Impact Force on Human Body resulted from Safety Harness during Slip-related Falls in Gait

BY

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THESIS

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This thesis is dedicated to my parents without whom it would never have been accomplished.

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SUMMARY

Real slip and fall reproduction in a lab environment is important to investigate the mechanisms behind slip-related falls as well to produce perturbation training for the reduction of future falls. To prevent an actual fall and any potential injuries to the subjects, a safety harness in combination with shock absorber ropes is essential during these experiments. Although a safety harness prevent the subjects from falling down on the ground, the faller does still take an impact when the harness deploys during falling. Sometimes this impact force could cause serious trauma to the affected area. For example, during a vertical fall, the impact force between the straps from the hips down to the crotch could be high; thus it can injury the tissues in the affected area. Further, the resulted impact force between body segments, such as at neck or between vertebras could lead to injuries at spinal cord too.

The purpose of this study was: (1) to examine the impact force between body segments resulted from safety harness during falling by computer simulation assisted by dynamic optimization; and (2) to analyze the possibility of injuries to human body bone, tissue, or ligament due to the impact force among healthy older adults.

Analyzing the results, a general trend was observed which indicated relatively higher impact forces in the thoracic-lumbar region as compared to neck or cervical spine region. Results also showed that the impact force would not cause any injury to either bone or soft tissue. However, for couple of subjects relatively higher forces were noted in the lower back or thoracic-lumbar region

1. INTRODUCTION

1.1 Introduction

"Falls, injurious or non-injurious have a significant health impact, often causing deterioration in mobility and performance of activities of daily living" [1]. "There is an increase in vulnerability to falls with age which possess a greater health threat to older adults" [2]. The statistics provided by the National Safety Council and the Bureau of Labor Statistics suggest that Slip-Falls are the second leading cause of accidental death and disability. And as per the study conducted by United States Occupation Safety and Health Administration, the total annual cost of all slip and fall injuries in the United States exceeded $60 billion in 2004. According to the Centers for Disease Control and Prevention (CDC), slip and fall accidents account for more injury deaths of elder Americans than any other form of injury. A Global Report on Fall Prevention indicates approximately one in three older people falls each year. In the United States alone in 2004, over 1.85 million elders were treated in the emergency room for fall-related injuries. The CDC estimates that 20% to 30% of people who experience a slip and fall will suffer moderate to severe injuries such as bruises, hip fractures, or head injuries. These injuries can inhibit mobility and hamper independent living. "There is no question that prevention of falls is a pressing problem that biomedical research must solve" [3].

"A better understanding of the mechanisms underlying slip-related falls will undoubtedly be a crucial step towards the development of such intervention strategies and eventually towards the prevention of such injuries and reduction of the cost resulting from the fall incidences" [4]. Slip occurs when there is too little friction or traction between the foot and the floor. During normal walking, two types of slips may occur. The first of these occurs as the heel of the forward foot contacts the walking surface. The front foot slips forward, and person falls backward. The second type of fall occurs when the rear foot slips backward. The force to move forward is on the sole of the rear foot. As the rear heal is lifted and the force moves forward to the front of the sole, the foot slips back and person falls [5].

Currently lots of ongoing research aims at understanding the processes associated with slip-induced falls, influence of the age, as well as the recovery responses. A wide array of research also focuses on developing perturbation training for fall prevention [2, 3, 4]. However, the reproducibility of real slip and fall in a lab environment is one of the key factors involved in understanding the mechanisms behind slip-related falls or in developing a fall prevention strategies. Most common method to replicate the real world slip induced falls scenario is the gait-slip experiments [4]. In such experiments, participants are subjected to slip by making them walk across a contaminated surface, or a motorized force plate but for subject's safety, a harness in combination with shock absorber ropes is used [4]. This safety harness prevents an actual fall and any potential injuries to the subjects by regulating deceleration when the end of the rope is reached. Typically a harness system distributes the impact force of a fall across as much as the body as possible, thus meaning no individual part of the body receives too much force. Although a safety harness reduces the force of a fall, the faller does still take an impact when the harness deploys. Sometimes this impact force could cause serious trauma to the affected area. The impact force is mainly exerted on the trunk during falling. It may incur injury to the spinal cord or neck which may result in serious effect like spinal cord injury. There have been instances where people have injured even with safety harness intact during fall. Most severe types of injuries occur from "jack-knifing" that results from what is called the "nose-to-toes" posture of the body being violently folded in half when the safety harness is only secured to the body at the waist with a belt [19, 20]. Other injuries that generally occur are: Impact Injuries, Blood Flow Restriction, Suspension Trauma, and Catastrophic Failure [20]. According to OHSA; “A harness if not designed or worn properly; can be useless and may increase your chances of serious injury” [19]. There is little research investigating the impact force exerted on the body by the harness during falling after a slip and the potential injuries to the body bones, soft tissues, or ligaments. It is still unclear how big the impact force is. It is also unknown whether the impact force causes any injury to human body. Because all the impact force between body segments are difficult to measure in vivo. Computer modeling and simulation has the capability to examine in detail and precisely the force between body segments. The derived forces could offer investigators required data to study if the impact force could cause any potential injury to the spinal cord. In addition, it could provide guidance in designing safety harness aiming at protecting subjects from body injuries during perturbation-based gait training. This study will also assist us in better understanding the body dynamic mechanical response to the impact stepped from actual falls.

Therefore, the purposes of this study would be: (1) to examine the impact force between upper body segments resulted from safety harness during falling by computer simulation assisted by dynamic optimization; and (2) to analyze the possibility of injuries to human body bone, tissue, or ligament due to the impact force among healthy older adults.

1.2 Organization of thesis

The thesis is organized as follows. Chapter 2 describes methodologies such as gait-slip experiments and computer simulations that have been used in this study. In first half of this chapter, it focuses on the laboratory instrumentation, experimental set-up and protocol essential for slip induced gait study. The latter half of this chapter focuses on computer simulation based on forward-dynamics. The analytical modeling of human body during impact, equations of motion governing the model, the inverse dynamic technique and the optimization routine required for successful forward-dynamics based computer simulation in this study are also discussed.

During an impact, the human body acts as a spring-damper-mass system [9]. In this slip-related fall study, the human head, upper trunk and lower trunk was represented by spring-damper-mass model. The equations of motion governing the developed biomechanical model were derived on the basis of Newton's second law. In order to solve these set of second order differential equations simultaneously, the equations were rewritten in the form of first order differential equations and were solved with MATLAB ODE solvers. From the experimental recordings, the movement data of all segments, the load cell force and the ground reaction force were applied as inputs to the forward-dynamics based computer simulation. A simulated annealing (SA) algorithm was used to perform the optimization routine. The objective of the optimization technique was to seek the best spring-damper coefficients so that the output kinematics from the forward-dynamics model could closely replicate each subject's measured body kinematics.

The results for this study are shown in chapter 3 and are grouped under: i) Forces acting on the model, ii) Simulation and Optimization, iii) Intersegment contact forces, and iv) Human Injury Criteria. The first section shows the forces acting on model: load-cell force and Ground Reaction Force. Plots for the forces acting on the model are shown at each time instance for one of the subject. In the subsequent section, the closeness of the fit of the optimal simulated kinematics and experimental data is shown to indicate that our method i.e. forward simulation and dynamic optimization could precisely reproduce the experimental kinematics. In the third section, the impact force between each segments were analyzed during slip induced fall trials and during normal walking trials for 15 subjects. The final section of chapter 3 shows human injury tolerance for neck/cervical spine and thoracic-lumbar spine region.

Discussions regarding this study are presented under chapter 4.

2. METHODOLOGY

As aforementioned, both gait-slip experiment and computer simulation techniques were used in this project to examine the inter-segment impact force during falling resulted from slip in gait. The gait-slip experiment provided the basic full body kinematics and kinetics during falling. The computer simulation would calculate the contact force between each pair body segments due to the difficulty to directly measure them.

