MODELLING AND ANALYSIS OF THE FALLING PROCESS BASED ON A FIVE-LINK GAIT MODEL

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Abstract

In this paper we develop a dynamic model to mimic and simulate human falling process when tripping over an obstacle. The work consists of three parts. First a five-link gait model is established and the joint trajectories of knee and hip are generated. Next several constraints including the physical, physiological and environmental constraints are taken into consideration. Hence the fall process can be modelled in a realistic manner. Next a number of important factors that characterize the fall process, such as reaction time, inclining angle, stable range, initial speed, are investigated. The whole falling process is formulated according to biological characteristics and stability analysis, which offers an effective way to identify and distinguish falling from normal activities, so that the falling can be predicted and a prevention device can be activated in advance.

Key Words

Falling, gait model, constraints, biological characteristics, stability analysis

1. Introduction

Nowadays in our modern society, the health conditions of elders are attracting more and more attentions. Falling accident becomes a major problem threatening elders’ health, in that almost one-third of 65-year old or elder people fall at least once a year [1], [2] and [22]. What is even worse is that 83% of fatal falls lead to the cause of deaths [3]. Furthermore, approximately $20.2 billion are spent each year in the treatment, which is a significant financial burden [4]. Thus many scientific researchers and institutions have been exploring ways to predict the falling ahead and ultimately to protect the elderly from falling.

A number of researches on detecting falls have been carried out in recent years. Using accelerometer-based systems to distinguish daily activities from accident situations such as falling is a common approach [5]–[8]. Some other approaches use cameras in the fall-detection systems based on surveillance of in-house movements, to make the detection more accurate [9], [10]. Another approach combines floor vibrations and acoustic sensing to identify human falls [11]. All these approaches are based upon the sensor data of human motions, acquired by implementing experiments on human normal activities and abnormal motions, so as to build appropriate algorithms to detect falls. After collecting data, appropriate algorithms such as neural network [12] are developed to classify normal motions from abnormal motions such as falling.

Another approach to investigate into the researches of human falling is based on the dynamic model of bipedal robot, such as [13], in which a 12 degree-of-freedom linkage model is set up to simulate a trip and fall during gait. A number of control schemes are proposed to generate and maintain the trajectories of the robot, such adaptive trajectory tracking control [14], neural network control [15], etc.

In this paper, a mathematically dynamic model is developed to simulate the motion of falling. This model is based upon a five-link bipedal robot, which can perform ordinary walking gait. The computed control method is applied to control the robot moving. Certain constraint equations, representing the physical, physiological and environmental constraints in real life, are imposed on the model to simulate the effects of possible impacts and constraints during the falling.

It is worthwhile highlighting the differences between our work and [13]. In [14], internal forces such as joint torques are generated from a locomotion profile that is measured from a sample subject. In our work, the internal forces or joint torques are generated according to the first principle, namely, through kinematic constraints including both physical and physiological constraints. In other words, the falling process in [13] is imitation based, whereas ours is an analytical model. Our model is generic in the sense that the model can be applied to any one with the individuals parameters. On the contrary, the imitation-based model cannot be applied to produce the falling process of other subjects whose kinematic behaviour differs from the sample subject. By virtue of the analytic model, the analysis on the stability and other important factors that characterize the fall process, such as reaction
time, inclining angle, stable range, initial speed can be investigated. Such analysis cannot be conducted using an imitation-based model.

The paper is organized as follows. The gait model with five links is given in Section 2. Section 3 demonstrates how the relative constraints are formulated and incorporated into the dynamic equations. Analysis and investigation of reaction time and balance conditions are discussed in Section 4. Simulation results are given in Section 5. Finally, the conclusion is given in Section 6.

2. Five-Link Gait Model

The bipedal robot model is taken from [16], which consists of five links including the torso and two legs. Each leg consists of two links as shown in Fig. 1. The five links are connected via four joints, which are joint 3 and joint 4 at the hip and joint 2 and joint 5 at the knees. Though feet ankle joints are not considered in this model, it is assumed that the actuated torques can be applied at each ankle. The locomotion of the robot is only considered within the sagittal plane. Also it is assumed that the joints are friction free and there is no slippage between the limbs and ground.

