Gravity Balancing of a Human Leg Using an External Orthosis

Gravity balancing is often used in industrial machines to decrease the required actuator efforts during motion. In this paper, we present a new design for gravity balancing of the human leg using an external orthosis. This external orthosis is connected to the human leg on the shank and its other end is fixed to a walking frame. The major issues addressed in this paper are as follows: (i) design for gravity balancing of the human leg and the orthosis, (ii) kinematic compatibility of the human leg and the external orthosis during walking, and (iii) comparison of the joint torque trajectories of the human leg with and without external orthosis. We illustrate feasible 2D and 3D designs of the external orthosis through computer simulations. Our results show that the 3D design has smaller inertia with respect to 2D design, which can be more helpful for typical stroke patients to walk in a balanced position. [DOI: 10.1115/1.4003329]

1 Introduction

In recent years, passive gravity balancing orthoses have been proposed for the upper arm [1–3]. For people suffering from neuromuscular diseases, there is a need for a device to support the weight of their arm against gravity and allow either their residual muscle force or external power to position the arm in space. Many of the current assistive devices are externally powered, tend to be very bulky, and have poor aesthetics. The nonpowered devices of the current assistive devices are externally powered, tend to be bulky, and have poor aesthetics. The nonpowered devices available suffer from limited range of motion and several other inconveniences [3]. Moreover, orthoses for the lower extremity are typically powered. For example, a single degree-of-freedom (DOF) powered gait orthosis using a direct dc motor was designed to provide bipedal locomotion [4]. In 2005, Aoyagi et al. developed a robotic device (pelvic assist manipulator (PAM)) that assists the pelvic motion during gait training on a treadmill. PAM allows naturalistic motion of pelvis actuated by six pneumatic cylinders, which combined with a nonlinear force-tracking controller [5]. Lokomat is an actively powered exoskeleton worn by patients during treadmill walking [6]. Mechanized gait trainer (MGT) is a single DOF powered machine that drives the leg through a gait pattern [7]. The disadvantage of these devices is that they force the patient to use a predetermined movement pattern rather than allowing the patients to move under their own control. Banala et al. [8] and Agrawal and Fattah [9] proposed a design of an exoskeleton for full or partial gravity balancing of a human leg during motion. The device is worn by the user and segments of the exoskeleton are strapped to the corresponding segments of the human leg. However, there are some issues with the existing exoskeletons [6,7,10,11], which motivate us to look at alternative designs. One such issue is the alignment of the human leg and exoskeleton segments. Also, it is hard to get full extension of the knee due to singular configuration of the exoskeleton. Robotic devices are developed to assist patients with lower extremity using alternate designs [5,12,13]. Galvez et al. [12] proposed a sensorized orthosis that measured shank kinematics and therapist forces during locomotor training. The orthosis is attached to one of the legs. Sirdulovic and Bernhardt [13] developed the String-Man, a tension controlled wire-drive system, which stabilizes the torso of a subject during stepping on a treadmill.

In this paper, we present a new design for gravity balancing of the human leg using an external orthosis. The key contribution of this paper is the design of an external orthosis to avoid issues with the existing exoskeletons, such as joint and segment misalignment. This orthosis is designed for 4DOF motion of the human leg in swing phase, namely, 2DOF motion in the sagittal plane, i.e., flexion and extension at the hip and knee, 1DOF for hip abduction/adduction in the coronal plane, and 1DOF for hip rotation in the transverse plane (see Fig. 1). The foot is considered as a point mass at the end of the shank segment. The external orthosis connected at the shank, together with the human leg, creates a kinematic closed loop.

