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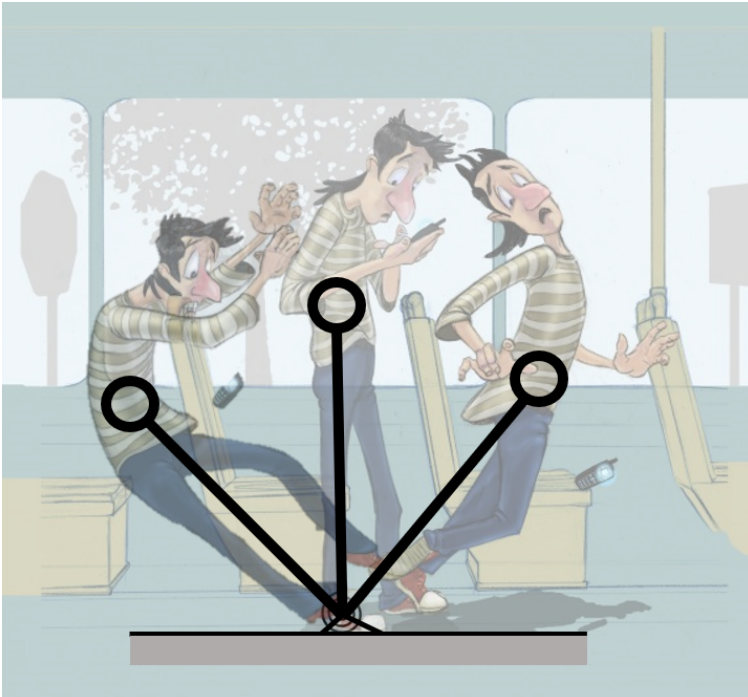

Model of a standing bus passenger: Modelling Safety for Non-Impact Collisions

Project in Automotive Engineering-TME180

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Project Report In Automotive engineering 2021:03

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Abstract

Public transport has been an efficient and convenient mode of transport for diverse population, though they are considered safe, but considerable injuries to passengers occur even due to non-crash related incidents. Evasive maneuvers, sudden accelerations or braking influence the posture of a standing passengers and result in them loosing balance. This project aims at developing a model to simulate the motion of a standing passenger influenced by distinct acceleration profiles and predict the time of when the passenger's foot is off the ground. Extreme limits for imbalance is decided based on the position of the center of mass with respect to the ankle to find the instance for the event to occur. In order to capture the passenger response during these incidents a mathematical model is designed based on Single inverted pendulum (SIP) with a torsion spring at the ankle. The SIP models the motion of the standing passenger. In the model the torques acting on the ankle are two different kinds, one due to the center of mass and the other due to the muscle influence. A feedback controller consisting of a proportional and a predictive part is designed to control the torque influence due to the muscle actions. The stiffness of the ankles are calculated based on the combined torques due to the muscles and the center of mass. The output i.e. the predicted time for imbalance is influenced by various factors such as the acceleration jerk, amplitude and duration along with the reaction time of the person. The results obtained can be further used in the implementation of passenger safety features in the transportation industry as the model in this project can help in detailed study of the factors affecting the posture during standing and the prediction of the time for imbalance. Knowledge of the passenger's state of balance can be an input signal to activate a safety feature. The model developed also has a lot of applications in the development of complex Human Body Models (HBM).

Keywords: Non-crash incidents, Evasive maneuvers, Single Inverted Pendulum, torsion spring,Feedback controller, HBM

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Contents

List of Figures

List of Tables

1
Introduction

1.1 Background

Public transport has been an efficient and convenient mode of personal mobility for humans spanning all ages, economic statures, and physical capabilities. Though public transport such as bus and tram are considered safe, considerable injuries to passengers occur even due to non-crash related incidents. These include sudden accelerations or braking and the passenger not being able to maintain balance due to their posture at the time of incident. i.e., whether they have balanced themselves by holding on to structures around them while standing, the direction they are facing, passenger condition etc. The balancing capability and acumen of individuals are dependent on various aspects, personal and vehicular. For example, age, gender, physical strength and so on, and aspects of the vehicle manoeuvres such as accelerations (jerks, amplitude, duration).

1.2 Problem Description

The public transport system sees a lot a of non-collision related incidents which lead to numerous injuries. This project aims at developing a model to simulate such incidents in order to design, develop, and simulate safety systems which are used to avoid such injuries in the future. In these future simulations, gender and age parameters will be especially accounted for as an analysis point.

2

Literature Review

Public transportation is important for personal health, social inclusion, and maintaining older adult independence. In many societies, the proportion of older people is rising and therefore it is important to ensure safety in public transportation [4]. Most bus passenger injuries happen when the passenger falls due to sudden acceleration, sudden braking , or sudden turning of the bus driver or when boarding the bus[4][5]. It is shown that the 54% of injuries obtained in busses are non-collision injuries. There is also an increase of 24% over the last 10 years [4]. As already mentioned, they identified two main scenarios when injuries happen: 37% of injuries happened when alighting or boarding a bus. In that scenario, the bus was stationary. The other scenario is balance loss while the bus is moving and accounts for 17% of injuries. 28% and 27% of these happened during sudden braking and sudden acceleration, respectively, of the bus. This is only considering the MAIS2 cases [4]. Female passengers are more sensitive to injuries than male passengers. Also the age group most affected is 65 and above for females [5]. For males the most affected age group is 25 to 64 [5]. The mean risk of injuries when falling in a bus or tram or other public transport vehicle has been calculated. It is around 0.3 to 0.5 per million passenger kilometers but a wide scatter around the mean exists. It ranges from 0.04 to 1.4 injuries per million passenger kilometers. The same has been calculated for the risk of injury when boarding or alighting a public transport vehicle. It is around 0.8 to 1.7 per million passenger. Again, a wide scatter exists with a range from 0.04 to 4.5 per million passenger [5].

Human movement and balance is a complex problem to be defined mathematically and its control aspects are even more complex to be modelled owing to the dynamic complexity of human muscle and skeletal system. Thus, to study any sort of human behavioural aspects in terms of gait, balance or other characteristics, many simplifications are made in building a mathematical model. When deciding on a model, previous studies chose between a single inverted pendulum (SID) or a double/dual inverted pendulum (DIP). Most related studies model the human as an inverted pendulum with the concentrated mass/pendulum bob depicting the total mass of the human concentrated at that point i.e., center of mass and the link of the pendulum being usually mass less rigid link[10]. Also the pendulum is considered to be at the saggital plane of the human i.e., a plane splitting the human body into left and right sections and does not take the lateral disturbances or influences into account. The length of the pendulum arm is based on the anthropometry of the human and is arrived at as a representative quantity based on some previous studies on human balance. [11]

Alternatively, a dual link inverted pendulum can be assumed with the intermediary link depicting either the hip or the ankle. This model would be extremely complex as it involves setting up of dependencies between the movement of the ankle and that of the knee or the hip, which ever is considered in the model. Understanding and controlling such a model would be extremely complex and thus will not be pursued in this study.