2.1 Joint Trajectories

The design of the joint trajectories is critical for this dynamic model as it defines the normal human walking gait. Different objective and constraints can be taken into consideration such that the planned locomotion or gait is more authentic, for example, in [17] the objective of minimum energy consumption, the constraints on stability, walking speed, etc. In this paper, the joint trajectories are derived according to [16]. Another set of constraints is considered including geometrical constraints, minimization of the effect of impact, repeatability of the gait and so on. One complete walking cycle includes a single support phase (SSP) and a double support phase (DSP). During SSP, the swing leg moves in the forward direction while the stance leg is pivoted on the ground. During DSP, both legs are pivoted on the ground and the body moves forward slightly. The trajectory profiles of the five joint angles are shown in Fig. 2, from which we can find that the desired angle profiles are smooth and continuous. The stick diagram of the normal walking cycle is shown in Fig. 3.

2.2 The Closed-Loop Dynamic System

Human joint is a closed-loop control process, in which the joint feedback is used to produce appropriate joint torques that subsequently generate the desired joint angle profiles. The dynamic equations of the model can be derived using Lagrangian’s equation of motion [18]. The general expression of the equation is shown below:

$$D(\theta)\ddot{\theta} + h(\theta, \dot{\theta}) + g(\theta) = ET_q$$

where $D(\theta) \in R^{5 \times 5}$ is the mass-inertia matrix, $h(\theta, \dot{\theta}) \in R^{5 \times 1}$ is a vector of centrifugal and Coriolis terms, $g(\theta) \in R^{5 \times 1}$ is a vector of gravity terms and $E \in R^{5 \times 5}$ is the transition matrix from actuated torques $T_q \in R^{5 \times 1}$ to generalized torques $T_\theta \in R^{5 \times 1}$, where $T_\theta$ is derived from $(\theta_1, \theta_2, \theta_3, \theta_4, \theta_5)$ and $T_q$ is derived from $(q_1 = \theta_1, q_2 = \theta_1 - \theta_2, q_3 = \theta_2 - \theta_3, q_4 = \theta_3 + \theta_4, q_5 = \theta_4 - \theta_5)$. Note that in human walking process, only relative joint angles $q_i(i = 1, \ldots, 5)$ can be controlled by the actuated torques.
Figure 3. Stick diagram of the normal walking cycle. The red stick denotes the anterior swing leg, which is also the posterior stance leg. The blue stick denotes the anterior stance leg, which is also the posterior swing leg. The green stick denotes the torso.

Figure 4. Block diagram of the control system. The control system is implemented by the computed torque method. The computed torque method is used to compute the joint torques, then generate the motion trajectories. Computed torque method is a linearizing control law, because it can cancel the nonlinear elements in the dynamic system such that the overall system is linear [19]. From (1) and applying the computed torque method, the actuated torques are:

\[ T_q = E^{-1}[D(\theta)u + h(\theta, \dot{\theta}) + g(\theta)] \]

\[ u = \dot{\theta}_d + K_D e + K_P e \]  

(2)

where \( K_D \) and \( K_P \) are the derivative and proportional gain matrices, and \( e \) denotes the position error vector, which is defined as \( e = \theta_d - \theta \). The block diagram of the resulting control system is shown in Fig. 4. It is assumed that a simple PD controller is adequate to generate accurate gait movement.

In this work, we assume that the feedback gains are \( K_D = diag(2\lambda) \in R^{5 \times 5} \) and \( K_P = diag(\lambda^2) \in R^{5 \times 5} \), which yield a critical damping performance. \( \lambda \) determines a desired bandwidth. \( \lambda \) should be selected sufficiently large to obtain a fast response. However, a large \( \lambda \) will result in large driving torques [20]. Thus it should not exceed the limits of actuation by the muscles.

3. Impacts and Constraints

Impacts and constraints occur during normal walking or falling process. They play important roles when computing normal walking and falling, and should be taken into consideration to make the computed gaits realistic. Generally speaking, there are seven physical constraints on impacts, including impact between the swing limb and the obstacle, impact between the joints and the ground. Further, there are two important biological constraints on the rotational angles of knee joints. These joints should not exceed the maximum angles due to the biological structure of the joints.