The kinematic loop constraint can be satisfied during walking by choosing appropriate link lengths for the external orthosis. Gravity balancing of the human leg and the external orthosis is achieved by making the potential energy of the combined system, human and the external orthosis, to be configuration invariant. First, the potential energy of the system is written in terms of the joint angles of the human leg and the external orthosis. Loop constraint equations are then substituted in the potential energy to express dependent joint angles in terms of independent ones. Finally, the coefficients of joint angle dependent terms in this expression are made to vanish to make the potential energy invariant and thereby achieve a gravity balanced system. These conditions are satisfied by choosing appropriate inertia parameters of the segments of the orthosis and addition of proper springs. The main advantages of the design of human leg with external orthosis as compared with the existing exoskeletons are as follows. (i) This design does not have the problem of alignment between the human leg and the orthosis because there is only one connection point between the external orthosis and human leg. However, in existing exoskeleton orthosis, the human leg and exoskeleton should be aligned at hip and knee joints. In other words, the axes through the centroid of knee and hip joints of the orthosis and the corresponding ones in human leg should be collinear. Since exoskeleton and human leg are considered as a single device, therefore, the joints at hip and knee should be completely aligned. (ii) We can also get the full extension of the knee with this design. (iii) It requires less hardware such as force-torque sensors between human leg and orthosis to compute the joint torques since there is only one connection point between the external orthosis and human leg. Nevertheless, there are also some drawbacks in this design such as the following. (a) It increases the inertia of the system during fast walking. (b) Modeling of human leg with the
external orthosis, having a kinematic closed loop, is more complicated than modeling of existing open-loop exoskeletons.

The organization of this paper is as follows. Section 2 describes the kinematic compatibility of the human leg and the external orthosis during walking. Gravity balancing of the human leg and the external orthosis is described in Sec. 3. Some feasible designs are then presented in Sec. 4. Joint torque computation is studied in Sec. 5.

2 Kinematic Compatibility

Figure 2 shows a schematic of 4DOFs of external orthosis and human leg in a walking frame. The walking frame is connected to orthosis. In fact, both walking frame and orthosis will help the patient to walk in a balanced position.

Our model of human leg has 2DOFs for the hip medial and lateral rotation and hip abduction at point $O_1$, i.e., $\theta_0$ and $\theta_1$, respectively, and 2DOFs for the flexion/extension at points $O_2$ and $O_3$, namely, $\theta_2$ and $\theta_3$, as shown in Fig. 3. Also, the external orthosis has 2DOFs at point $O_4$, i.e., $\theta_4$ and $\theta_5$, respectively, and 2DOFs at points $O_6$ and $O_7$, i.e., $\theta_6$ and $\theta_7$. The human leg is connected to the external orthosis through a universal joint at point $O_8$. The human leg is connected to the walking frame at point $O_9$. The vertical motion of the hip is quite smaller than the vertical motion of the other joints of the leg and thus it can be easily neglected. Hence, it is assumed that the human leg is also connected to walking frame at point $O_9$.

The distance between leg and external orthosis is shown as $d_0$ in Fig. 3. Using the Kutzbach–Gruebler formula for this system, it is easy to show that this system creates a kinematic closed-loop mechanism comprising external orthosis and human leg to have the same DOF as the human leg, we connect the two systems by a universal joint instead of a spherical joint to simplify the kinematic loop constraint equations. Regarding Fig. 3, the loop constraint equation for the system is written as

$$O_1O_2 + O_2O_3 + O_3O_4 + O_4O_5 + O_5O_6 + O_6O_7 + O_7O_8 + O_8O_1 = 0$$

This equation can be written in terms of its components along the axes of inertial frame (see Appendix A for detailed computation), namely, $X$, $Y$, and $Z$ as

$$l_0s_\theta + l_1c_\theta + l_1s_\phi s_\theta + (l_2 - l'_1)c_\phi c_\theta + (l_2 - l'_3)s_\phi s_\theta s_\phi - l'_2c_\phi c_\theta - l'_3c_\phi c_\theta c_\phi - d_0 = 0$$

(2)

$$-l_0s_\phi + l_1c_\phi s_\phi + l_2c_\phi - l'_1c_\phi c_\phi + l'_2c_\phi c_\theta c_\phi - l'_3c_\phi c_\theta c_\phi c_\phi = 0$$

(3)

$$l_0c_\phi + l_1s_\phi c_\phi + l_1c_\phi s_\phi - (l_2 - l'_1)s_\phi c_\phi + (l_2 - l'_3)c_\phi s_\phi - l'_2c_\phi - l'_3c_\phi c_\phi = 0$$

(4)

where $s_\phi$ and $c_\phi$ stand for $\sin \theta_\phi$ and $\cos \theta_\phi$, respectively. Here, $l_0$ is the offset of the hip abduction and hip flexion joint axes and $l_1$ and $l_2$ are link lengths of the thigh and the shank segments of the human leg, as depicted in Fig. 4. Also, $l'_1$, $l'_2$, and $l'_3$ are the link lengths of external orthosis. Furthermore, $l'_3$ locates the distance between $O_4$ and $O_5$, as shown in Fig. 3.