Studies show that the difference in accuracy of the foot off time during an imbalance in posture does not differ much between the SIP and DIP as SIP sufficiently captures all the dynamics of posture adjustment for a standing passenger [18]. Also, cerebral palsy patients tend to move their trunk as a single rigid body. That means a SIP is sufficient enough to meet the requirements for this study. [17]

Another study was noted where the maximum voluntary joint torque as a function of joint angle and angular velocity applied to the lower limp was researched. They developed a model using different gender and age groups varying in height, weight and mass. Using averages of the above mentioned age groups they came to the conclusion of about 200 N-m for the maximum hip extension and hip flexion [19]. A study on balance criteria says that it is a required to confine projection of the CoM within the Base of Support (BoS) in order for the body to remain balanced while standing [20]. Also, the horizontal velocity of the CoM should be considered in describing the feasible movements for the control of balance because it governs the destiny of the horizontal position of the CoM over the BoS [20]. Change in support strategies are used to recover or maintain balance. Those strategies are for example stepping forward or backwards or moving the upper body to shift the location of the CoM [14]. The fixed- support hip strategy involves the use of the hip flexors or extensors to generate shear forces at the feet that act to decelerate the CoM [14]. Numerous studies exist that aim to understand and objectively define the parameters that affect human motion and balance of which few relevant ones have been mentioned and used in this project.

3

Methods

3.1 Modeling Approach

As determined earlier, a single inverted pendulum (SIP) model is assumed as adequate and accurate enough for this study. The SIP model has a few attributes that determine its behaviour which are derived from the human anatomy and anthropometry. These attributes are discussed further.

3.1.1 Anthropometry

To represent the human body in terms of an SIP the mass and height of the human has to be simplified in terms of the mass of the pendulum and the length of the link. This approximation is done by using a simple proportionality which has been taken from previous human balance studies [11] where in the mass of the human is concentrated at a perceived centre of mass (CoM). This height is calculated as in eqn 3.1 and the representation of such SIP is approximately depicted in Figure 3.1 where M represents the CoM mass of the human and L is the center of mass height.

$$CoM = 0.575 * height \tag{3.1}$$

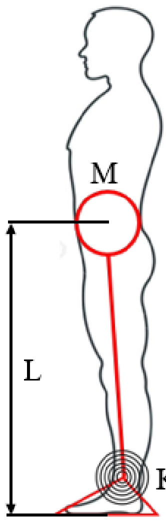

Figure 3.1: CoM Height

Through the above described formula one can determine an approximate height of pendulum for individuals, but this cannot be the case when a whole demographic of people is to be considered i.e., people of various age, physical build, heights, gender etc. An average or mean of the heights and weights does not constitute a proper representation of the demographic being studied. Averages are too simplistic and one-dimensional. The 50^{th} percentile gives a good understanding of the real performance characteristics [9]. This is mitigated by classifying the required demographic of heights and weights using anthropometry. Through anthropometric data the 50^{th} percentile male and female are taken into consideration for the preliminary model, whose dimensions are given in table 3.1 [6]. The reason for this choice is based on the available volunteer test data that is to be used to compare, tune and verify the results in this project. Also, the 50^{th} percentile model will be used as a baseline to tune the model to represent other demographics using suitable scaling factors.

Such scaling factors need an extensive study in its own right to be able to sufficiently depict the behaviour of varied demographic. For example, older people tend to be physically weaker and less agile owing to their physical degradation than their younger counterparts. Also, young children and inebriated people tend to have weaker reaction times and strength to recover themselves in cases described in this project. All these parameters cannot be accurately compared or verified as there is no volunteer data for these categories of people. Ethical and moral constraints prevent us from conducting volunteer tests with such categories of people and is not even realistic to consider such trials.

Table 3.1: Antrophometry of Human Body Models

	height $[cm]$	**CoM height** $[cm]$	**weight** $[kg]$
male	175.3	100.79	77.3
female	161.8	93.03	62.3

3.1.2 Ankle model

An aspect that can be used to determine the loss of balance of a human being is to determine the limits of the human anatomical aspects that contribute for the function. One such aspect of the human anatomy is the ankle, an integral part of human gait, balance and balance recovery. Human ankle is one of the first joints that tend to be affected when there is an input that tends to push a standing person off balance i.e., vehicle acceleration in the case of this project.

As mentioned earlier there are different strategies to study the human balance, ankle strategy, hip strategy and step strategy [16]. In this project, the ankle strategy is used where in the human body above the ankle is considered rigid akin to an SIP and any disturbance or motion to this balanced body will be resisted to an extent by the stiffness of the ankle provided by the muscles and ligaments involved around the ankle as well as its mechanical structure (bones). Also, there is will be a physical limit to the flexion of the ankle, different when flexing upward (dorsiflexion) or downward(plantar flexion). A person falling forward would experience dorsiflexion and a person falling rearward would experience plantar flexion of the ankle. Due to the anatomy of the ankle dorsiflexion is mush smaller than plantar flexion resulting in lower balance recovery window than for the case of plantar flexion.

The ankle joint has certain stiffness which is dependant on its position i.e., flexion and it is neither a constant nor does it vary linearly. It is a nonlinear parameter dependent on various factors such as its inclination - given by ankle angle θ as shown in figure 4.1 , the torque induced in the joint etc. This ankle torque and and position dependent stiffness has been extensively studied and one such study gives a complex transfer function given in equation 3.2 [13] as below.

$$\frac{\tau(s)}{\theta(s)} = Is^2 + Bs + K \tag{3.2}$$

Where, s is the Laplace variable, τ is the the ankle torque, θ is the ankle angle, I is the angular moment of inertia of the foot, B is the angular viscosity of the ankle and K is the angular elasticity of the ankle

This is a function arrived at in a sophisticated study to determine the influence of ankle torque and ankle angle on the ankle stiffness. A equation inspired from this is used in this project to incorporate the dynamic stiffness parameter, which is as in 4.15 and is discussed ahead in the report.

3.1.3 Reaction time for muscle action

Stability of human posture is dependent on numerous active and passive mechanisms, joint's position, impedance and the control of the nervous system controlling them. This is inherent and mostly subconscious in all humans. The ankle has certain stiffness even when the person is in standing position and when there is a disturbance of minute magnitudes of sway the ankle stiffness reduces a bit in part to accommodate for some flexibility of the ankle to adjust balance to compensate for the sway as other muscles become active. When the ankle angle crosses a threshold, the stiffness of the ankle varies and becomes much higher to restrict the motion of the body from falling - a process that is both dependent on the anatomical restrictions as well the nervous system acting to counter the imbalance [21].

In the upcoming SIP model the ankle stiffness is taken as a constant of 50 Nm/deg [14] until there is an input disturbance which adds a torque onto the ankle and thus changing the ankle angle. When this happens the stiffness change can be taken from the 3.2 and in the SIP model the 4.15 is used. The threshold time until when the ankle stiffness remains constant or changes proportionally is the reaction time of the body to register imbalance intrinsically is the 'reaction time' that is incorporated in the SIP model. This parameter of time is discussed and arrived at by monitoring the volunteer test data.