2.1 Gait Slip Experiments

All the gait experiments for this study were conducted in Clinical Gait and Movement Analysis Laboratory; University of Illinois at Chicago. The data collection area included a lowered, 10 x 3 meter instrumented walkway with ramp access. Kinematics and the ground reaction forces (GRF) were gathered using an eight camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) and four synchronized force-plates (AMTI, Newton, MA), respectively [6]. A set of twenty eight light-reflective markers were used for this experiment. Specifically, markers were affixed at vertex, ears, rear neck (the spinous process of the seventh cervical vertebra), shoulders (the acromion of the scapulae), midpoint of the right scapula, elbows (the lateral humeral epicondyles), wrists (the radial styloid processes), sacrum, greater trochanters, mid-thighs, knees (the lateral femoral epicondyles), mid-legs (the tibial tubercles), ankles (the lateral malleoli), heels (calcaneal tuberosities), and the fifth metatarsal heads [6].

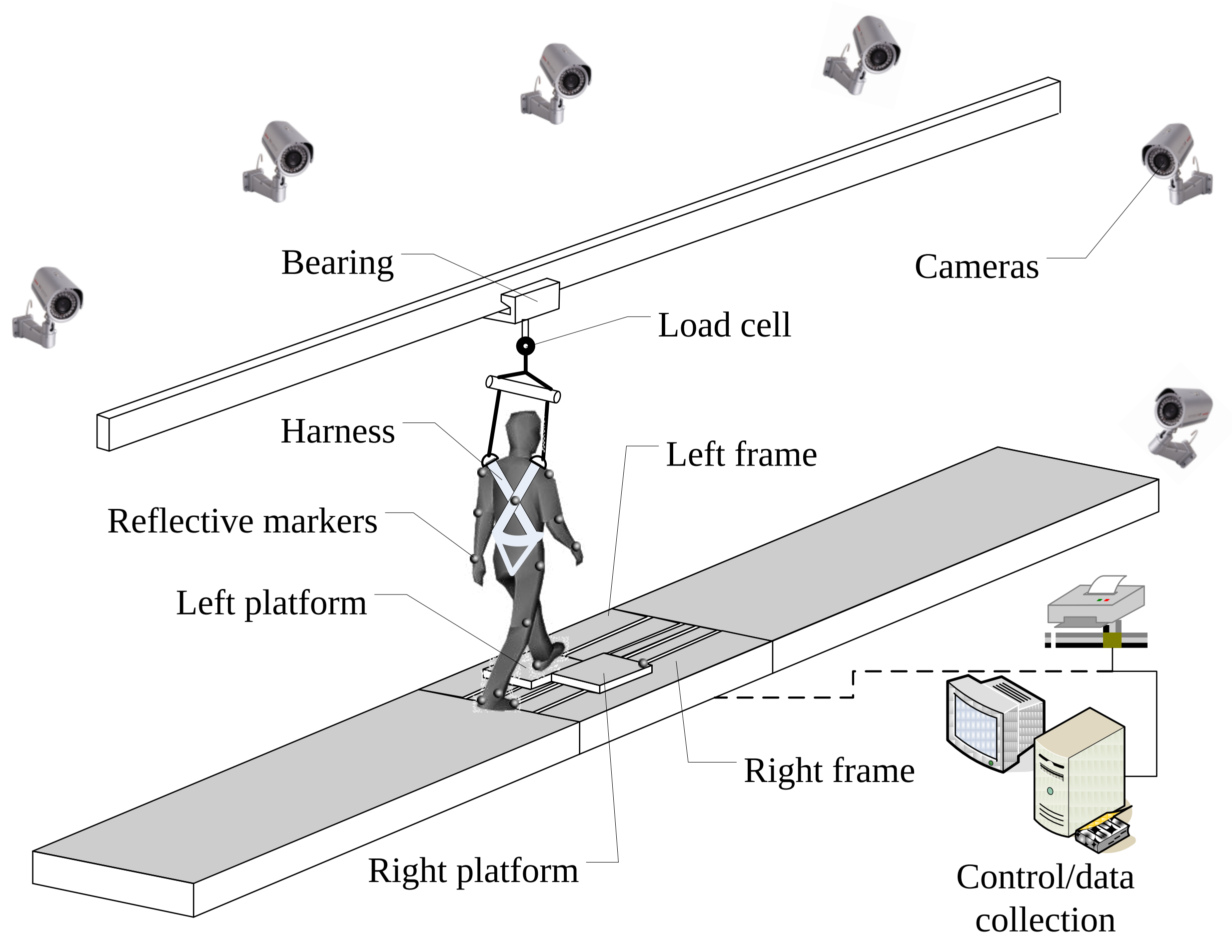


Figure 1: Schematic illustration of the experimental setup for inducing unannounced slips in gait. A slip is induced by releasing two low-friction movable platforms. Each of the two moveable platforms is mounted on a frame with four linear bearings, and the frame was bolted to two force plates to measure the ground reaction force. The low-profile movable platforms (and the force plates beneath, not shown here) were embedded in a 7-m walkway with decoy platforms (not shown) to reduce its visibility. The right- and left-side moveable platforms can be unlocked electronically after the landing of the corresponding foot. A set of 28 light-reflective markers were placed on bilateral upper and lower extremities, torso, and platforms. Their spatial positions were captured by an 8-camera motion capture system. The subjects were required to wear a safety harness which is individually adjusted to prevent a fall to the ground. A load cell was used to measure the force exerted on the harness [8].

Fifteen healthy older subjects who fell in response to an unannounced slip in gait were randomly chosen from an existing subject's pool. As per the study conducted, all these subjects participated in the study after being screened for exclusionary factors such as neurological (e.g. stroke, Parkinson's disease, spinal cord injury), musculoskeletal (e.g. osteoarthritis, rheumatoid arthritis, fractures onset ≤ 6 months), cardiopulmonary (e.g. angina, emphysema, lung cancer), other systemic disorders (e.g. diabetes), and selected drug usage (e.g. sedatives, anti-anxiety, anti-histamines). Also, prior to participation, all subjects gave informed consent as approved by the Institutional Review Board [1].

Subjects were informed that they would be performing normal walking initially and would experience simulated slip later, but no information was given as to where, when, or how the slip would occur. A full-body harness, attached by shock-absorbing ropes at the shoulders and waist to a low-friction linear bearing moving along a ceiling-mounted track, was employed for subjects' protection while imposing negligible resistance or constraint to their movement. The ropes were adjusted for each subject so that should they fall and suspend from the track after slip occurrence, their palms, knees, and buttocks would not contact the walking surface. After adjustments of the rope lengths, every subject was asked to perform a standard sitting-in-harness trial for 8 sec to ensure their safety. A load cell connecting the rope of the bearing was used to measure the force exerted on the person. The experimenter would adjust each subject's starting position so that his or her future slipping (right) foot would land entirely on the movable plate at touchdown. All subjects were able to take at least three steps before stepping on the movable platform. Unannounced slips were induced as subjects walked along a 7-m long walkway in which a sliding device was embedded. The sliding device consisted of a side-by-side pair of low-friction, passively movable platforms each mounted on a supporting metal frame with four linear bearings. The platforms were free to independently slide ≥ 0.75m forward upon computer controlled release of their locking mechanisms. Each metal frame was supported by two individual force plates (via two hinges in order to measure ground reaction forces to determine initial foot contact of a step). In the slipping trials, the right platform was released when the right foot contacted the movable platform; then the left platform was triggered when left foot touched it to perturb the recovery (left) step [1, 4].

2.2 Data Collection and Analysis

The collected marker displacement data were low-pass filtered at marker-specific optimal cutoff frequencies using a recursive second-order Butterworth Filter. Force plate and harness load cell data and trigger-release onset signals were collected at 600 Hz using a 64-channel, 16-bit A/D converter. The GRF and motion data were time synchronized at the time of data collection [1, 7].

Head, upper trunk, and pelvis movements were described using respectively the trajectories of the midpoint between the head markers (head level), of the shoulder markers, and of the pelvis markers. For each slip, three essential events were identified. They were slipping (right) foot touchdown (R-TD), the recovery (left) foot lift off (L-LO), and the instant of the recovery foot touchdown (L-TD). Slip outcomes were classified as falls when the peak force exerted on the load cell exceeded 30% body weight and were unambiguously confirmed via visual inspection of recorded video [8].