Equations (2)–(4) are three nonlinear equations to be solved numerically to express dependent joint angles $[\theta_0, \theta_1, \theta_6, \theta_7]$ in terms of independent ones $[\theta_2, \theta_3, \theta_4, \theta_5]$. The link lengths of external orthosis are chosen such that above equations are satisfied during normal walking. Assuming predetermined length for exter-
nal orthosis links \( (l'_i; i=0,\ldots,3) \), one can solve the nonlinear equations (2)--(4) to express dependent joint angles in terms of independent ones using a nonlinear equation solver in MATLAB like fsolve. If all of these equations are satisfied during a complete normal walking cycle, then selected lengths for orthosis limbs are acceptable.

3 Gravity Balancing of the Human Leg and External Orthosis

The objective is to design an external orthosis with appropriate geometry, inertia, and springs such that the combined system of human leg and the external orthosis becomes gravity balanced. To this end, it is tried to make the total potential energy of the system comprises of gravity and spring configuration invariant. The potential energy of the human leg and external orthosis due to the gravity can be written as

\[
V_g = - \sum_{i=1}^{3} m_l \mathbf{g} \cdot \mathbf{r}_{oc_i} - \sum_{i=1}^{4} m'_l \mathbf{g} \cdot \mathbf{r}_{oc'_i} \tag{5}
\]

where \( m_l \) is the mass of link \( i \) of the human leg and \( \mathbf{r}_{oc_i} \) is the location of center of mass (COM) of link \( i \) from its origin. Also, \( m'_l \) is the mass of link \( i \) of the external orthosis and \( \mathbf{r}_{oc'_i} \) is the location of center of mass of link \( i \) from its origin. These expressions are given in Appendix B.

Upon substitution of \( \mathbf{r}_{oc'_i} \) and \( \mathbf{g} = -g\mathbf{j} \) into Eq. (5), the total gravitational potential energy \( V_g \) can be written in terms of joint angles as

\[
V_g = m_0 g (-l_0 s_1) + m_1 g (-l_1 s_1 + l_1 c s_2) + m_2 g (-l_2 s_1 + l_2 c s_3) + m_3 g (-l'_3 s_1 + l'_3 c s_2) \\
+ l_2 c s_3 + m_3 g (-l_3 s_1 + l_3 c s_2 + (l_3 - l'_3) c s_3) + m_3 g (-l'_3 s_1 + l'_3 c s_2) \\
+ l'_1 c s_5 + l'_2 s c_6 + m'_1 g (-l'_5 s_8 + l'_1 c s_7) + m'_2 g (-l'_5 s_8) \tag{6}
\]

where \( l_0 = O_1 C_0, l_1 = O_2 C_1, l_2 = O_3 C_2, l_3 = O_4 C_3, l'_3 = O_0 C'_3, l'_1 = O_7 C'_1, \) and \( l'_5 = O_9 C'_1 \) are locations of COM of the links from their origins. Also, \( d_0 = O_0 O_1 \) is a constant distance between points \( O_1 \) and \( O_0 \), as shown in Fig. 3. A point mass \( m_0 \) is added to human leg at the hip to be able to make the joint angles of the human leg and the external orthosis gravity balanced. The point mass \( m_0 \) is required because all design parameters are related to external orthosis and we do not have any design parameters in human leg. Without having this extra mass, the system of the equations that leads to the conditions of gravity balancing of the system is not compatible and thus it is not solvable.

