actual time trajectories of pelvic rotations, i.e., rotation, obliquity, tilt movements are derived by forward dynamic solution of problem. Also, ψ_d, θ_d, ϕ_d express the desired values. Using such an objective function results in the actual trajectories to be placed in a tunnel around the desired ones [11]. However, we may not get to the best configuration where $f=0$. 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 problem, these parameters are chosen to have appropriate tunnels around desired trajectories. These parameters are:

$$\begin{aligned}&n_\psi=6, n_\theta=6, n_\phi=6\\&d_\psi=0.01\,rad, d_\theta=0.005\,rad, d_\phi=0.005\,rad\end{aligned} \quad (8)$$

At the end of the optimization process, we have the passive motion of pelvis close to the desired motion as well as optimum design parameters. As said, the design parameters are bounded during optimization process. However, the limits on parameters should be chosen from 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 an optimization problem subject to restrictions on design variables in addition to dynamic equations. The appropriate lower and upper bounds of design variables of passive device are given in Table 1.

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

Design parameters	Lower bounds	Upper bounds
β_i (rad)	$-\pi$	π
x_{o_i} (m)	0.166	1.50
y_{o_i} (m)	0.166	1.50
z_{o_i} (m)	-0.8	1.50
k_i (N/m)	0	150
δ_i (m)	-0.35	0.35
m_i (kg)	0	1.5
d_{m_i} (m)	0.166	1.00
d_{s_i} (m)	0.166	1.00
$f_g(N)$	0	686.7

Here, we present the simulation results for two cases. First, we consider the interaction of subject and device assuming several levels of pelvic motion ability for patient. Next, we present optimum passive device for a patient without any ability in pelvic motion. It is noted that we allow the device to be adjustable corresponding to optimized variables relevant to subject with different geometric and inertia parameters. . However, this issue will be considered during the fabrication of device performed in near future.

A. *Subject with given pelvic motion ability*

In this case, we assume that the patient can provide a part of desired hip joint torques to move the pelvis during gait cycle. The simulation results from the optimization technique are shown for three angular orientations of the pelvis during the whole gait in Fig. 4-7 corresponding to several levels of pelvic motion ability. The actual trajectories are shown by solid blue lines as compared to the desired ones, dashed red lines. It may be noted that the desired trajectories are obtained using third model of prediction of the human pelvic motion for considered healthy subject as stated in [7].

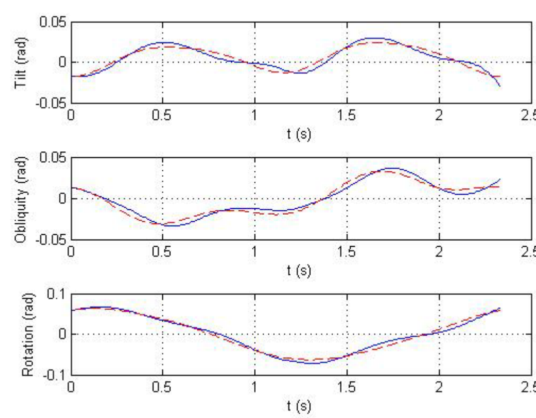

Fig. 4. Tilt, Obliquity and Rotation trajectories of pelvis during gait cycle for a subject with 30% of pelvic motion ability. Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories. Body support weight is obtained to be 76% of subject's weight.

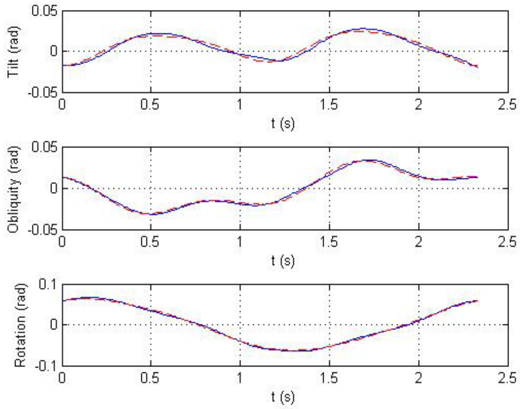

Fig. 5. Tilt, Obliquity and Rotation trajectories of pelvis during gait cycle for a subject with 60% of pelvic motion ability. Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories. Body support weight is obtained to be 52% of subject's weight.

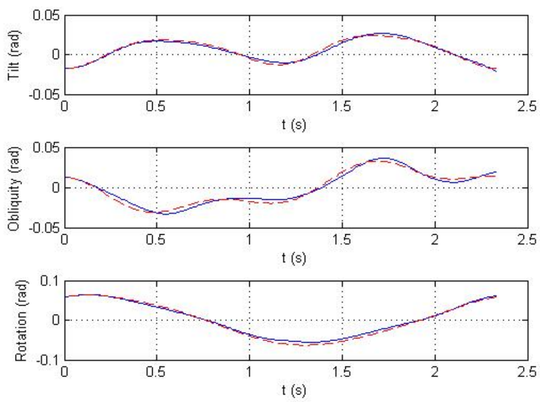

Fig. 6. Tilt, Obliquity and Rotation trajectories of pelvis during gait cycle for a subject with 90% of pelvic motion ability. Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories. Body support weight is obtained to be 14% of subject's weight.

In all above designs, the body support weight is considered as a design parameter derived during optimization process. However, if we intend to train the patient with a specified body support that unloads a proper value of human body weight, we should try to solve this problem by eliminating the parameter f_g from design. It means that we fix the value of body support weight during optimization. We present the results related to this condition of design for patient with 60% of pelvic motion ability. Here, we fix the body support weight to 30% of body weight and obtain the optimum passive device parameters. The actual rotation angles of pelvis are obtained in near to desired ones as shown in Fig. 7.