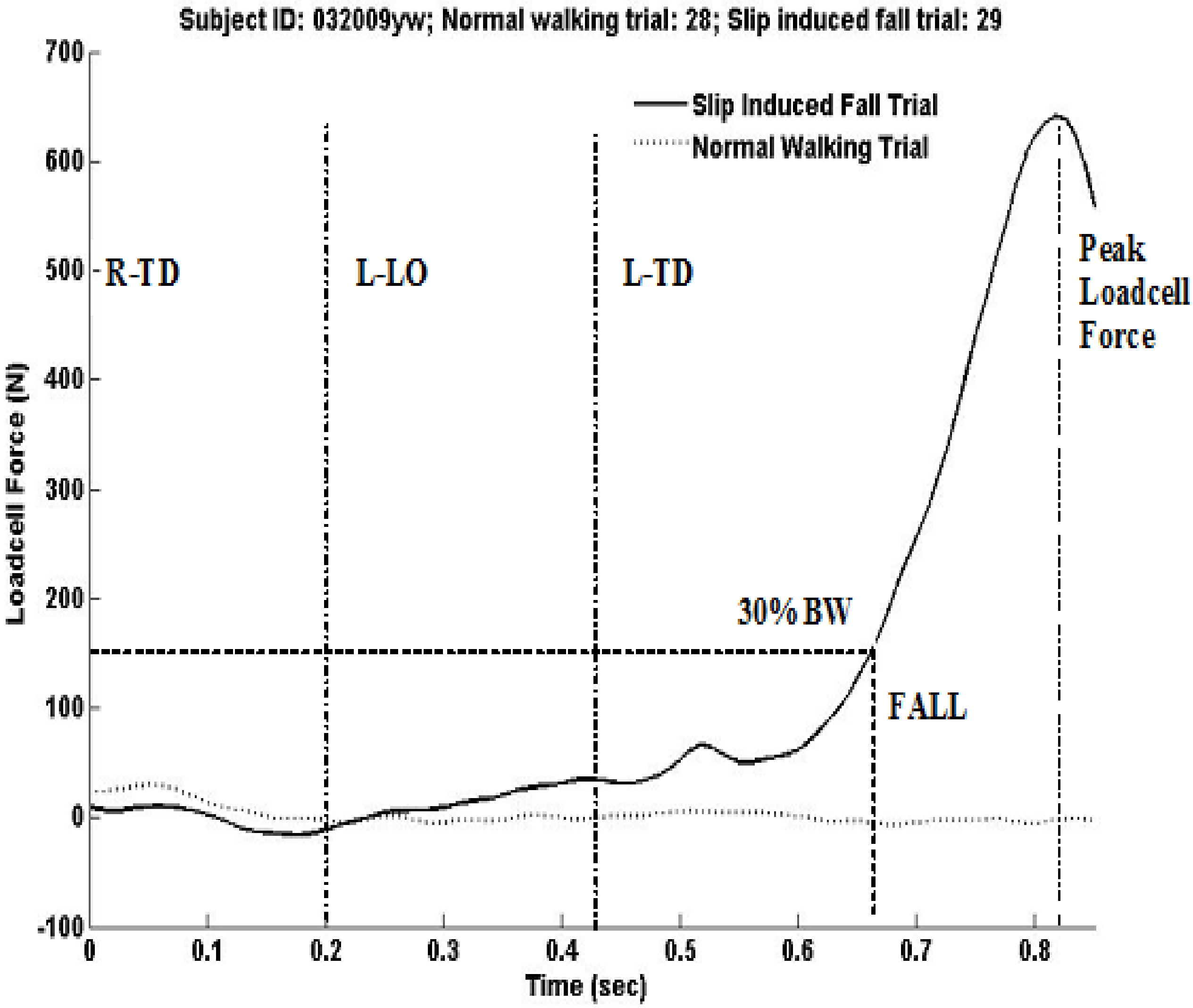


Figure 2: The force recorded by the load-cell, present in the over-head-harness system is plotted for every time instance of a normal walking and slip induced fall trial and is shown in figure above for one of the subject. The sudden increase in the force at a particular time instance indicates a fall experienced by the subject. Time t = 0sec indicates the slip onset period. Falling ended at 0.851 sec for this subject. During a slip trial, three events were identified namely slipping (right) foot touchdown (R-TD), the recovery (left) foot lift off (L-LO), and the recovery foot touchdown (L-TD).

2.3 Computer Simulation

In this study, computer simulation based on forward dynamics has been used to investigate human movement mechanisms. "In a forward dynamic simulation, differential equations of motion are numerically integrated forward in time subject to gravity, inertial and velocity-dependent effects, and muscle forces. It is a 'forward' simulation in the sense that forces produce motions and is distinct from inverse dynamic analyses in which internal (muscular) moments are computed from measured motions and external forces. One advantage of solving for motion through numerical integration of equations of motion rather than applying conditions of equilibrium in a static or quasi-static formulation is that there is no theoretical limitation on the number of degrees of freedom or the number of unknown forces that must be determined. If state equations can be written that describe the multibody dynamics of the body segments and joints as well as the computation of forces applied to those segments, then those equations can be used to predict positions and velocities going forward from some initial state" [9].

The reason forward dynamics based computer simulation is better suited over inverse dynamic technique in slip-related fall studies is aptly described by Dr. Feng Yang and Dr. Clive Pai as "The inverse-dynamics approach does not directly quantify the causal relationship between the inter-segmental forces and the motion state (i.e., the position and velocity) of body segments. Further, the inverse dynamics approach is unable to assess, control, or eliminate the confounding impact of the kinematic variability, which is common in empirical sampling, on the inversely derived forces. Even small inter subject differences in the initial posture (or body segment motion state) could yield large subsequent differences in inter-segmental forces. Moreover, the derived forces from the traditional inverse-dynamics approach often cannot reproduce the original body segment movement, when such derived forces and forward dynamics are used to drive the same individualized human model. The existence of such discrepancy, which likely results from inherent measurement errors, not only raises doubts on the validity of this approach, but also presents challenge for the simulation" [6].

2.3.1 <u>Human Upper Body Model:</u>

A modeling study is a useful adjunct to experimental studies, and may provide an insight view of the influence of body masses through simulations with unchanged springs and dampers of the system [10]. The limitations of experimental study and thereby the need for mass-spring-damper models in the study of human locomotion is very well summarized by Dr. Ali Asadi Nikooyan and Dr. Amir Abbas Zadpoor in their article: "Mass-spring-damper modeling of the human body to study running and hopping: an overview" and is as follows: "Although experimental techniques are valuable in understanding many aspects of locomotion, they have some limitations. First, not every quantity can be easily measured. For example, there is no non-invasive easy way of measuring muscle forces *in-vivo*. Second, experiments cannot be used to study the isolated effect of parameters. For instance, it is hardly possible to study experimentally the independent effects of body mass distribution on locomotion, because one cannot change body mass distribution while keeping the other parameters constant. It is much easier to run parametric studies using a model. Third, experiments may need access to a large number of participants and to specific experimenting conditions" [11].

Human upper body during the impact can be considered as a mechanical system with masses and connecting springs and dampers [10]. A spring-damper-mass model used in this study consisted of three rigid masses namely the head (m_0), the upper-trunk (m_1) and the lower-trunk (m_2). Spring-damper systems (c_1, k_1, c_2, k_2) were used to connect the three masses of the model. The masses represent the inertia properties of the different segments of the human body while springs and dampers represent the mechanical properties of the different segments including bones, muscles, tendons, and ligaments [11]. The model is as shown in the figure 3. The model is passive i.e. it does not include the effects of any active elements such as muscle, environmental factors such as stiffness of the ground or the type of footwear. In figure 3, the diagram on the left represents the front view of the model and that on the right represents the side view of the model. The Z-axis is oriented vertically. The Y-axis is in the plane of progression of gait. The X-axis is orthogonal to the plane of YZ. Movement trajectories of each segment were represented by Z0, Z1, Z2 in the vertical direction and Y0, Y1, Y2 in the horizontal direction. The forces acting on the model are the load-cell force and the ground reaction force. The load-cell force is the force acting on the body due to deployment of the harness during the fall. F_0 and F_1 are the load-cell force acting on the shoulder and waist respectively. The other forces acting on the model are the forces (F_2) exerted on lower back from both thighs.