3.1 Constraint between Swing Limb and Obstacle

When the limb of the swing leg contacts the obstacle such as a wall, a resultant impact occurs. A set of holonomic constraints prevent the swing limb from moving forward in the sagittal plane, but the limb can move freely in the vertical plane. The holonomic constraint equation is:

\[ \phi_{end,obstacle}(\theta) = x_{end}(\theta) - x_b = S_{impact} \]  

(3)

where \( \phi_{end,obstacle}(\theta) \) denotes the horizontal position of the swing limb after the impact happens, \( x_{end} \) denotes the displacement of the swing limb along \( x \) direction, \( x_b \) denotes the origin and \( S_{impact} \) denotes the position of the limb when impact occurs along \( x \) direction. By differentiating equation (3) twice with respect to (w.r.t.) time, we can obtain:

\[ \dot{\phi}_{end,obstacle} = J_{end,obstacle}(\theta) \dot{\theta} = 0 \]

\[ \ddot{\phi}_{end,obstacle} = \ddot{J}_{end,obstacle} \dot{\theta} + J_{end,obstacle}(\theta) \ddot{\theta} = 0 \]  

(4)
where }J_{\text{end,obstacle}}(\theta)\text{ is the Jacobian matrix. Thus, the
dynamic equation for DSP can be expressed by:

\[
D(\theta)\ddot{\theta} + h(\theta, \dot{\theta}) + g(\theta) = ET_q + J^T_{\text{end,obstacle}}(\theta)\lambda_{\text{end,obstacle}}
\]

where }J_{\text{end,obstacle}}\text{ is the Lagrange multiplier, which can be expressed as:

\[
\lambda_{\text{end,obstacle}} = (J_{\text{end,obstacle}}D^{-1}J^T_{\text{end,obstacle}})^{-1}
\]

\[
\{J_{\text{end,obstacle}}D^{-1}[h(\theta, \dot{\theta}) + g(\theta) - ET_q] - J_{\text{end,obstacle}}\dot{\theta}\}
\]

The impact results in instantaneous changes of the joint velocities. In [21], a method is proposed to calculate the
velocity change in the below form:

\[
\dot{\theta}_+ - \dot{\theta}_- = D^{-1}J^T_{\text{end,obstacle}}(J_{\text{end,obstacle}}D^{-1}J^T_{\text{end,obstacle}})\Delta \dot{s}
\]

\[
\Delta \dot{s} = \dot{s}_+ - \dot{s}_-
\]

where }\dot{\theta}_-\text{ and }\dot{\theta}_+\text{ represent the velocities of the joints before and after the impact, respectively, }\Delta \dot{s}\text{ represents the instantaneous velocity change due to the impact point in the Cartesian Space. }\dot{s}_+\text{ represents the velocity of the limb after the impact and }\dot{s}_-\text{ represents the velocity of the limb before the impact. In our case, the speed of the free end of the swing leg after the impact is zero, which means }\dot{s}_+ = 0. \text{ Thus we can derive the instantaneous velocity change }\Delta \dot{s}\text{ as:}

\[
\Delta \dot{s} = 0 - \dot{s}_- = -J_{\text{end,obstacle}}\dot{\theta}_-
\]

Combining relations (7) and (8), the joint velocities after the impact can be calculated as below:

\[
\dot{\theta}_+ - \dot{\theta}_- = D^{-1}J^T_{\text{end,obstacle}}(J_{\text{end,obstacle}}D^{-1}J^T_{\text{end,obstacle}})(-J_{\text{end,obstacle}}\dot{\theta}_-)
\]

\[
(\theta)\text{ denotes the Lagrange multiplier, which can be expressed as:}

\[
\phi_{\text{end,ground}}(\theta) = y_{\text{end}}(\theta) - y_g = 0
\]

3.2 Constraints between Body and Ground

During the falling process, the swing limb will hit the ground. Analogous to previous derivations, the dynamic equation of gait is subject to the holonomic constraints. Thus we need to determine first the holonomic constraint equations. When hitting the ground, the swing limb is prevented from moving downward. The holonomic constraint equations can be derived accordingly.