Using Eq. (3), \( c_g s_7 \) can be written as

\[
c_g s_7 = -l_0 s_1 + l_1 c s_2 + (l_2 - l'_3) c s_3 - l'_3 s_6 s_8 + l'_0 s_8 \tag{7}
\]

Upon substitution of Eq. (7) into Eq. (6), the term \( c_g s_7 \) is expressed in terms of other joint angles, and the expression for the potential energy can be written in terms of joint angle variables as

\[
V_g = K_1 s_1 + K_2 c s_2 + K_3 c s_3 + K_4 s_6 c s_8 + K_5 \tag{8}
\]

where

\[
K_1 = -g l_0 \left( m_0 g l_0 + m_1 + m_2 + m'_1 + m'_2 + m'_3 + m'_4 \right) \tag{9}
\]

Please note that Eq. (8) involves five variables and is not in the minimal set of coordinates. Hence, any conditions obtained later are necessary conditions. Should we want to use minimal set of coordinates, all three loop constraint equations, i.e., Eqs. (2)--(4), should be used. This leads to a nonlinear equation, and the final form of \( V_g \) becomes very cumbersome to satisfy the conditions. Hence, we use only the loop constraint equation, which is in terms of sine of joint angles to eliminate one of the independent joint angles.

To achieve invariant potential energy in all configurations of leg and orthosis, we can add springs to the external orthosis, as shown in Fig. 5. One spring is connected to point \( P_1 \) on link \( l'_0 \) at one end and fixed to point \( H_1 \) on an axis parallel to gravity vector at the other end. The second and third ones are connected to points \( P_2 \) and \( P_3 \) on links \( l'_1 \) and \( l'_2 \) at one ends and fixed to points \( H_2 \) and \( H_3 \) on an axis parallel to gravity vector at the other ends, respectively. The vertical axes for the second and third springs are acquired by using two parallelograms, as depicted in Fig. 5.

The total potential energy due to the gravity and springs is given by

\[
V = V_g + V_s \tag{10}
\]

in which \( V_g \) is defined in Eq. (8) and

\[
V_s = \sum_{i=1}^{3} \frac{1}{2} k_i x_i^2 \tag{11}
\]

Here, \( k_i \) and \( x_i \) are the stiffness and extension of the \( i \)th spring.

In this work, it is assumed that the undeformed length of the spring is zero. In other words, the spring force is zero when the deformation of the spring is zero. In the physical implementation of zero free length, nonzero free length spring can be used behind the pulley where the spring force can be transmitted through a wire [15]. Therefore, the extensions of the springs are written as

\[
x_1 = d'_1 + d'_2 + 2d'_3 d'_5 s_8 \\
x_2 = d'_5 + d'_2 - 2d'_4 d'_5 s_8 \tag{12}
\]

\[
x_3 = d'_5 + d'_2 - 2d'_4 d'_5 s_8 \tag{12}
\]

where \( d'_i, i=1,\ldots,6 \), are the end point locations of the springs from \( O_5, O_7, O_9 \), respectively. Upon substitution of \( V_g \) from Eq. (8) and inserting Eqs. (7) and (12) into Eq. (10), one obtains
where $V = C_1 s_1 + C_2 s_2 + C_3 s_3 + C_4 s_4 + C_5 s_5 + C_6$

$$V = C_1 s_1 + C_2 s_2 + C_3 s_3 + C_4 s_4 + C_5 s_5 + C_6$$

with $C_i, i = 1, \ldots, 6$, defined as

$$C_1 = K_1 + k_2 d_1 d_1' l_0 / l_1'$$

$$C_2 = K_2 - k_2 d_1 d_1' l_0 / l_1'$$

$$C_3 = K_3 - k_2 d_1 d_1' (l_2 - l_1') / l_1'$$

$$C_4 = K_4 + k_2 d_1 d_1' l_2' / l_1' - k_3 d_2 d_6$$

$$C_5 = K_5 - k_2 d_1 d_1' l_2' / l_1' + k_1 d_1' d_2$$

Therefore, the coefficients $C_i$ for $i = 1, \ldots, 5$ are functions of geometric and inertia parameters as

$$C_1 = C_1(m_0, m_1, m_2, m_3, l_0, l_1', l_2', k_2, d_1, d_1')$$

$$C_2 = C_2(m_1, m_2, m_3, l_0, l_1', k_2, d_1, d_1')$$

$$C_3 = C_3(m_1, m_2, m_3, l_1', l_2', k_2, d_1, d_1')$$

$$C_4 = C_4(m_1, m_2, l_1', l_2', k_2, d_1, d_1')$$

$$C_5 = C_5(m_0, m_1, m_2, l_1', l_2', k_2, d_1, d_1')$$

These coefficients should be vanished to make the total potential energy configuration independent. Considering the coefficients $C_i$ from Eq. (15), the design variables are stiffness of springs, orthosis links’ mass, point mass $m_0$ and its location on hip joint, location of COM of orthosis links, and the end point locations of springs.