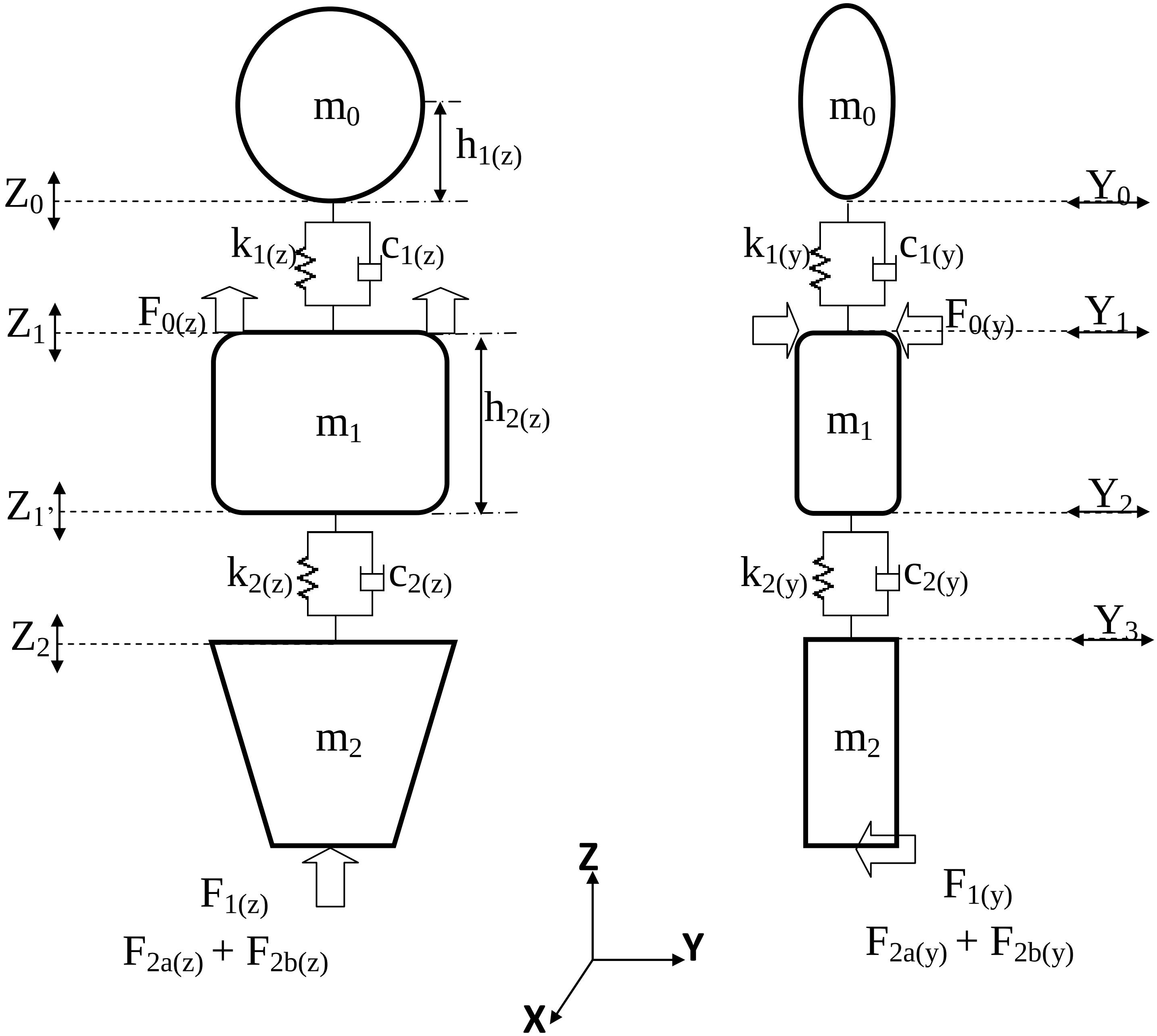


Figure 3: Spring-damper-mass model used to calculate the impact force when a subject falls under the protection of a safety harness during a slip-related fall induced in gait. The figure on the left represents the front view of the model and that on the right represents the side view of the model. The 3 segments in the model are the head, upper trunk and lower trunk from top to bottom respectively. The Z-axis is oriented vertically. The Y-axis is in the plane of progression of gait. The X-axis is orthogonal to the plane of YZ.

2.3.2 <u>Equations of Motion:</u>

The equations of motion governing the model, as showed in Figure 3, are as follows:

$$
\left.
\begin{aligned}
&m_0\ddot{Z}_0 = -m_0 g - k_{1(z)}(Z_0 - Z_1 - h_{1(z)}) - c_{1(z)}(\dot{Z}_0 - Z_1)\\
&m_1\ddot{Z}_1 = -m_1 g + F_{0(z)} + k_{1(z)}(Z_0 - Z_1 - h_{1(z)}) + c_{1(z)}(\dot{Z}_0 - \dot{Z}_1) - k_{2(z)}(Z_1' - Z_2 - h_{2(z)}) - c_{2(z)}(\dot{Z}_1 - \dot{Z}_2)\\
&m_2\ddot{Z}_2 = -m_2 g + F_{1(z)} + F_{2a(z)} + F_{2b(z)} + k_{2(z)}(Z_1' - Z_2 - h_{2(z)}) + c_{2(z)}(\dot{Z}_1 - \dot{Z}_2)\\
&m_0\ddot{Y}_0 = - k_{1(y)}(Y_0 - Y_1 - h_{1(y)}) - c_{1(y)}(\dot{Y}_0 - \dot{Y}_1)\\
&m_1\ddot{Y}_1 = F_{0(y)} + k_{1(y)}(Y_0 - Y_1 - h_{1(y)}) + c_{1(y)}(\dot{Y}_0 - \dot{Y}_1) - k_{2(y)}(Y_1' - Y_2 - h_{2(y)}) - c_{2(y)}(\dot{Y}_1 - \dot{Y}_2)\\
&m_2\ddot{Y}_2 = F_{1(y)} + F_{2a(y)} + F_{2b(y)} + k_{2(y)}(Y_1' - Y_2 - h_{2(y)}) + c_{2(y)}(\dot{Y}_1 - \dot{Y}_2)\\
&F_{\text{Load-cell}} = \text{sqrt}\,(F_{0(z)}^2 + F_{1(z)}^2 + F_{0(y)}^2 + F_{1(y)}^2)
\end{aligned}
\right\} \quad (1)
$$

where; m_0, m_1 and m_2 are the segmental masses for head, upper trunk and lower trunk respectively. [$F_{0(Z)}$; $F_{0(y)}$] and [$F_{1(Z)}$; $F_{1(y)}$] are the load cell force acting at the shoulder and waist respectively. [$F_{2a(Z)}$; $F_{2a(y)}$] and [$F_{2b(Z)}$; $F_{2b(y)}$] are the ground reaction forces acting on right and left hip respectively. [Z_0; Y_0], [Z_1; Y_1], [Z_2; Y_2] are the movement trajectories of the head, upper trunk and lower trunk respectively. [$c_{1(Z)}$; $c_{1(y)}$], [$k_{1(Z)}$; $k_{1(y)}$], and [$c_{2(Z)}$; $c_{2(y)}$], [$k_{2(Z)}$; $k_{2(y)}$] are the spring/damper coefficients of the model between head-trunk and upper trunk – lower trunk respectively.

The equations of motion governing the developed biomechanical model were derived on the basis of Newton's second law. These equations are second order differential equations. In order to solve these set of second order differential equations simultaneously, the equations were rewritten in the form of first order differential equations and were solved using MATLAB ODE solvers. As there are more unknowns than the set of equations, it is an under-determined system. Thus it has an infinitely many solutions. In order to obtain the optimal solution from a set of possible solutions, we use forward dynamic based computer simulation assisted by optimization.