**Constraint between swing limb and ground:**

\[
\phi_{\text{end,ground}}(\theta) = y_{\text{end}}(\theta) - y_g = 0
\]

where }\phi_{\text{end,ground}}\text{ denotes the vertical displacement of the swing limb after the impact, }y_{\text{end}}\text{ is the displacement of the swing joint along y axis, and }y_g\text{ denotes the coordinate of the ground, which is chosen to be 0.}

**Constraint between stance knee and ground**

After completely falling down, the stance knee will hit the ground. This constraint is expressed as:

\[
\phi_{\text{stanceknee,ground}}(\theta) = y_{\text{stanceknee}}(\theta) - y_g = 0
\]

where }\phi_{\text{stanceknee,ground}}\text{ denotes the vertical displacement of the stance knee after the impact and }y_{\text{stanceknee,ground}}\text{ is the displacement of the stance knee along y axis.}

**Constraint between swing knee and ground:**

\[
\phi_{\text{swingknee,ground}}(\theta) = y_{\text{swingknee}}(\theta) - y_g = 0
\]

where }\phi_{\text{swingknee,ground}}\text{ denotes the vertical displacement of the swing knee after the impact and }y_{\text{swingknee,ground}}\text{ is the displacement of the swing knee along y axis.}

3.3 Joint Constraints

From biological structure, each joint has a limited range of rotation [23].

From the model shown in Fig. 1, a normal gait requires that }\theta_1 > \theta_2\text{ and }\theta_5 > \theta_4.\text{ These two conditions are used as the joint constraints. When }\theta_1 = \theta_2\text{ and }\theta_5 = \theta_4,\text{ some constraints should be imposed on the knee joints so that they will not exceed the extreme condition to perform an unreasonable gait style, such as bending towards the opposite direction. The constraint equations can be derived in a similar way as the previous cases.}

**Stance knee joint constraint:**

\[
\phi_{\text{joint,1}}(\theta) = \theta_1 - \theta_2 = 0
\]

where }\phi_{\text{joint,1}}\text{ denotes the relevant angle of link 1 and link 2.}

**Swing knee joint constraint:**

\[
\phi_{\text{joint,2}}(\theta) = \theta_4 - \theta_5 = 0
\]

where }\phi_{\text{joint,2}}\text{ denotes the relevant angle of link 3 and link 4.}

3.4 Synthesis of Constraints

During the falling process, several constraints may occur simultaneously. For example, when both }\phi_{\text{end,obstacle}}(\theta)\text{ and }\phi_{\text{end,ground}}(\theta)\text{ occur, the augmented constraint equation is:

\[
\phi(\theta) = \begin{bmatrix}
\phi_{\text{end,obstacle}}(\theta) \\
\phi_{\text{end,ground}}(\theta)
\end{bmatrix} = \begin{bmatrix}
x_{\text{end}}(\theta) - x_b \\
y_{\text{end}}(\theta) - y_g
\end{bmatrix} = S_{\text{impact}}
\]

\[
\begin{bmatrix}
S_{\text{impact}} \\
\lambda_{\text{impact}}
\end{bmatrix}
\]

Accordingly we can compute the Jacobian }J = \frac{\partial \phi}{\partial \theta} \in \mathbb{R}^{2 \times 5}\text{ and the augmented Lagrange multiplier }\lambda = \lambda(DJ^{-1}J^T)^{-1}(JD^{-1}[h(\theta, \dot{\theta}) + g(\theta) - ET_q] - J\dot{\theta}) \in \mathbb{R}^{2 \times 1}.

Similarly, if multiple constraints (3) and (10)–(15) occur, the augmented constraint equation is obtained by adding more constraint equations into (16). As a consequence, the constrained dynamic equation can be derived analogous to (3)–(9) with the augmented constraints.
4. Stability Analysis

4.1 Reaction Time

During normal walking cycles, the control torque $T_q$ keeps the joints to follow the desired trajectory. While in the falling process, the situation is different. At first, when the falling starts, i.e., the swing leg hits the obstacle, the joint torques try to keep the body to follow the normal trajectory. After a short interval, the fall body begins to react to the constraints. This makes sense from the biological point of view, because there is a delay for the human body to react to stimuli. During this delay time, called reaction time, human beings try to keep doing things as normal. After the reaction time, human body becomes aware and begins reacting to the stimuli by taking actions. During the reaction time, the body tries to keep the normal walking cycle despite the constraint of the obstacle. After the reaction time, the body tries to hold the body to prevent from falling forward.