Here, we have an underdetermined system with less number of equations than variables. We have five nonlinear equations and 19 variables to solve this set of equations.

### 4 Feasible Designs

As an example, the inertia and geometric parameters of the human leg are considered for a normal subject [16], as shown in Table 1. The link lengths of external orthosis are derived such that the loop constraint equations, i.e., Eqs. (2)–(4), are satisfied during walking. Their expressions are $l_0' = 0.3$ m, $l_1' = 0.4332$ m, $l_2' = 0.4634$ m, and $l_3' = 0.2316$ m. The lengths of external orthosis links are chosen such that they are close to human limb dimensions. Snapshots of animation of the human leg and external orthosis during walking are shown in Fig. 6.

#### 4.1 A Feasible Design: 4DOF Case (3DOF for Hip and 1DOF for Knee)

As presented in the previous section, we have five nonlinear equations, namely, Eq. (14) and 19 design variables. We can assume some of the variables to be given. Upon substitution of Eq. (9) into the first two equations of Eq. (14) and simplifying the results thus obtained, one obtains

$$m_{dik} = m_{10} \left( \frac{l_{1i}}{l_1} - 1 \right) = -0.654015$$

From Eq. (16), it is easy to find out that $l_0$ must be negative. So, we can assume that $m_{10} = 1.45337$ kg and $l_0 = -0.45$ m. A physical interpretation for the negative value of $l_0$ can be considered as...
locating $m_0$ at a point with distance $l_0$ in the opposite direction of $O_1O_2$. This point mass can be attached to hip via a corset worn by the user.

The other design variables are obtained by vanishing $C_i$, $i = 2, . . . , 5$, from Eq. (14). Results are shown in Table 2. The design parameters are obtained based on the following physical constraints: (a) minimization of the total mass of external orthosis, (b) physical limits to the placement of spring connections, and (c) maximum practical values of spring stiffnesses.

### 4.2 A Feasible Design: 3DOF Case (2DOF for Hip and 1DOF for Knee)
In this case, we consider 2DOFs of hip joint (hip abduction/adduction and hip flexion/extension) and 1DOF for knee joint (knee flexion/extension). The movement is considered only in the sagittal plane and coronal plane. The conditions for the gravity balancing of the system are similar to 4DOF case presented in Sec. 4.1 because $\theta_h$ and $\theta_k$, the rotation degrees of freedom, did not appear in the expression of total potential energy, i.e., Eq. (8). So, we can use three springs and the same parameters, presented in Sec. 4.1, to gravity balanced this system.

### 4.3 A Feasible Design: 2DOF Case (1DOF for Hip and 1DOF for Knee)
In this case, we consider 1DOF of hip joint (hip flexion/extension) and 1DOF for knee joint (knee flexion/extension). Therefore, we eliminate the joint angles $\theta_h$, $\theta_k$, $\theta_l$, and $\theta_u$, associated with the hip abduction and hip rotation, by vanishing them from Eqs. (2) and (3). To achieve constant potential energy in all configurations of leg and orthosis, we can consider two cases: case (a) add one spring to the external orthosis, as shown in Fig. 7, or case (b) add two springs, as shown in Fig. 8.

Using Eq. (10) and Figs. 7 and 8, the total potential of the external orthosis and leg can be written as follows:

\[ V = C_0s_2 + C_1s_3 + C_2s_4 + C_3 \]  

The coefficients $C_i$ for $i = 1, . . . , 3$ are functions of geometric and inertia parameters of external orthosis and leg. Their expressions are given for cases (a) and (b) by the following equations, respectively:

\[ C_0 = K_0 - k_1d'_1l'_1/l'_1 \]

\[ C_1 = k_1d'_2l'_2l'_1/l'_1 \]

\[ C_2 = k_2d'_2l'_2l'_1/l'_1 \]

\[ C_3 = 1/2 k(d'_1^2 + d'_2^2) \]

\[ C_0 = K_0 - k_1d'_1l'_1/l'_1 \]

\[ C_1 = k_1d'_2l'_2l'_1/l'_1 \]

\[ C_2 = k_2d'_2l'_2l'_1/l'_1 \]

\[ C_3 = k_1d'_1l'_2l'_1/l'_1 - k_2d'_1l'_4 \]  

\[ \theta = \angle PD, H = 90° \]

\[ \angle P_ObH_1 = \theta_h - \pi/2 \]

\[ \angle P_ObH_2 = 5\pi/2 - \theta_b \]
and two springs, one can infer that using two springs decreases the inertia of the external orthosis. On the other hand, this adds to the complexity of the design.