3.2 Mathematical model

The above described inverted pendulum model of a human being needs to be mathematically described to study its behaviour. The equations of motions required to describe the inverted pendulum model can be arrived at by a few approaches, the classical/Newtonian approach, the Lagrangian approach and the Hamiltonian approach etc. The appropriate approach for the inverted pendulum problem was decided after an initial trial by defining the equations of motion in both the classical mechanics approach and the Lagrangian approach. The Newtonian approach directly addresses the equations of motion and its always advantageous to have some known forces in the system. It also requires to find accelerations in all 3 directions and equate the F=ma (where, F is the force acting on the system, m is the mass and a is the acceleration of the system) and solve for the constraint forces. The motion or equilibrium is determined from the scalar equations. It is helpful in solving problems that can be defined in Cartesian form and would be tedious to define the equations at every point of the system. This becomes highly inconvenient and thus Lagrangian mechanics was considered to solve the problem described in this project.

The Lagrangian approach uses energies of the system to derive the equations of motion. And the Lagrangian function is defined as L=T-V (where, L is the Lagrangian function, T is the kinetic energy and V is the potential energy of the system). The Lagrange's equation can be described using partial differential equations as,

$$\frac{d}{dt}\left(\frac{\partial \mathcal{L}}{\partial \dot{\theta}}\right) = \frac{\partial \mathcal{L}}{\partial \theta} \tag{3.3}$$

3.2.1 Feedback Control

To replicate the complexity and non-linearity of the muscle-skeletal system involved in human balance in the mathematical model, a control system is also needed to control the muscle torque properties. To make a more realistic influence of muscles in the model a feedback control loop is needed to adjust the rate and time of action of the extra torque. The concept of a proportional integral derivative (PID) controller is used to develop predictive and a proportional feedback loops for the model. The proportional part gain is set based on the error rate increase, based on the rate of change in the error of the model the gain which represents the muscle torque acting rate in the model is changed. In the predictive part the previous data are extrapolated to predict the future possible value of θ, based on this the decision to whether the gain value of the controller will be changed accordingly. The final value of the gain of the controller will be based on both the parts to give a more realistic representation of the muscle action in the model. This will be further used to calculate the ankle stiffness and thus the angle of tilt and the loop continues further. The construction and the parameters of the controller is further explained in the section 4.3.5.

3.3 System Identification and Verification

In this section the parameters that are observed within the model and the parameters that are used to validate the observations or results from the project are discussed and identified.

One of the parameters observed is the ankle angle as discussed above and is modelled into the SIP with certain degree of sophistication. A PID inspired control system is defined such that limits of ankle stiffness to predict points of balance loss, are put in place. The time at which this limit is reached based on the input motion of the sled as in volunteer tests is computed and tuned to match with the observed data from these tests.

Along with this an additional parameter is observed to understand loss of balance, the protrusion or longitudinal motion of the pelvis before the person takes corrective action to regain balance. From volunteer data it is observed that oftentimes due to the motion of the platform the torso of the human reacts late to the the impulse i.e., the ankle gets rotated and the torso remains stationary ever so slightly before the person reacts and takes corrective action. This is when the person tries to keep their knees relatively unbent. There is a chance that the person bends their knee and the pelvis remains very close to the original position and might not give a realistic value that can be used to define a loss of balance parameter for the model built in this project. Such data will is ignored for the time being to simplify the criteria for loss of balance.

For the system to identify point of imbalance a limit or a threshold has to be defined for the amount of tilt of the center of mass (CoM). Based on a study on support strategies for human balance [14] there is a clear cut threshold set for imbalance. Here we define a specific base of support (BoS) that will be the dimension of the ankle in our model and then the CoM. The model is said to be imbalanced when the protrusion of the CoM forward or backward exceeds the threshold of the base of support. In this study volunteer tests were conducted and it was decided that the within the optimum distance of about 0.23 meters for an acceleration of about 2.6 m/s^2 from CoM to the ankle in the horizontal direction, when the protrusion is forward is said to be that the body is balanced. the tilt angle corresponding to this distance based on the height of the CoM for a 50^{th} percentile male is about 10 degrees. Similar calculation is done for falling backwards and a limit of falling/balance is considered to be about 8 degrees. Though there is not accurate data on the limits for falling backwards the threshold is based on the uncertainty and behavioral assumption made for an individual. Such as, while falling backwards the person is more cautious so the decrease in the limit for imbalance.

4
Model

This chapter describes the mathematical model of the standing passenger built in MATLAB.The assumptions, numerical derivation and the implementation of the equations is explained in this chapter.

4.1 Assumptions

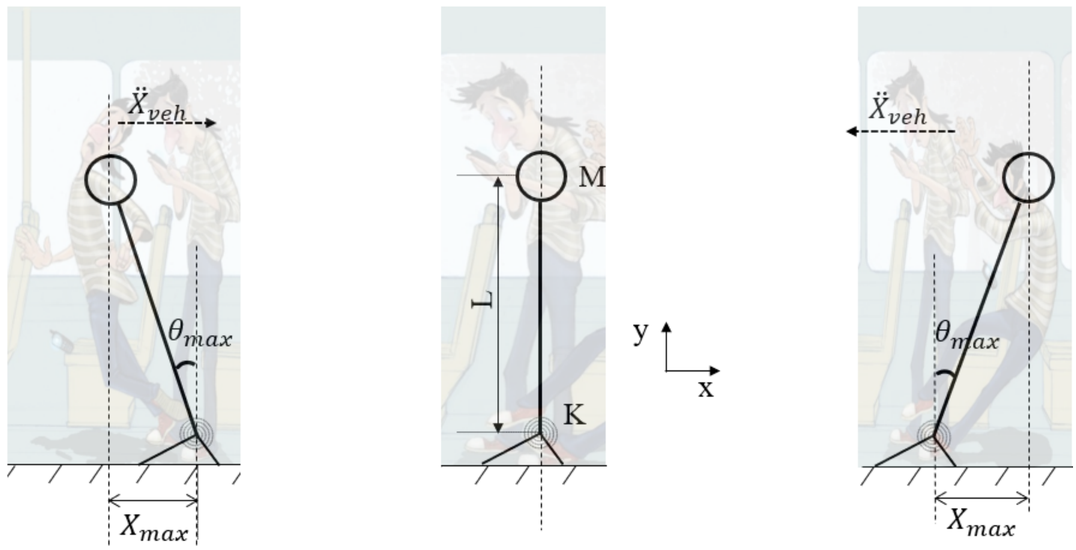

Figure 4.1: Single Inverted pendulum model in steady position and with acceleration $\ddot{X}$ acting in either directions.