2.3.3 Determination of Modeling Parameters:

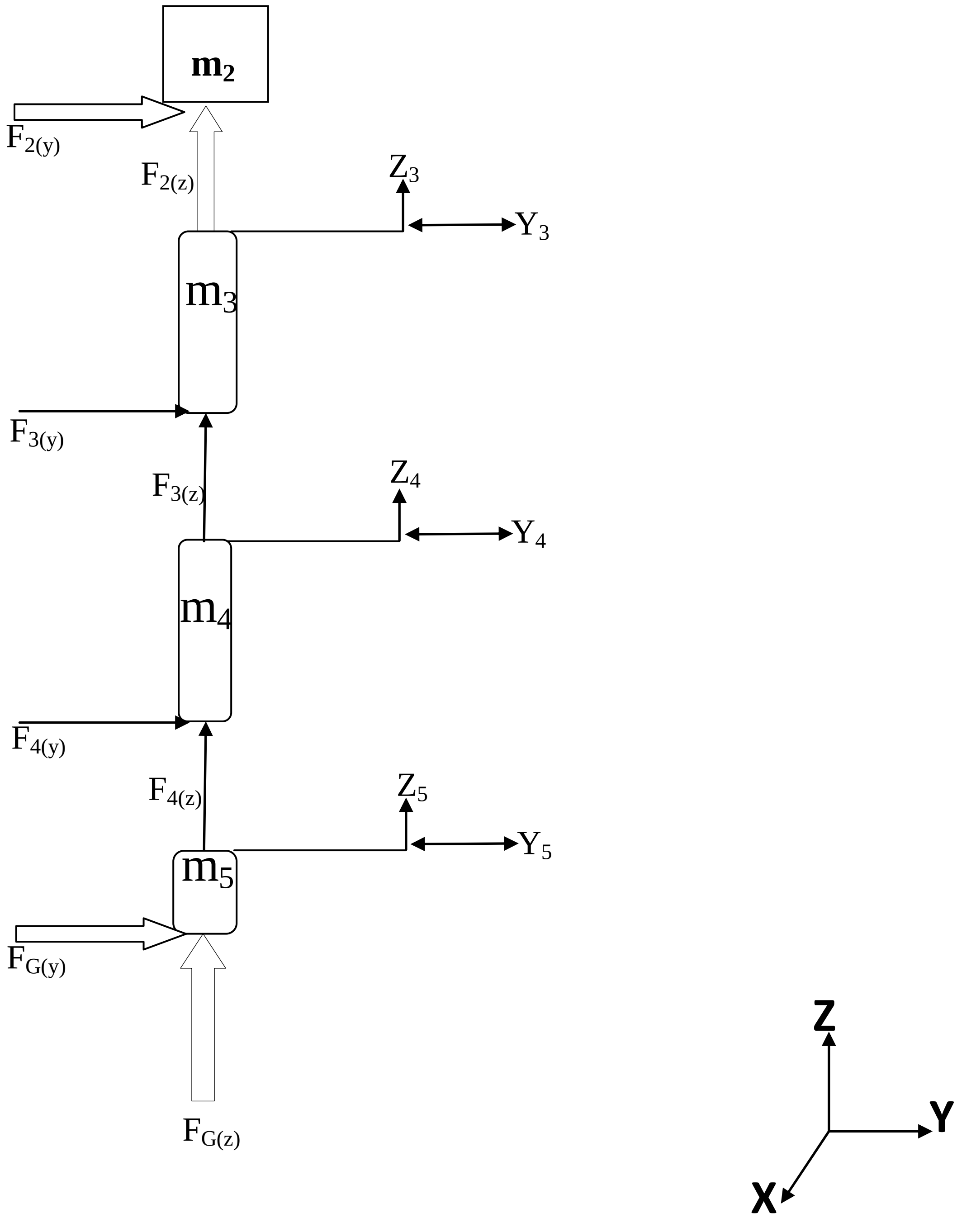
All the parameters which form the input to forward dynamics computer simulation are either obtained or calculated from experimental recordings. Step-by-step procedure for determining the modeling parameters are as shown below:

a. The masses (m_0, m_1 and m_2) and ($h_{1(z)}$; $h_{1(y)}$; $h_{2(z)}$ and $h_{2(y)}$) are calculated from the collected anthropometric data for each individual subject. Thereby allowing each subject's individual biomechanical model to simulate and optimize.

b. Movement data of head, upper-trunk and lower-trunk segments ($Z_{0,1,2}$; $Y_{0,1,2}$) are obtained from the experimental recordings.

c. Load-cell force is the force measured by the over-head-harness systems, equipped with load cell sensors. The time histories of the load cell force can be known experimentally. Typical load-cell force curve for slip-induced gait fall is shown in the Figure 2. The load-cell force measured is an overall force without knowing its direction and distribution. The sudden increase in the force at a particular time instance indicates a fall experienced by the subject.

d. Contact forces acting on the lower trunk from both thighs ($F_{2(z)}$; $F_{2(y)}$) is the result of ground reaction force and is computed using inverse dynamics technique. The model used to compute forces using inverse dynamic is as shown in the Figure 4.



<u>Figure 4</u>: Inverse Dynamic Technique used to calculate the forces acting on lower trunk due to ground reaction force through foot. The Z-axis is oriented vertically. The Y-axis is in the plane of progression of gait. The X-axis is orthogonal to the plane of YZ.

In figure 4, masses m_2, m_3, m_4 and m_5 are representing the lower-trunk, thigh, leg and foot respectively. These masses are calculated anthropometrically. [Z_3; Y_3], [Z_4; Y_4], [Z_5; Y_5] are the movement trajectories of the thigh, leg and foot respectively. These movement trajectories are obtained from the experimental recording of the kinematic data. [$F_{G(z)}$; $F_{G(y)}$] is the ground reaction force acting on the foot. The ground reaction force is recorded experimentally from force platforms. [$F_{2(z)}$; $F_{2(y)}$], [$F_{3(z)}$; $F_{3(y)}$] and [$F_{4(z)}$; $F_{4(y)}$] are the forces due to ground reaction force. These forces are computed using the following equations:

$$\left.\begin{aligned}
m_5\ddot{Z}_5 &= - m_5 g + F_{G(z)} + F_{4(z)} \\
m_4\ddot{Z}_4 &= - m_4 g - F_{4(z)} + F_{3(z)} \\
m_3\ddot{Z}_3 &= - m_3 g - F_{3(z)} + F_{2(z)} \\
\\
m_5\ddot{Y}_5 &= F_{G(y)} + F_{4(y)} \\
m_4\ddot{Y}_4 &= - F_{4(y)} + F_{3(y)} \\
m_3\ddot{Y}_3 &= - F_{3(y)} + F_{2(y)}
\end{aligned}\right\} \quad (2)$$

Using the above set of equations (2), forces acting on the lower trunk due to right or left leg ($F_{2(z)}$; $F_{2(y)}$) can be calculated. It is important to note that the vertical forces combine the effect of the gravity and the effects of the body's movements and acceleration in the vertical direction while the horizontal forces represent only the effects of the body's movements and acceleration in the horizontal direction.

2.4 Optimization

In the forward dynamics method, forces applied to the segments are used as the inputs to calculate the corresponding body motions. However determining a set of forces that produce a desired movement is one of the major challenges in creating a forward dynamic simulation. One approach is to use dynamic optimization to determine a set of muscle excitations that generate a simulation that best reproduces experimental data. Using this approach, the optimization objective function is typically a global measure of the error between measured and simulated biomechanical quantities [12]. In this study, Simulated Annealing technique was used to optimize the values for spring-damper coefficients and the load-cell force distribution factors which served as the control variables to minimize the difference between the experimental data and the calculated values for the movement trajectories of the head, upper trunk and lower trunk [6].

2.4.1 Simulated Annealing Algorithm:

The simulated annealing algorithm is a way of finding optimum solutions to problems which have a large set of possible solutions, in an analogous fashion to the physical annealing of solids to attain minimum internal energy states [13]. Simulated annealing (SA) is a method for solving unconstrained and bound-constrained optimization problems. The method models the physical process of heating a material and then slowly lowering the temperature to decrease defects, thus minimizing the system energy [14]. The basic idea is to generate a path through the solution space, from one solution to another nearby solution, leading ultimately to the optimum solution. In generating this path, solutions are chosen from the locality of the preceding solution by a probabilistic function of the improvement gained by this move. So, steps are not strictly required to produce improved solutions, but each step has a certain probability of leading to improvement; at the start all steps all equally likely, but as the algorithm progresses, the tolerance for solutions worse than the current one decreases, eventually to the point where only improvements are accepted. In this way the algorithm can attain the optimum solution without becoming trapped in local optima [13].