5 Joint Torque Computation

Joint torques are not needed to keep the human leg in equilibrium at any configuration of the leg using external orthosis design. However, joint torques are needed during walking. To this end, the joint torque trajectories of the human leg during walking are computed and compared for the following two cases: (I) human leg and the external orthosis design for the feasible designs of 2DOF and 3DOF cases and (II) human leg without external orthosis. Case (I) is a closed-loop system while case (II) is an open-loop system. Given the hip and knee joint trajectories and their time derivatives during walking, the joint torques of the human leg for case (I) are computed as follows: The dependent joint angles and their time derivatives are derived in terms of independent joint ones using the loop constraint equations and their time derivatives. Next, the joint torques at the hip and knee joints are derived using inverse dynamics of the closed-loop system [17]. Also, given the hip and knee joint trajectories and their time derivatives during walking, the joint torques of the human leg for case (II) are computed using inverse dynamics of the open-loop system. As an example, consider the geometric and inertia parameters of the human leg and the external orthosis designs with one spring and two springs (2DOF cases) given in Tables 1, 3, and 4, respectively. Also, the hip and knee joint trajectories during normal walking are given the same as Ref. [18]. The effects of muscles' elasticity and friction are neglected. Using these data, the joint torque trajectories of the human leg at the hip and knee are computed for the human leg and the external orthosis design with one and two springs as well as for the human leg only. The results are shown for the design with one spring at different walking speeds of 0.92 m/s (fast walking), 0.451 m/s (normal walking), and 0.23 m/s (slow walking) in Figs. 9(a)–9(c). These velocities correspond to fast, normal, and slow walking speeds for the purposes of stroke rehabilitation. It can be concluded from these figures that higher joint torques are required at fast walking because of the inertia of the external orthosis. Also, as shown from the results, the positive hip torque represents the flexion torque and tries to move the leg forward and the negative hip torque is extension torque and it slows down the forward motion of the leg.

As an example, the joint trajectories of the hip and knee at the specific walking speed of 0.23 m/s were shown in Fig. 10. Moreover, the results for both designs with one and two springs at a specific walking speed of 0.92 m/s at the hip and the knee joints are depicted in Figs. 11(a) and 11(b), respectively. As shown, the overall joint torques in design with two springs are less than the joint torques in the design with one spring. Therefore, the design with one spring, which has less complexity, can be used at slow walking.

The simulation results corresponding to 20% of the beginning and 10% of the end of the walking period cannot be reliable since high joint torques near the start and end of the trajectories result from finite difference approximation of velocity and acceleration; high calculated acceleration thus yields erroneously high joint torque estimates. Hence, they are not shown in the figures.

Table 3 Geometric and inertia parameters of the external orthosis with one spring

<table>
<thead>
<tr>
<th>Stiffness of springs (N/m)</th>
<th>Mass (kg)</th>
<th>Length (m)</th>
<th>Length (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$k_1 = 1150$</td>
<td>$m_1' = 1.534896$</td>
<td>$I_1' = -0.11015$</td>
<td>$d_1' = d_2' = 0.2726$</td>
</tr>
<tr>
<td>$m_2 = 12.384842$</td>
<td>$I_2' = -0.32346$</td>
<td>$I_3' = -0.00434$</td>
<td>$\Sigma m_i' = 17.33$</td>
</tr>
</tbody>
</table>

Total mass of external orthosis

Table 4 Geometric and inertia parameters of the external orthosis with two springs