The model passenger posture is represented using a Single Inverted Pendulum (SIP). The SIP setup for the standing passenger posture has certain assumptions in its working for this project. The model comparison and representation is as shown in figure 4.1. The entire mass of the human is assumed to be a point mass, M and is located at the center of mass (CoM) of the human body. In the SIP the distance of the CoM from the ground is taken as L. The link is assumed to be rigid with certain inertia. The ankle joint is assumed to have an torsion spring which provides resistance depending on the angle. The stiffness of the spring is denoted as K. X is the displacement of the foot which changes with respect to angle θ. The θ is a function of time t, hence the spring stiffness and the displacement also varies with time.

In the figure 4.1 the θ varies with respect to time and is influenced by the magnitude of the acceleration, $\ddot{X}$. The equations for this motion is derived in the following section which will be used to predict the unbalance in the system.

4.2 Manual Calculations

The mathematical representation for the change in angle is derived in this section. We use Lagrangian's energy relativity which is defined as,

$$\frac{d}{dt}\left(\frac{\partial\mathcal{L}}{\partial\dot{\theta}}\right) = \frac{\partial\mathcal{L}}{\partial\theta} \tag{4.1}$$

Here the $\mathcal{L}$ is the Lagrangian function which is given as ,

$$\mathcal{L} = E_k - E_p \tag{4.2}$$

E_K is the Kinetic energy and the E_p is the potential energy. In our model the Kinetic energy,E_K is influenced by the mass and the position of the CoM. The Potential energy,E_p changes by the stiffness of the torsion spring in the ankle and the height of the CoM. The equations are,

4.2.1 Kinetic Energy

$$E_k = \frac{1}{2}MV^2 + \frac{1}{2}I\omega^2$$

here the velocity V is defined as,

$$V^2 = v_x^2 + v_y^2$$

The components of velocity are,

$$v_x = \frac{d}{dt}(x) = \frac{d}{dt}(X - Lsin\theta) = \dot{X} - \left(Lcos\theta * \dot{\theta}\right)$$

$$v_y = \frac{d}{dt}(y) = \frac{d}{dt}(Lcos\theta) = -\left(Lsin\theta\dot{\theta}\right)$$

so V^2 is,

$$V^2 = \left(\dot{X}^2\right) + \left(L^2\dot{\theta}^2\right) - \left(2Lcos\theta\dot{\theta}\dot{X}\right)$$

Substituting the velocity for Kinetic energy,

$$E_K = \frac{1}{2}M\left[\dot{X}^2 + \left(L^2\dot{\theta}^2\right) - \left(2Lcos\theta\dot{\theta}\dot{X}\right)\right] + \frac{1}{2}\left[\frac{1}{12}ML^2\right]\dot{\theta}^2$$

$$E_K = \frac{1}{2}M\dot{X}^2 + \frac{13}{2}ML^2\dot{\theta}^2 - MLcos\theta\dot{\theta}\dot{X} \tag{4.3}$$

4.2.2 Potential Energy

The potential energy has two components, the CoM and the torsion spring so based on them both, the equation for potential energy,

$$E_p = (MgLcos\theta) + \frac{K\theta^2}{2} \tag{4.4}$$

4.2.3 Equate using Lagrangian equation

Using equation 4.2, 4.3 and 4.4 we get

$$\mathcal{L} = \frac{1}{2}M\dot{X}^2 + \frac{13}{2}ML^2\dot{\theta}^2 - MLcos\theta\dot{\theta}\dot{X} - MgLcos\theta - \frac{K\theta^2}{2} \tag{4.5}$$

partially differentiate 4.5 w.r.t $\dot{\theta}$

$$\frac{\partial \mathcal{L}}{\partial \dot{\theta}} = 13ML^2\dot{\theta} - MLcos\theta\dot{X}$$

differentiate this w.r.t time t

$$\frac{d}{dt}\left(\frac{\partial \mathcal{L}}{\partial \dot{\theta}}\right) = 13ML^2\ddot{\theta} + MLsin\theta\dot{\theta}\dot{X} - MLcos\theta\ddot{X} \tag{4.6}$$

partially differentiate 4.5 w.r.t θ

$$\frac{\partial \mathcal{L}}{\partial \theta} = MLsin\theta\dot{\theta}\dot{X} + MgLsin\theta - K\theta \tag{4.7}$$

Using the 4.1 equation and substituting the 4.6 and 4.7 into the equations and simplifying it,

$$13ML^2\ddot{\theta} - MLcos\theta\ddot{X} = MgLsin\theta - K\theta \tag{4.8}$$

partially differentiate 4.5 w.r.t $\dot{X}$

$$\frac{\partial \mathcal{L}}{\partial \dot{X}} = M\dot{X} - ML\dot{\theta}cos\theta$$

differentiate this w.r.t time t

$$\frac{d}{dt}\left(\frac{\partial \mathcal{L}}{\partial \dot{X}}\right) = M\ddot{X} - ML\ddot{\theta}cos\theta + ML\dot{\theta}^2sin\theta \tag{4.9}$$

partially differentiate 4.9 w.r.t θ

$$\frac{\partial \mathcal{L}}{\partial X} = 0 \tag{4.10}$$

Using the 4.1 equation and substituting the 4.9 and 4.10 into the equations and simplifying it,

$$M\ddot{X} - ML\ddot{\theta}cos\theta + ML\dot{\theta}^2sin\theta = 0 \tag{4.11}$$

In the above equation,

- θ represents the tilt angle in $[rad]$.
- g represents the acceleration due to gravity in $[m/s^2]$.
- L length of the link in i.e. the height of the CoM $[m]$.
- K represents the spring constant in $[Nm/rad]$
- M is the mass of the passenger in $[kg]$
- $\ddot{X}$ is the acceleration in $[m/s^2]$

4.2.4 Explicit Equations

To implement the equations without using symbolic solving method in MATLAB, explicitly equate $\dot{\theta}$ and $\ddot{\theta}$ with respect to the known input variables.
As seen in equation 4.10, X is a cyclic coordinate so using initial parameters,

$$M\dot{X} - ML\dot{\theta}cos\theta = constant$$

finding the value of constant in initial conditions, initially the values,$\dot{X} = 0, \dot{\theta} = 0$ and $\theta = 0$ by this we get the value $constant = 0$. so we get,

$$\dot{X} = L\dot{\theta}cos\theta$$

Using the conservation of energy and the the above equations we get the explicit equations for $\dot{\theta}$,

$$\dot{\theta} = \sqrt{\frac{MgL(1 - cos\theta) + \frac{K\theta^2}{2}}{ML^2\left(\frac{1}{2}cos^2\theta + \frac{13}{2} - cos\theta\right)}} \tag{4.12}$$

Solving the equations 4.8 and 4.11 to get the equation for $\ddot{\theta}$

$$\ddot{\theta} = \frac{\ddot{X} + L\dot{\theta}^2 sin\theta}{Lcos\theta} \tag{4.13}$$

4.2.5 θ calculation

Both equations 4.12 and 4.13 depend on θ. The value of θ practically does not vary linearly it has dependence on both the external acceleration on the center of mass and the ankle stiffness.
The final equation can be represented by substituting 4.12 and 4.13 in equation, 4.8.

$$13ML^2\left(\frac{\ddot{X} + L\left(\frac{MgL(1-cos\theta)+\frac{K\theta^2}{2}}{ML^2\left(\frac{1}{2}cos^2\theta+\frac{13}{2}-cos\theta\right)}\right)sin\theta}{Lcos\theta}\right) - MLcos\theta\ddot{X} - MgLsin\theta + K\theta = 0 \tag{4.14}$$

The above equation has only the θ as an unknown and using `dsolve` the instantaneous angular positions is found.