Reaction time is a very important factor in preventing falling. If the response time is short, human body can react to the tripping at an earlier phase and gain a longer interval to adjust posture to avoid falling. If the reaction time is long, human body do not have enough time to adjust posture to react to the tripping. Thus the shorter the response time, the higher possibility the falling can be prevented. It is known that the reaction time of a normal person is about 0.1s and shorter for a trained person.

4.2 Inclining Angle

The torques human joints can produce are limited. If the joints cannot produce adequate torques, the posture cannot be held and the body will incline forward. The angle that human can incline without losing stability is limited. When the body inclines forward exceeding a critical angle, the body loses balance and falls down as a free falling body. During the free falling process, the joints cannot produce torques, which behaves like a muscle-relaxed fall.

In this paper, we assume that the critical angle is 30°, considering the supporting effect of human foot. The inclining angle is defined as the angle between the vertical axis and the line that connects the centre of gravity (COG) and the supporting limb. If the stance limb is the supporting limb as is shown in Fig. 5(a), the expression of inclining angle is defined as $\alpha_1 = \arctan \frac{x_c}{y_c}$, where $(x_c, y_c)$ are the coordinates of the COG.

If the anterior swing limb is the supporting limb as is shown in Fig. 5(b), the expression of inclining angle is defined as $\alpha_2 = \arctan \frac{x_c - x_{end}}{y_c - y_{end}}$.

4.3 Balance Analysis

When the swing limb hits the ground, a supporting force is generated by muscles and points upwards along the limb. There are two factors that determines the balance after tripping, the initial posture or the position of the COG and the momentum.

First investigate the influence of the initial posture. Two possible initial postures and force distributions are shown in Fig. 6.

In the case shown in Fig. 6(a), the swing limb is taken as the reference point to analyse the torques that determine the balance condition. There are two torques w.r.t. the reference point, that is, the torque generated by force $f_1$ in the clockwise direction, $\tau_1$, and the torque generated by the gravity force $g$ in the clockwise direction, $\tau_g$. The net torque is $\tau_{net} = \tau_1 + \tau_g$ in the clockwise direction.

We can find that in this case the net torque is not zero, thus the body inclines forward. Furthermore, the effect of the force $f_1$ will make the body to incline further. Thus we can conclude that when the swing limb hit the ground, if it is behind the COG, the net torque can never be zero. Thus nothing can be done to prevent the body from falling down.

Now consider the case shown in Fig. 6(b), where the swing limb is ahead of the COG in the sagittal plane. We take the stance limb as the point of reference to analyse the torques. There are also two torques w.r.t. the reference point, that is, the torque generated by force $f_2$ in the clockwise direction, $\tau_2$, and the torque generated by the gravity force $g$ in the clockwise direction, $\tau_g$. The net
Figure 7. Force analysis.

The torque is $\tau_{net} = \tau_2 - \tau_f$, from which we can find that if $\tau_2$ is sufficiently large, an equilibrium state can be obtained, and it is possible to prevent falling.

We can further analyze the balance condition with the initial posture (b) shown in from Fig. 7, but with a nonzero initial horizontal speed. Suppose, when the swing limb is blocked at $X_f$, the COG lies at $P_0$, and the initial horizontal speed is $v_0$. To keep balance, the final velocity of the COG should decrease to 0 when it reaches the critical position $P_f$ that is right above the contact point between the foot and obstacle. Then we have the relationship by law of energy conservation:

$$mv_0^2/2 = (l - l \cos \alpha_0)mg + \bar{w}$$  \hspace{1cm} (17)

where $\alpha_0$ denotes the initial angle between the sustaining link and the vertical axis, $\bar{w}$ represents the energy generated by the force $f$ of the supporting limb. The energy $\bar{w}$ can be calculated as below:

$$\bar{w} = \int_{X_0}^{X_f} f \, dx = \int_{0}^{l} \sin \alpha_0 \, f \, dx$$
$$= \int_{\alpha_0}^{\pi} -mg \tan \alpha \, d\alpha = \int_{\alpha_0}^{\pi} -mg \sin \alpha \, d\alpha$$  \hspace{1cm} (18)
$$= mgl(1 - \cos \alpha_0)$$