<table>
<thead>
<tr>
<th>Stiffness of springs (N/m)</th>
<th>Mass (kg)</th>
<th>Length (m)</th>
<th>Length (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$k_1 = 1150$</td>
<td>$m_1' = 1.555899$</td>
<td>$I_1' = -0.103648$</td>
<td>$d_1' = d_2' = 0.22921$</td>
</tr>
<tr>
<td>$k_2 = 1000$</td>
<td>$m_2' = 6.39878$</td>
<td>$I_2' = 0.310931$</td>
<td>$I_3' = 0.24274$</td>
</tr>
<tr>
<td>$m_3 = 3.410124$</td>
<td>$I_3' = -0.000755$</td>
<td>$\Sigma m_i' = 11.3675$</td>
<td></td>
</tr>
</tbody>
</table>

Fig. 9 (a)–(c) Joint torque trajectories at the hip $\tau_2$ and the knee $\tau_3$ for the design with one spring (2D case) at different walking speeds: (a) 0.92 m/s, (b) 0.451 m/s, and (c) 0.23 m/s
The results are shown for the design with three springs for the 3DOF case at different walking speeds of 0.92 m/s (fast walking), 0.451 m/s (normal walking), and 0.23 m/s (slow walking) in Figs. 12(a)–12(c) for both human leg and external orthosis design and human leg without external orthosis. As shown, the joint torques are higher for the human leg and external orthosis as compared with their values for the human leg only while we have a high speed walking, as shown in Fig. 12(a). Conversely, there are less joint torques for the human leg and external orthosis in a slow walking, as depicted in Fig. 12(c). Therefore, the results verify that the joint torques are much smaller for human leg and external orthosis design at slow walking. This design will be used for training a typical stroke patient to improve his/her ability of walking. Therefore, the walking speed is not high.
6 Conclusion

The paper provided the design of an external orthosis to remove gravity load on the joints of a human leg during walking. This design connects to the human leg at a single point and thus does not have the issue of joint alignment between the human leg and the device. However, it increases the inertia of the system, which may be a drawback during fast walking. The results showed that the 3D design has smaller inertia with respect to the 2D design, see Table 2. Moreover, the 3D design supports one extra degree of freedom for walking than the 2D case. In the 2D design, the results showed that the design with two springs is more desirable at fast walking. Based on the results, the best range of human walking speed for stroke patient is between 0.23 m/s and 0.451 m/s. Noteworthy is that the normal walking speed of a typical stroke patient lies within the mentioned range.

Appendix A: Position of Human Leg and External Orthosis

To obtain human leg position, frame $i$ with $X_i$, $Y_i$, and $Z_i$ axes was assigned at point $O_i$, $i=1,2,3,4$. Upon derivation of the rotation matrix of each frame with respect to reference frame ($F$), the position of each link was obtained. Reference frame ($F$) was attached to the walking frame. Axes of each frame of $F$, 0, 1, 2, and 3, were defined by $(x_0,y_0,z_0)$, $(x_1,y_1,z_1)$, $(x_2,y_2,z_2)$, and $(x_3,y_3,z_3)$, respectively. The rotation matrices of each frame with respect to prior frame using Euler angles [19] are shown in Figs. 13–16.

Having these rotation matrices, the position of each point can be defined by multiplying the rotation matrix of the frame attached to each point with respect to reference frame by the posi-
The orthosis links can be derived using the same trend.

Appendix B: Location of Center of Mass of Human Leg and External Orthosis

The expressions are

\[ r_{oc} = (l_0 s_0 c_1 - l_1 c_0 c_2) \hat{z} + (l_0 s_1 + l_1 c_1 s_2) \hat{j} + (l_0 c_0 c_1 - l_1 s_0 c_2) \hat{k} \]

\[ r_{oc'} = -(d_0 l_0' s_0 c_8 + l_1' c_0 c_7) \hat{z} + (l_0' s_8 + l_1' c_0 s_7) \hat{j} + (l_0' c_0 c_8 + l_1' s_0 s_7 - l_1' s_0 c_7) \hat{k} \]

\[ r_{oc''} = -(d_0 l_0'' s_0 c_6 + l_1'' c_0 c_5) \hat{z} + (l_0'' s_6 + l_1'' c_0 s_5) \hat{j} + (l_0'' c_0 c_6 + l_1'' s_0 s_5 - l_1'' s_0 c_5) \hat{k} \]

where \( \hat{z}, \hat{j}, \) and \( \hat{k} \) are unit vectors along X, Y, and Z, respectively.

References