2.4.2 Optimization Implementation:

The objective of the optimization routine was to seek the best spring-damper coefficients and load-cell distribution factors so that the output kinematics from the forward-dynamics model could closely replicate each subject's measured body kinematics. Following steps were computed to reach the optimized value for spring-damper coefficients:

- Make an initial guess for the desired parameters as follows:

  c_1 = 650kg/s, k_1 = 1000N/m, c_2 = 650kg/s, k_2 = 1000N/m;

- Subscribe these guessed values and initial position of each segment into Eq. (1);
- Translate the 2nd order differential equation to 1st order equations and use MATLAB ODE solvers to solve them;
- Compute the difference between the experimental data and their calculated values for $Z_{0,1,2}(t)$ and $Y_{0,1,2}(t)$;
- The cost function of the optimization routine is to make the difference between the experimental data of $Z_{0,1,2}(t)$; $Y_{0,1,2}(t)$ and their calculated values minimum which can be expressed:

$$\min f = \sum_{i=0}^{2} \int_{t_0}^{t_f} [|Z_i(t) - \bar{Z}_i(t)| + |Y_i(t) - \bar{Y}_i(t)|]\, dt$$

where t_0 and t_f represents the time duration between the slip onset and the peak of the load cell force. $Z_i(t)$; $Y_i(t)$ respectively represents the measured vertical and horizontal kinematic variable i at time instant t from the experiments and $\bar{Z}_i(t)$; $\bar{Y}_i(t)$ is the simulation data coresponding to $Z_i(t)$; $Y_i(t)$ respectively.

- Four parameters (p_0, p_1, $q_0$and q_1) were used to distribute the load cell force between head-upper trunk and between upper trunk – lower trunk segments of the model in the vertical and horizontal direction such that $\sqrt{p_0^2 + p_1^2 + q_0^2 + q_1^2} = 1$. p_0, p_1, q_0, q_1 respectively represent $F_{0(z)}$, $F_{1(z)}$, $F_{0(y)}$, $F_{1(y)}$.
- When the cost function is minimum possible, the simulated annealing algorithm terminates with the optimized value for spring-damper coefficients and load cell force distribution parameters.

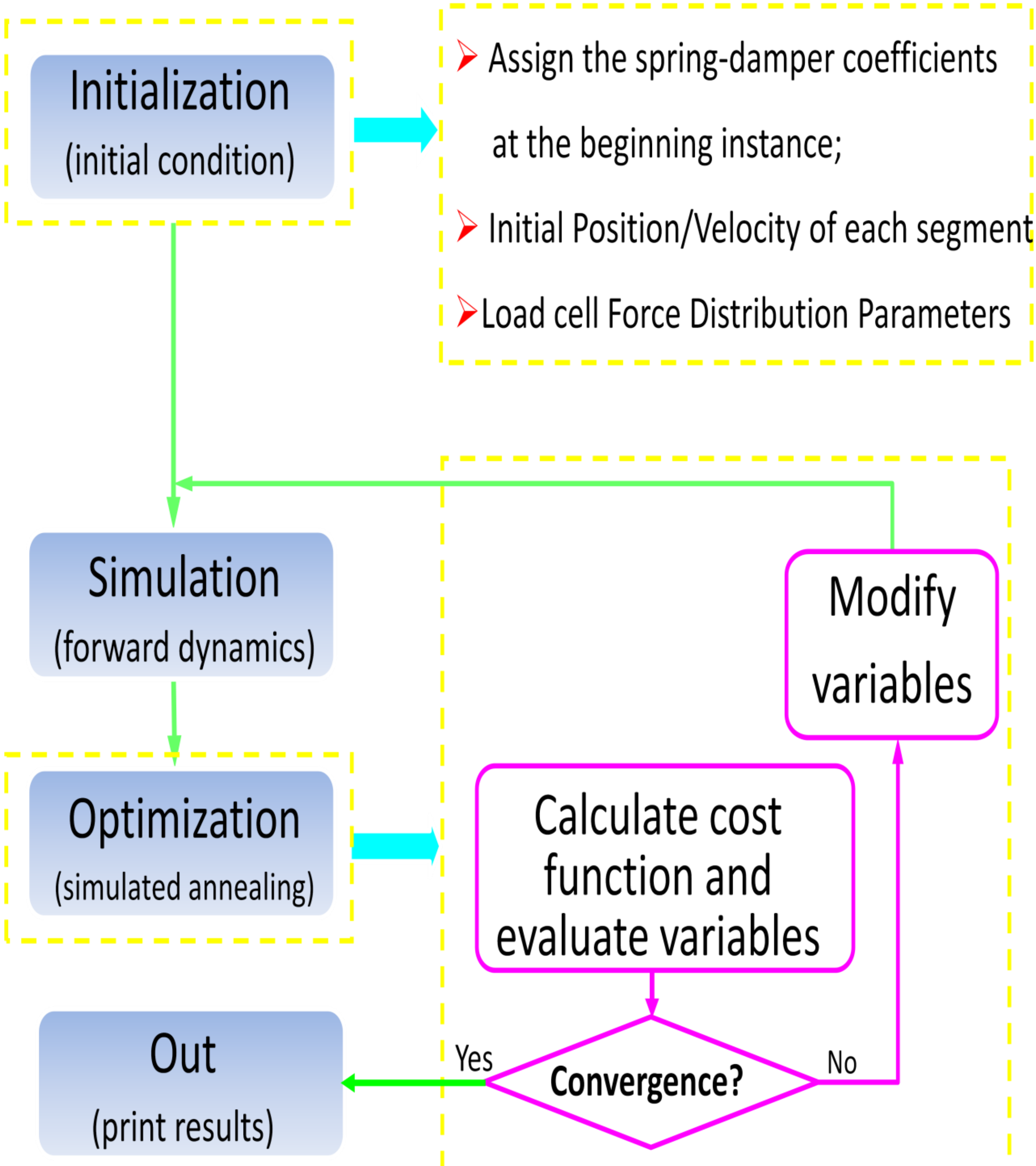


Figure 5: Schematic of the procedure to derive segmental movements which can reproduce the experimentally measured kinematics during a slip in gait. The procedure included two main parts: forward-dynamics simulation and optimization. A simulated annealing (SA) algorithm was used to perform the optimization routine.

2.5 Solution

The inter-segmental impact forces during slip related fall experiments can be computed using the results of forward dynamics based computer simulation. The detailed procedure is as follows:

- The masses (m_0, m_1 and m_2) and ($h_{1(z)}$; $h_{1(y)}$; $h_{2(z)}$ and $h_{2(y)}$) can be calculated from the collected anthropometric data.
- From experimental data, the time histories of the load cell force $F_{\text{Load-cell}}$, contact forces acting on the lower trunk ($F_{2(z)}$; $F_{2(y)}$), and movement data of all segments ($Z_{0,1,2}$(t) and $Y_{0,1,2}$(t)) can be known (section 2.2.3).
- Substitute all these information into equations of motion (section 2.2).
- The optimal values for the spring/damper coefficients can be derived by forward simulation assisted with optimization routine. The objective (or cost) function of the optimization routine is to make the difference between the experimental data of ($Z_{0,1,2}$(t) and $Y_{0,1,2}$(t)) and their calculated values minimum.
- Based on the calculated spring/damper coefficients and the movement trajectories, the impact force between every two adjacent segments can be calculated as:

$$\left.\begin{aligned}
F_{R_01(z)} &= k_{1(z)}(Z_0 - Z_1 - h_{1(z)}) + c_{1(z)}(\dot{Z}_0 - \dot{Z}_1) \\
F_{R_12(z)} &= k_{2(z)}(Z_1' - Z_2 - h_{2(z)}) + c_{2(z)}(\dot{Z}_1 - \dot{Z}_2) \\
F_{R_01(y)} &= k_{1(y)}(Y_0 - Y_1 - h_{1(y)}) + c_{1(y)}(\dot{Y}_0 - \dot{Y}_1) \\
F_{R_12(y)} &= k_{2(y)}(Y_1' - Y_2 - h_{2(y)}) + c_{2(y)}(\dot{Y}_1 - \dot{Y}_2)
\end{aligned}\right\} \quad (3)$$

- Compare the inter-segment forces with their corresponding injury threshold values

3. RESULTS

3.1 External forces acting on the Model:

Except the gravity, forces acting on the model are the load-cell force and the ground reaction force. Load-cell force acts across the shoulder and waist due to the harness system. Ground reaction force acts on the lower back from thigh.

3.1.1 Load-cell Force:

The load-cell force is the force acting on the body due to deployment of the harness during the fall. This force is distributed across the shoulder (F_0) and waist (F_1). Four parameters (p_0, p_1, $q_0$and q_1) were used to distribute the load cell force between each segment of the model in the vertical and horizontal direction such that $\sqrt{p_0^2 + p_1^2 + q_0^2 + q_1^2} = 1$. Load-cell Force along with its four components is shown in Figure 6 and 7 for slip induced fall trial and normal walking trial of a subject respectively.