Combining (17) and (18) we can get the below relationship:

$$mv_0^2/2 = (l - l \cos \alpha_0)mg + mgl(1 - \cos \alpha_0)$$
$$= 2mgl(1 - \cos \alpha_0)$$  \hspace{1cm} (19)

Thus the initial speed limit can be calculated according to the above relationship:

$$v_0 = 2\sqrt{gl(1 - \cos \alpha_0)}$$  \hspace{1cm} (20)

We can conclude that the balance can be achieved if the initial speed of COG does not exceed $v_0 = 2\sqrt{gl(1 - \cos \alpha_0)}$ when the impact occurs.

5. Simulation Results

In this section, tripping over an obstacle of the five-link gait model is simulated. Figure 8 shows the falling process in one situation in which the obstacle blocks the limb of the swing leg while the limbs of two legs are parallel.

Figure 9 shows the falling process through 12 phases. At the beginning the gait is normal, which is shown in phases (a) and (b). When the limb of the swing leg reaches to the place parallel to the stance limb, it is tripped by the obstacle and cannot move forward, which is shown in phases (c)--(e). The phase (f) shows that the limb of the swing limb hits the ground. The phases (g) and (h) show the phenomenon that the swing leg drags along the floor,
Figure 9. The falling gait profile in 12 phases.

Figure 10. Acceleration profiles of the hip motion during normal gait and falling along $X$ axis.

which is a likely phenomenon [14]. Then the phase (i) shows that the stance knee hits the ground and the phase (k) shows that the hip and the swing knee are in contact with the ground.

Figure 11. Acceleration profiles of the hip motion during normal gait and falling along $Y$ axis.

Figure 12. Animations for the normal gait and tripped falling gait from [13].

Figure 10 shows the acceleration of hip during normal gait and falling process along the $X$ axis. Figure 11 shows the acceleration of hip during normal gait and falling process along the $Y$ axis. From Figure 10 and 11 we can see that there are sudden changes in acceleration which are not observed in other normal activities, and the magnitude of acceleration during falling is much larger than other
situations. This observation is consistent with the experimentally measured acceleration variation patterns [8].

A dynamic model was built in [13] to simulate a trip and fall during gait. In this approach, a 12 degree-of-freedom linkage model is built to simulate the human gait. The foot-obstacle contact force for a healthy subject tripping on an obstacle is measured in an experiment and applied to the swing limb of the forward dynamics model. In their experiment, a muscle-relaxed fall was simulated with the muscles of the body assumed to be inactive, which is similar to our falling model. In this way the animations for the normal gait and tripped falling gait were generated, as is shown in Fig. 12. Comparing Figs. 9 and 12, we can find that the falling gait generated by our analytic model is similar to the real falling process obtained in [13], even though the exact values of model parameters in [13] are not available, thus are not used in our analytic model.

6. Conclusions

The dynamic modelling of human falling process is presented in this paper. Based on a five-link gait model, a number of important constraint equations are applied to the dynamic motion model to realize the possible impacts and constraints during a falling. Important factors such as reaction time and inclining angle are considered to make the process more realistic. An analysis is performed to find out how to prevent falling from the point of forces and torques. Comparing with other experimental results of falling performed by human, the simulation results show that the acceleration obtained from this model can reflect the characteristic of falling. The simulated falling process appears reasonable.

The bipedal model consists of only five links, i.e., the torso, two thighs and two shanks. In our future work, the arms and the head would be taken into the model, because they play an important part in the balance-keeping. Other falling processes will also be investigated, such as fall due to slipping or fall due to stepping into hollow without awareness.

References


Biographies

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