4.2.6 Torsional Spring stiffness

The spring stiffness is influenced by two kinds of toque,
1. Torque due to the center of mass position w.r.t time.
2. Torque due to muscle reactions.

The torque due to the instantaneous position of the center of mass is directly calculated by the angle variation and the muscle torques are controlled by the the feedback control gains by the controller.

4.3 MATLAB Implementation

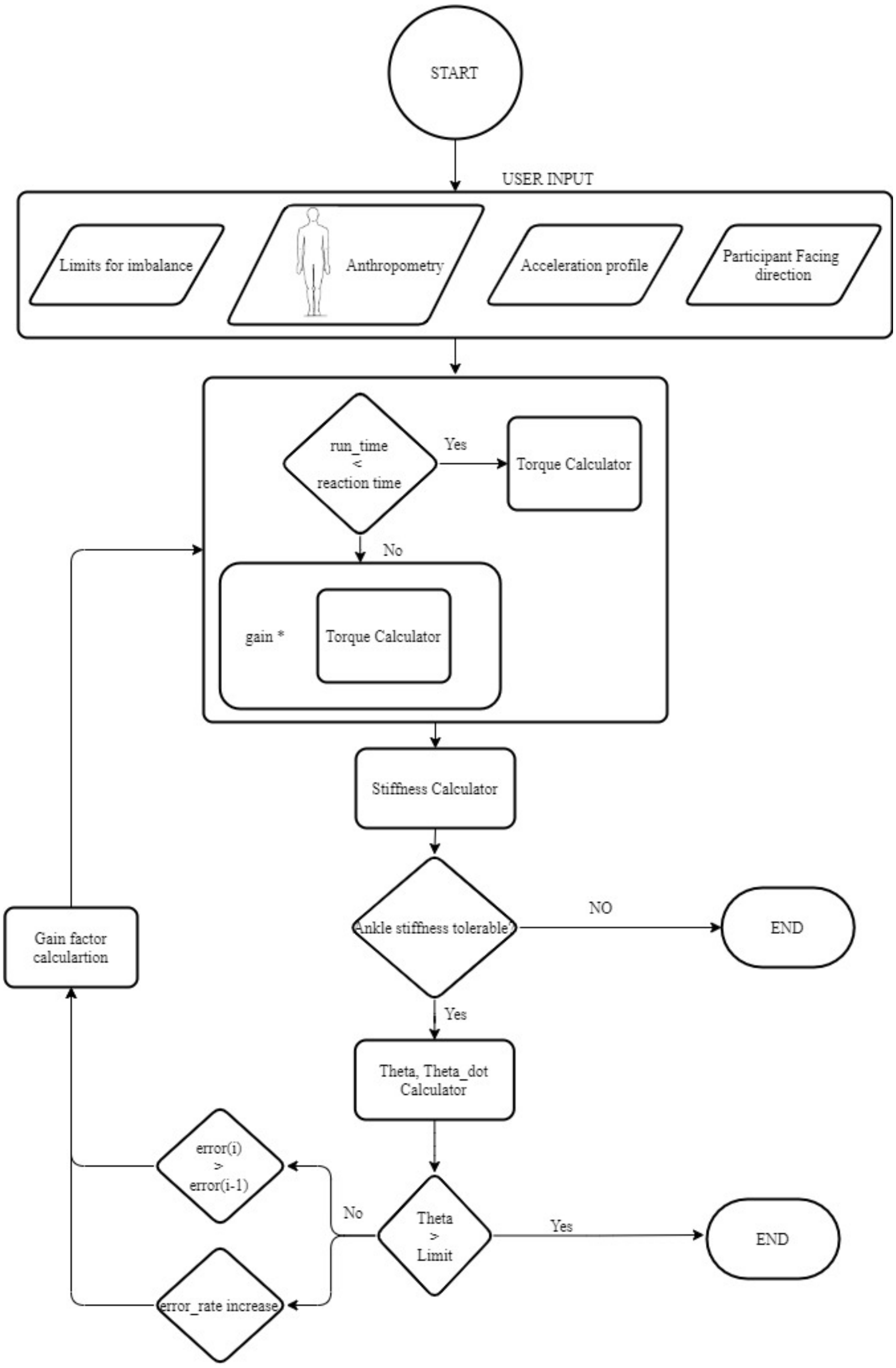

Figure 4.2: Flowchart representing the model for predicting imbalance

The above flow chart describes the flow structure of the program designed to predict the imbalance based on the criteria mentioned in system identification. The equations discussed in the earlier sections to predict the time and magnitude of acceleration for the imbalance are used. The main components of the program are explained as follows,

4.3.1 User Input

As represented in the flow chart in figure 4.2 the input for the program are,

1. Human body anthropometry used from section 3.1.1.
2. Acceleration profile from physical/volunteer testing.
3. Participant facing direction, Whether the participant during the physical testing is facing forward or backward w.r.t the acceleration direction.
4. Limits set for imbalance discussed in section 3.3.

This part is in the `main.m` in the final program file.

4.3.2 Torque calculator

As mentioned in the previous section two different kids of torques will be acting on the ankle. The basis of the torque acting will be depending on the reaction time of the volunteer. The reaction time is a factor which delays the time of the muscle torque to kick in which substantially increases the stiffness of the ankle giving more opposing force to fall. The factor for the percentage increase in the muscle forces in being controlled by the feedback loop. This operation is part of the `main.m` program.

4.3.3 Stiffness calculator

This part of the function `stiffness_calculator.m` which has the inputs $\theta, \dot{\theta}$ and Torque. The output for this program is the ankle stiffness. The stiffness of the ankle is found using the equation 3.2 but the term with the angular acceleration can be ignored for a spring-damper setup and the following equation is obtained,

$$\tau = B\dot{\theta} + K\theta \tag{4.15}$$

In the above equation τ is the torque, B is the coefficient of angular velocity and its value is found by pre-determined volunteer studies[13] which is available as linear regression data where the slope and the y-intercept($y = mx+c$) is available, so as a result the values for B is obtained. Value for θ and $\dot{\theta}$ is got from `theta_calculator.m`. In the first loop the theta values are not yet calculated so a initial assumption for stiffness is defined for when the model is at rest or the initial conditions.

4.3.4 Theta calculator

A function `theta_calculator.m` is built using equations 4.14 and 4.13. The inputs for this function is anthropometric data, acceleration profile and the ankle stiffness. The outputs are $\dot{\theta}$ and θ.