3.1.2 Ground Reaction Force:

In this study, an inverse-dynamics approach was applied to derive the force exerted on lower back from both thighs. Method for computing the four forces using inverse dynamic technique is shown under section 2.2.3 (d). These forces were computed at each instance of time for all the subjects. For one of the subject, time histories of the forces acting on the lower back due to ground reaction force are shown in Figure 8. These forces are the reaction forces supplied by the ground in response to the force exerted by the body on the ground. The vertical forces combine the effect of the gravity and the effects of the body's movements and acceleration in the vertical direction while the horizontal forces represent only the effects of the body's movements and acceleration in the horizontal direction [15]. At the instant of heel contact, there is zero force between the foot and the floor due to the fact that the heel strike is defined the moment when contact is generated but there is no force production at that time instant. For a very small increment of time after a heel strike, the force will start to increase rapidly as the body begins to be supported by the foot. Soon after, full body weight will quickly be generated between the foot and the floor resulting in maximum forces. Finally the weight drops to zero as the opposite limb takes up the body weight. The anterior – posterior reaction force represents the horizontal force exerted by the force plate on the foot. This force acts in a direction of human walking and is smaller in magnitude than vertical ground reaction force component [16].

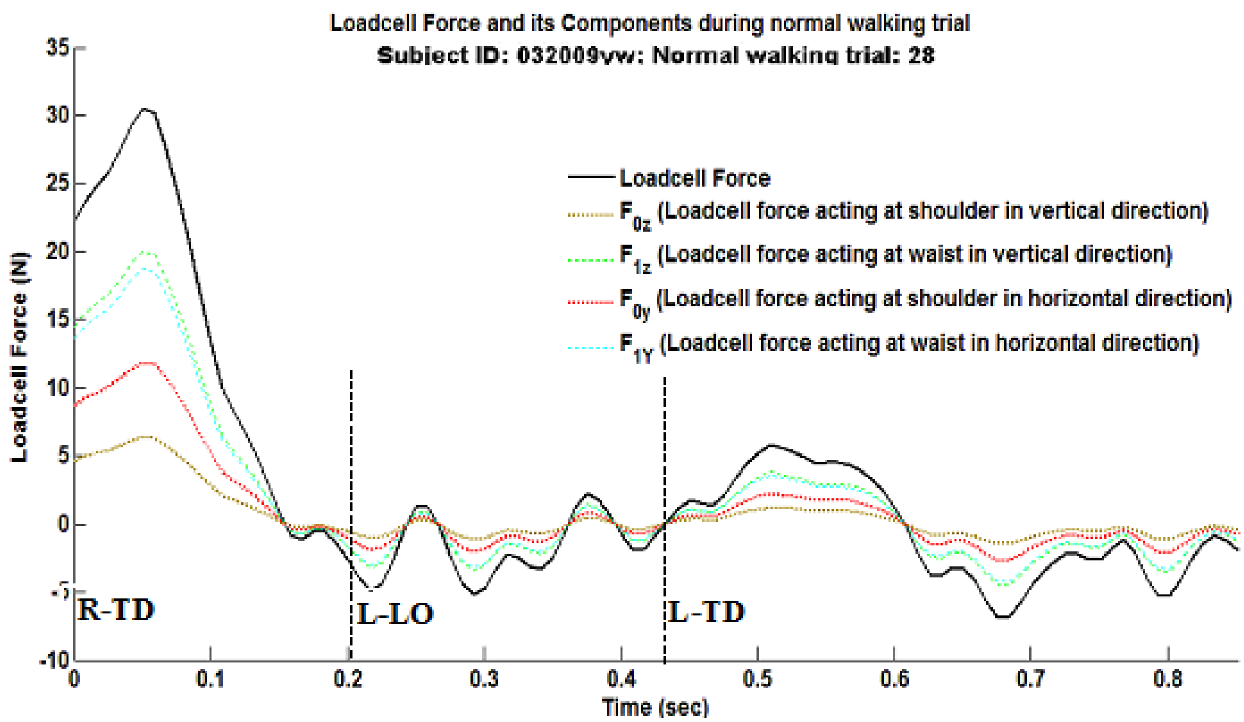


Figure 6: Loadcell force and its components plotted for a subject during normal walking trial

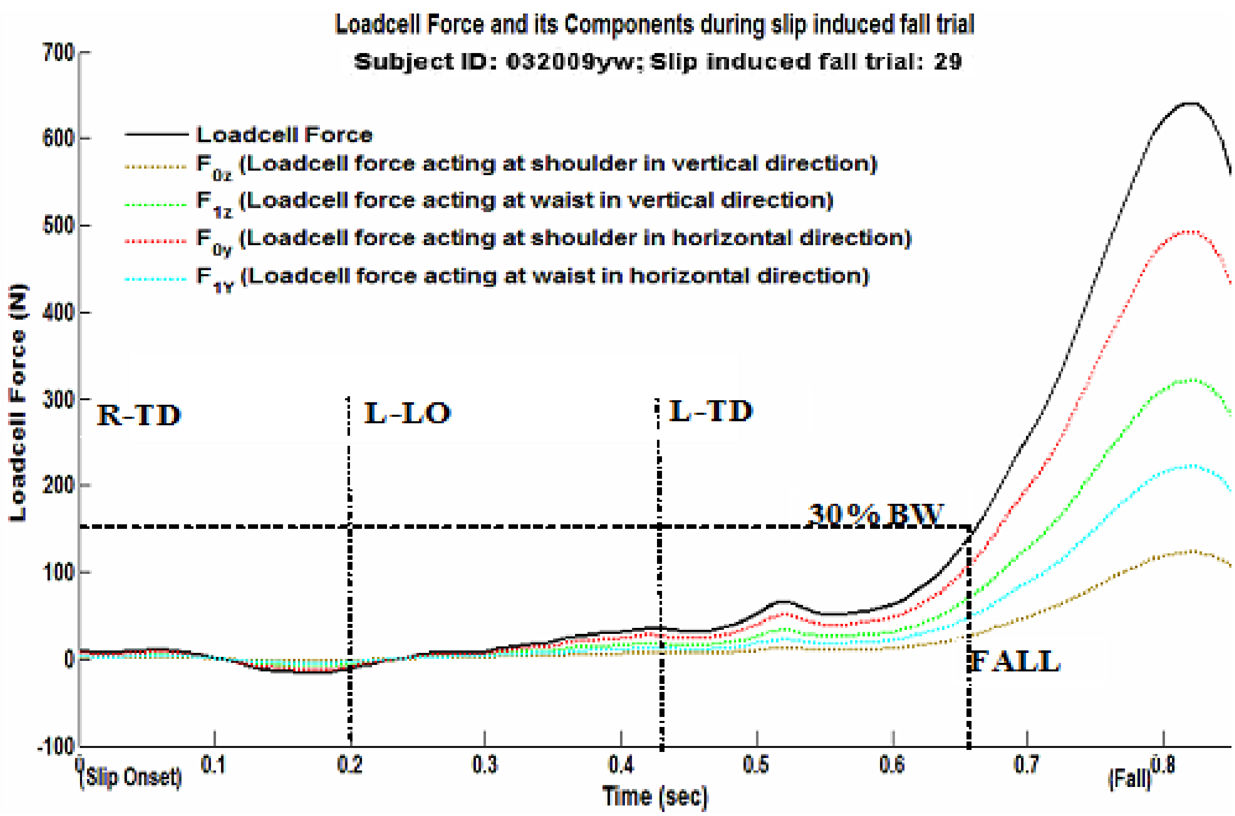


Figure 7: Loadcell force and its components plotted for a subject during slip induced fall trial