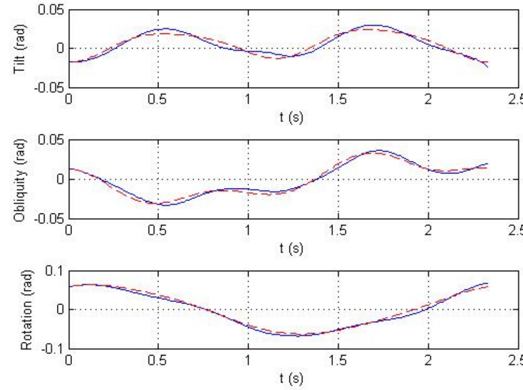

Fig. 7. Tilt, Obliquity and Rotation trajectories of pelvis during gait cycle for a subject with 60% of pelvic motion ability and 30% body support weight. Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories.

B. *Subject without any interaction with device*

Next, we present the passive device for a subject who has no ability to move the pelvis during gait. Table 2 shows the obtained values of design parameters after optimization.

TABLE 2
THE OPTIMUM DESIGN PARAMETERS OF PASSIVE DEVICE FOR A SUBJECT WITHOUT ANY INTERACTION WITH DEVICE

	i=1	i=2	i=3	i=4	i=5	i=6
β_i (rad)	-1.22	-0.95	2.26	-1.73	1.08	-2.85
x_{o_i} (m)	0.35	0.51	-0.68	-0.14	1.36	-1.04
y_{o_i} (m)	-0.94	-0.63	0.67	-0.38	0.62	-0.27
z_{o_i} (m)	0.05	0.05	0.38	-0.01	0.25	0.06
k_i (N/m)	110.96	120.63	110.46	143.44	122.90	150.00
δ_i (m)	-0.32	0.35	0.35	0.35	0.16	-0.10
m_i (kg)	1.50	1.50	0.01	1.50	1.22	1.16
d_{m_i} (m)	1.00	1.00	0.35	1.00	1.00	1.00
d_{s_i} (m)	0.98	0.80	0.91	0.36	1.00	0.97

Also, the proper value of body support weight is 98% of human body weight, namely f_g=675.6 *N*. The simulation results corresponding to optimum device is shown in Fig. 8 as pelvic rotations during walking.

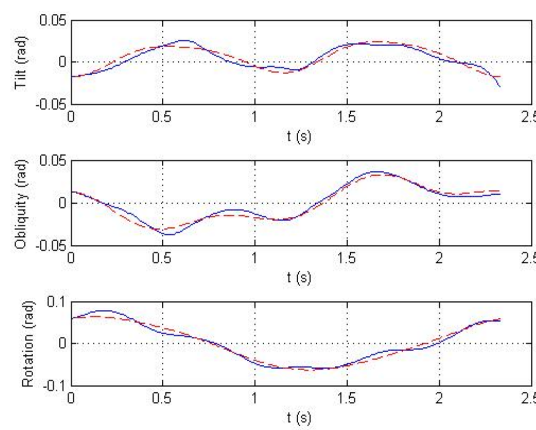

Fig. 8. Tilt, Obliquity and Rotation trajectories of pelvis during gait cycle for a subject without any interaction with device. Solid blue lines: Actual Trajectories, Dashed red lines: Desired Trajectories.

As observed in Fig. 4-8, the actual rotations of pelvis are very close to desired ones, particularly, in states where the subject has more pelvic motion ability. Also, by bounding the variables, feasible design parameters are obtained for fabrication process.

C. Sensitivity analysis

We study the performance of passive device by varying the design parameters of pelvic orthosis from their optimum values by the percentage presented in Table 3.

TABLE 3
THE PERCENTAGE OF DESIGN PARAMETERS VARIATIONS

	β_1	x_{o_1}	y_{o_1}	z_{o_1}	k_1	δ_1	m_1	d_{m_1}	d_{s_1}
percentage of variation	2	3	1	5	4	2	1	3	2

Fig. 9 shows the deviance of actual pelvic angles with respect to desired ones due to variation of the design parameters from their optimum values.

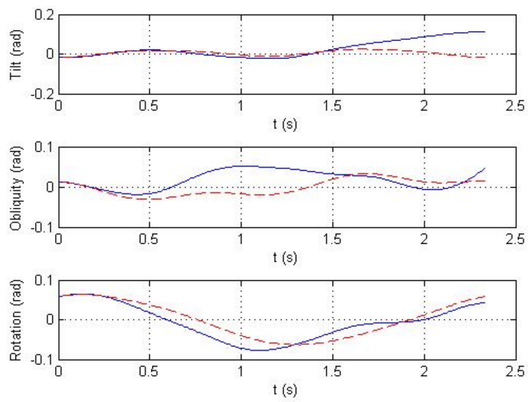

Fig. 9. Tilt, Obliquity and Rotation trajectories of pelvis during gait cycle for a subject with 60% of pelvic motion ability. Solid blue lines: Actual Trajectories,, Dashed red lines: Desired Trajectories.

As shown, although we change the optimum values of design variables by 1-5 percent, we still get rather good performance of device particularly about rotation angle. We have also observed the same result by changing the subject parameters. To avoid repetitive, we do not include it in the paper.

VI. CONCLUSIONS

In this research, we studied the design of an assistive passive device to help a subject with 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 a tunnel 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 the user to have different levels of ability in pelvic motion. Also, we presented an appropriate optimum design for a fixed body support weight. However, we allow the design of the device to be adjustable for subject with different geometric and inertia parameters as well as fixed body support weight. We will account for these issues practically during fabrication in near future. 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. In this work, we considered prescribed motion for the COM of pelvis. However, we will relax these assumptions in the future.

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