4.3.5 Feedback Loop

The feedback control is mainly set to implement realistic muscle torques acting after the reaction time. Two main kinds of control in used proportionality and the predictive.

4.3.5.1 Proportionality Control

The proportionality works based on a direct relation of the deflection angle with time. When the error i.e., the instantaneous angle subtracted by the initial or the preferred position. As the error increases a gain factor is introduced to compensate or to get to the original position. The further influence of this factor and its tuning is discussed in the results section.

4.3.5.2 Predictive Control

This works by using the `interp1` function in MATLAB to extrapolate the current data to find or predict whether the next data will be preferred or not. The gain value for the predictive is combined along with the proportionality to get the final gain value.

4.3.6 Program break identifier

The program break is identified based on the system identification section, section 3.3. When the Instantaneous value of the angular displacement reaches the set limits the loop breaks and the program ends. The time stamp when the program stops is saved and presented.

This MATLAB program is used for verification of the model working with the available volunteer test data and the results are presented and discussed in the next chapter.

5

Results

This chapter presents the results for the previously presented model. The model results are also compared with the volunteer test data and the effects of the change in results due to changes in the controller gain values are also presented in this chapter.

5.1 Simulation Parameters and overall Results

The results are compared with the volunteer test data and the anthropometric and the acceleration curves are mated to that with the physical test setup and the parameters are,

1. Length of CoM=0.98 m
2. Mass = 77.8 Kg

The acceleration profile used here are of two types,

1. Low profile Braking/Acceleration : acceleration profile with a gradual slope and relatively takes longer to reach the maximum magnitude for acceleration.
2. High profile Braking/Acceleration : acceleration profile with a abrupt slope and relatively takes less time to reach the maximum magnitude for acceleration, high jerk value.

The tests conducted physically are of the same kind so the results or the time of foot off can be compared. The main results from this are presented in the following Table 5.1.

Acceleration profile	**Facing Direction**	**Model predicted time [s]**	**Test Result [s]**	**Time difference [s]**
Low profile	Front	3.33	3.36	0.03
Low profile	Back	3.19	3.21	0.02
High profile	Front	0.390	0.420	0.03
High profile	Back	0.430	0.520	0.09

Table 5.1: Overall Results

Results for four cases are presented in the Table 5.1 and they have a similar trend of working so showing the elaborate results for volunteer facing in the same direction with two different acceleration profiles. The limits are set based on the section 3.3.

5.2 Low profile Braking- Forward facing

Using the aforementioned parameters the simulation is run for a low profile braking curve when the passenger is facing forward. The results for the case is presented in the following plots. Figure 5.1 refers to a low profile braking curve we see that to reach maximum magnitude of the acceleration some time is taken, this refers to a low profile curve. The two markers in the plot red represents the time-step captured by the program while the green represents the time obtained during volunteer tests. Figure 5.2 refers to an illustration on ho the pendulum moves with time.

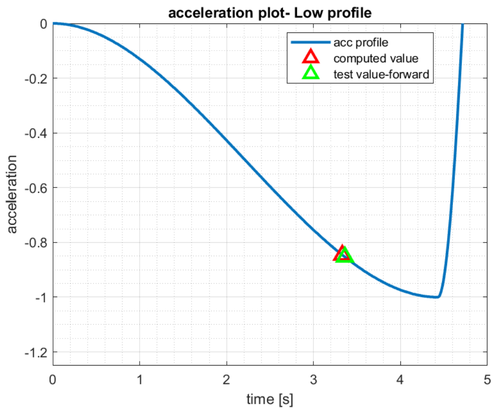

Figure 5.1: Low profile acceleration with the step off markers

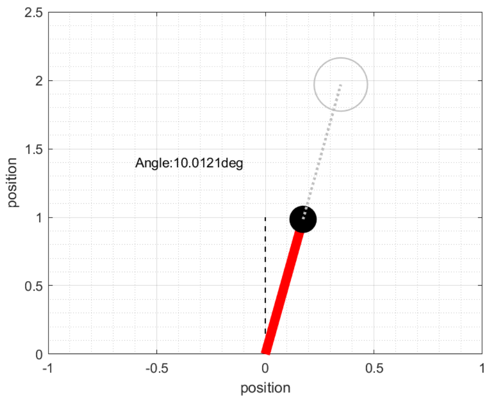

Figure 5.2: Final SIP motion illustration plot

Figure 5.3 refers to the change in angle plot. Data is plotted exactly till the point when the foot is off the ground. It is seen that the variation of the angle is non-linear with time and is represented in this plot. The plot is based on the equation 4.14.

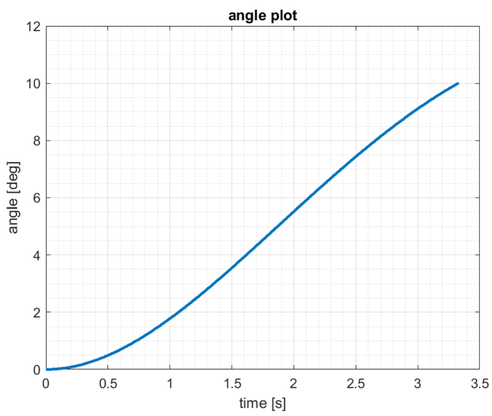

Figure 5.3: Change in angle with time plot

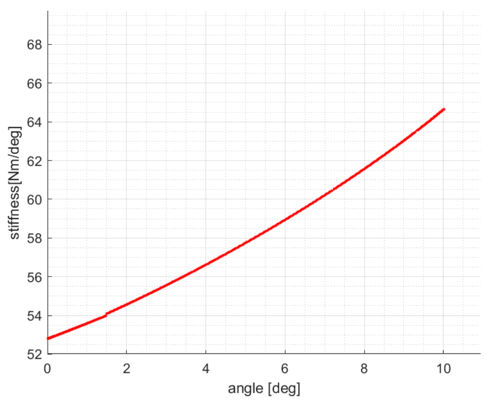

Figure 5.4: Stiffness change with angle plot

Figure 5.4 shows the change in the ankle stiffness with time and the plot is based on the equation 4.15. There is a slight step-up in the stiffness around 1.5 seconds, his is due to the reaction time and when there is a sudden surge in torque due to the muscle stating to act against falling.

5.3 High profile Braking- Forward facing

Similar simulation as the previous is run but with a different acceleration profile. Figure 5.5 shows the high profile acceleration plot, we see that the time taken to reach the peak magnitude is much less than the low profile case. The red marker in the plot represents the computed time-step by the developed mathematical model and the green marker represents the volunteer test result data.

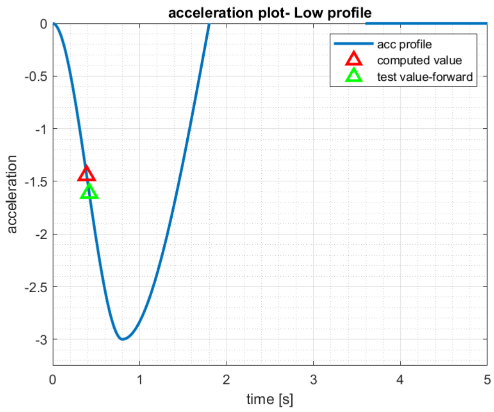

Figure 5.5: High profile acceleration with the step off markers

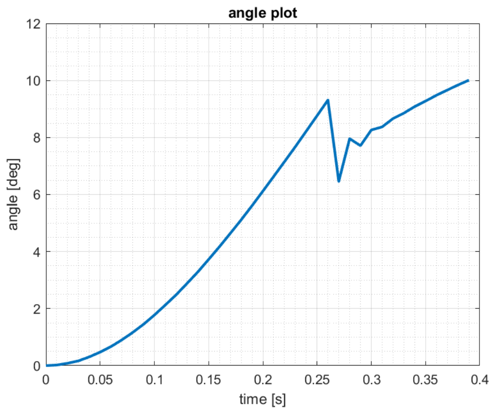

Figure 5.6: Change in angle with time plot

Figure 5.6 refers to the angle plot with time. It can be seen that there is a sudden change in the angle in the opposite direction. About 1.5-2 degree reflex change in the angle due to the sudden increase in the torque value due to the muscle starting to act to prevent the fall. This change in angle might seem to be very drastic but it is a degree change by the lower limb and as there is very less time for the body to react to the external force. Further illustration on this is done in the discussions chapter.

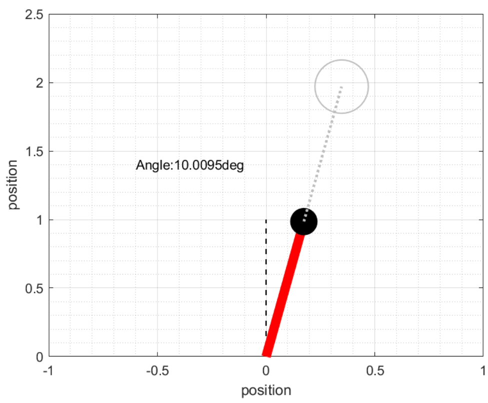

Figure 5.7: Final SIP motion illustration plot

Figure 5.7 Represents an illustrated plot on how the SIP acts on the used acceleration profile.

5.4 Gain Parameter Effect

The value of gain parameters affects the final results because the rate of muscle forces acting varies on the two occasions. The rate of the muscle force acting will be more when the magnitude of the acceleration is higher. There is no accurate relation between the two but due to this factor the tuning of the gain parameter is not accurate. When the same parameter or an average value cannot be used as the results will be dangerous in some situations, i.e. when the tuning is done for higher profile the stiffness curve ends up being very steep and that might cause the predicted time to be much more than the tested time for the lower profile acceleration curves. This affects the safety margin set and might lead to a misleading data to design safety systems.

5.4.1 High profile curve with the gain tuned for Low profile curve

The simulation is run for the high profile acceleration curve with the gain parameters tuned for the low profile curve and we get the following plots.

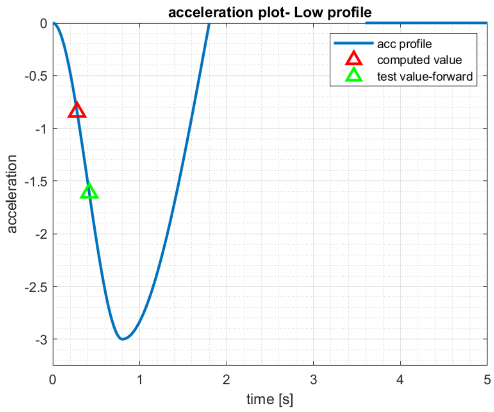

Figure 5.8: High profile acceleration with the step off markers

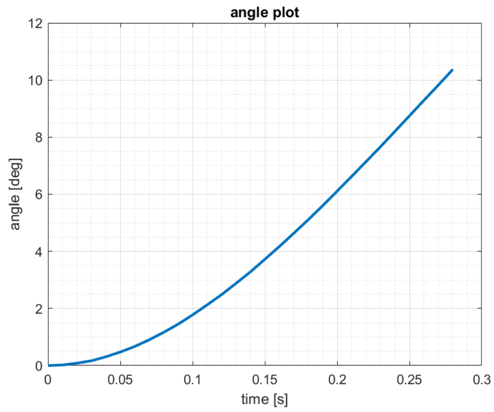

Figure 5.9: Change in angle with time plot

Figure 5.8 shows the difference between the model predicted time (Red marker) and the volunteer data (green marker) is greater than that of when the model was specifically tuned for high profile acceleration. As the smallest of the difference in the time will have a impact in the safety system design there needs to be specific tuning for different profiles in the designed model.

6
Discussion

A through study on posture of a standing human done both using a Single inverted pendulum (SIP) with a joint only in the ankle and a double inverted pendulum (DIP) with joints at both the ankle and the hip giving a more relative interactive motion variable [18]. Although the SIP model is is not representative as it ignores the ankle-hip coordination, it is functionally correct and practically acceptable for experimental studies that focus on the postural oscillations of the center of mass [18]. Also, an SIP successfully captures all the dynamics and the kinematics for the variables we have used in the project to compare with the volunteer results. The restricted degree of freedom resolves the issue of unnecessarily increasing the unknowns while using a DIP.

6.1 Effect due to Gain value

The gain value of the controller has some dependencies with the muscle torque acting on the model. The torque determines the stiffness of the ankle and hence it will have an effect in the angle change with time plot. It is seen that the imbalance time is sensitive to the gain value of the controller. From the results presented in the previous chapter, the gain value also is dependent on the following two factors,

6.1.1 Acceleration profile

By the two different acceleration profiles presented, the gain value is higher for the more abrupt profile of acceleration. Anatomically speaking, when there is a higher push to an individual the resisting muscle forces varies with the duration you get pushed. It is similar with the model as if the jerk value is higher then the rate of the muscle force acting will be higher hence the gain value will be higher for a low profile acceleration curve. Similarly when this is seen on a low profile acceleration curve the relative muscle acting rate will be less so in-turn the gain value will be relatively smaller.

6.1.2 Reaction Time

Reaction time also has an effect on the way the muscle acts. This value also is individual specific as the realisation time will be a sensory input for an individual and the outcome of this also depends on the surroundings of the person. For example if a person is told he will be pushed beforehand then involuntarily the body prepares itself to prevent fall due to this force. Same way for a lesser reaction time the the passenger is more safer and the time to imbalance due to this increases.

6.2 Effect due to the passenger facing direction

The result is related to the direction the passenger will be facing during the simulation. Generally when a person falls backward he/she is more careful as it is not on the sight of view and the surroundings are unknown. So the limit to which the person losses balance is less compared to facing forward. Also falling backward is less apparent and due to this the reaction time for this fall also might be more. In case passenger is falling forward the realisation of fall is more apparent and earlier also the limit to loose balance is more. The facing direction also changes the reaction time which ensures the start of the muscle torque acting.

6.3 Ankle Stiffness

Ankle stiffness is a important factor as it determines the angle of tilt of the passenger. it is seen that the stiffness is a non linear function, the equation for the curve 4.15 and the numerical value for stiffness is obtained. These values are also realistic as from studies in [19] the maximum achievable ankle toque possible by a human is about 200 Nm, as per the results the maximum ankle torque in the model reaches close to 172 Nm. By this it can be said that the model has given humanly possible torques to resist fall.

6.4 Safety margin

Safety margin is an important matter assumed in the consideration of the gain values for the model. The aim reason for keeping a safety margin is that theoretical results are not always practically right.
In Table 5.1 there is some difference in the predicted time by the model and the time by volunteer testing. In all the cases the model predicts a time before the actual event i.e. the safer side of the plot.
The reason the theoretical results deviates from the actual results is because of the failure of the assumptions made for the problem. some of the ones which might fail to give safety issues are,

1. This model does not have freedom in the perpendicular plane and vertical plane but in an actual scenario this might not be right.
2. Assuming the inertia of only the main body the upper limbs also have some inertia when the imbalance occurs.
3. surrounding environment affects the way the fall occurs like obstacles might aggravate the situation of fall.
4. Some corrective measures taken by the upper limbs might prevent or increase fall.

These are some of the possible reasons to introduce a safety margins as the prediction from this could be used directly used to design safety systems for passengers.

6.5 Model Limitations

6.5.1 Further Developments in the Model

Any more complexity is not necessary for the accuracy and precision of the results. An upgrade to the current model can be done introducing a second joint, maybe in the knee or in the hip to get a more realistic motion and by doing so the number of assumptions made can be reduced. The increase in the model complexity can certainly give better and more 'realistic' results but for the scope and the restriction in the project, the model developed currently is sufficient.

The model designed and simulated captures all the required data for the test and comparison but there are certain factors which are holding back the accuracy and precision of the results. These factors have an important influence in the output but the data for this is unavailable or inaccessible,

6.5.1.1 Joint relations

As discussed before DIP certainly is more representative of the human movements than SIP which has a lot of assumptions made to simulate. But the major drawback we faced to implement the DIP was the uncertainty of the joint location that is if the second joint was in the knee or at the hip. This ambiguous choice led to eliminate further progress in this direction. Also if for the use of DIP the constraints on each joint and also the relative motion between the joints represented by a mathematical equation was very complex due to the lack of sources in this topic.

6.5.1.2 Reaction time

The reaction time has an affect on the gain values of the controllers and hence with the output of the simulation. This parameter being non controllable by the test volunteer and it being a character of the individual, it is very difficult to generalise this variable in the program. Specific values can be used to run the simulation but the lack of usage of studies conducted on the reaction time of human nervous system made it difficult to incorporate an equation to generalise the model.

6.5.1.3 Ankle Torques and stiffness

Ankle torque curves are non-linear and have many factors influencing the values though practical tests are conducted on this topic these are age specific as the volunteers are restricted. Due to this more accurate torques could not be incorporated in the model, regardless of which the muscle and the center of mass torques were calculated to give acceptable results.
Ankle stiffness is also an important factor in the calculations in the equation 3.2 for the calculation but here the coefficients are still unknown. With limited source

[13] was used to find the coefficient of angular velocity. Due to lack of data, the coefficient of angular acceleration was ignored. As a result the stiffness is represented as $K(\theta, \dot{\theta})$ i.e. a spring damper system to compensate the muscle contributions. But with more data and literature, a better, efficient and detailed calculation can be done to improve the results.

Other than these factors, parameters like age which have a major influence on muscle properties, the limits of imbalance and the resistance torques produced. Age has an important relation with the control gain parameter, complexity in the model and not finding enough data on this limited its scope.

6.6 Further Applications

The model designed in this project has many applications in the vehicle safety domain. Firstly, there has been a lot of work on development of Human Body Models (HBM) to simulate complex crash scenarios, but the current models are not designed for the posture analysis criteria which is one of the major scenarios evaluated to assess the cause of injuries and accordingly develop safety systems for standing passengers in non crash events. With the model created in this project the working of and the forces acting on the passenger in a standing posture can be studied in more detail. The prediction of probable time of fall gives better data to help design and optimise the safety system to activate at a more accurate instant. The simple model developed here with acceptable prediction time for imbalance gives a better platform in the understanding the factors acting during the event. The model developed has a lot of applications in the development in complex HBM's and with sufficient amount of data the results can be further improved.

Further improvements of the model can be implemented to develop a system to control the drive line properties to reduce the amount of jerk and its duration to restrict the causes of imbalance. Based on the imbalance criteria the amount of tilt information, the passenger can be monitored and if any need for evasive action occurs, the best and a controlled approach can be utilized to have a smoother acceleration/deceleration profile to minimize imbalance of standing passengers. Incorporating this in public transport can reduce the present situation and reduce the number of injuries caused.

7
Conclusion

There are a myriad of complexities that determine human balance. All the contributing factors vary from person to person and across demographic. A model such as the one defined in this project takes into account various approximations and retains sophistication enough to predict with some level of accuracy, the loss of balance time which is corroborated with volunteer test data. These predictions need to be scaled to the most affected demographic i.e., the elderly and other groups which is a complex task taking into consideration that there is no volunteer data to corroborate the results of such scaling. A reason for this can be the lack of volunteer and/or real time data from incidents.

The effects of few of the major non-collision events are studied in this project. These events were modeled to a certain degree of sophistication based on volunteer test data which used inputs resembling possible scenarios of said events in real life. The level of sophistication used in this project was able to produce results that were satisfactory. To be able to directly implement the dynamic nature of a real life non-collision event would require modifying the current algorithm to be more robust and add more complex functions to process the raw input.

The results obtained in terms of timing the loss of balance were agreeable (with some tolerance) if not exact to the timing observed in the volunteer test data. The loss of balance criteria when using an SIP was drawn from various reliable and detailed experiments and studies. In this project there were assumptions made and simplified approach that was followed which could be accounted for the difference in the results from the test data.

Balance recovery strategies i.e., (re)actions by a person to regain balance can be studied along with the consequences of said actions using the model built in this project by implementing the results in a more complex human body model (HBMs). These advanced HBMs can be used to design safety features in the future that could mitigate injuries. Also, it is not the peak acceleration of the vehicle that causes the imbalance and consequent injury but the profile of the acceleration curve that induces it. Thus a vehicle control can be built around this predictive model to actively adjust the troublesome acceleration profiles instead of a flat limit on the peak acceleration.

8
Future Scope

The research of our supervisor Robert Thomson focuses on head/fall injuries of passengers in public transport and can benefit from this study. Old and current models for head/fall injures use a limp model falling over. The work done and the results of this project can help to determine the timing of active reactions of the model due to fall inducing inputs on public transportation vehicles. This then can lead to getting closer to the reality of human balancing behavior.

Further development of this model can be done to control the drive line attributes to decrease the possibility of imbalance in passengers by controlling the acceleration profiles of the vehicle.

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