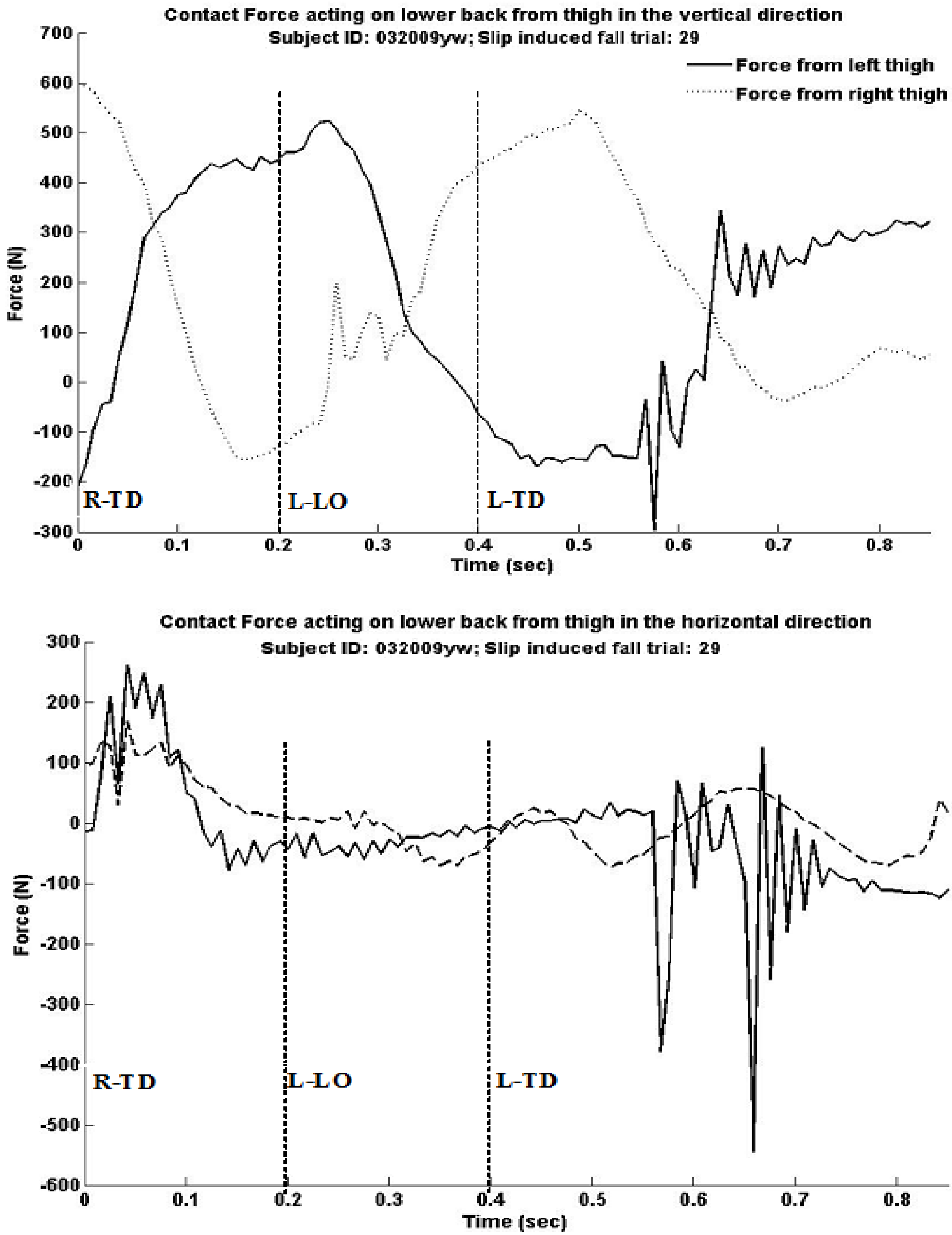
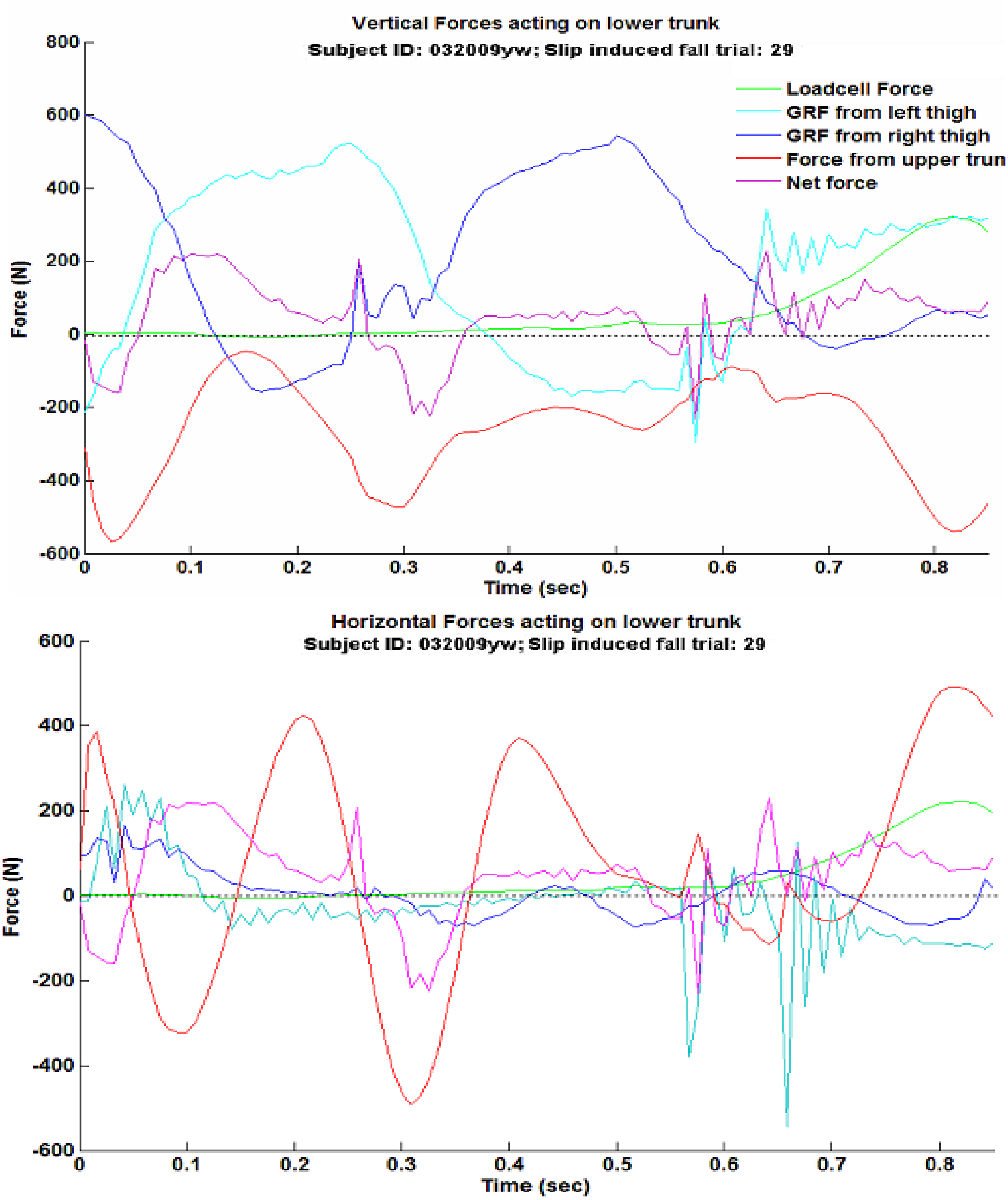


Figure 8: Ground Reaction forces acting on the lower trunk from hip for a subject during slip induced fall



<u>Figure 9</u>: For one of the subject, Forces acting on lower trunk are shown at each instance of time during slip induced fall trial

3.2 Simulation and Optimization Results

Results indicated that our methods, i.e. forward simulation and dynamic optimization could precisely reproduce the experimental kinematics. Based on each subjects' individual biomechanical model, computer simulations and optimization were performed. The experimental error was successfully reduced by optimization routine in each individualized forward-dynamics model, such that the simulated movements of body segments could closely reproduce the originally recorded joint motion [6]. A representative simulation sample of slip-initiated fall is shown in Figure 10. This figure shows that the optimization-derived result closely tracks the experimental kinematics, by the developed optimization/forward-dynamics simulation procedure. The closeness of the fit of the optimal simulated kinematics and experimental data was estimated by their correlation coefficient (σ) [6]. Coefficient of correlation is a measure of how well the simulated values from a model "fit" with the experimental kinematics. Coefficient of correlation was computed for each subject and is shown in Table I. As the values fall in the range of +0.7 to +1.0, it shows a strong positive association. The root mean square (RMS) of the residual error between the simulated results and experimental data was adopted to indicate the magnitude of the error [6]. RMS between the experimental data and their calculated values were computed for each subject and is shown in Table II.

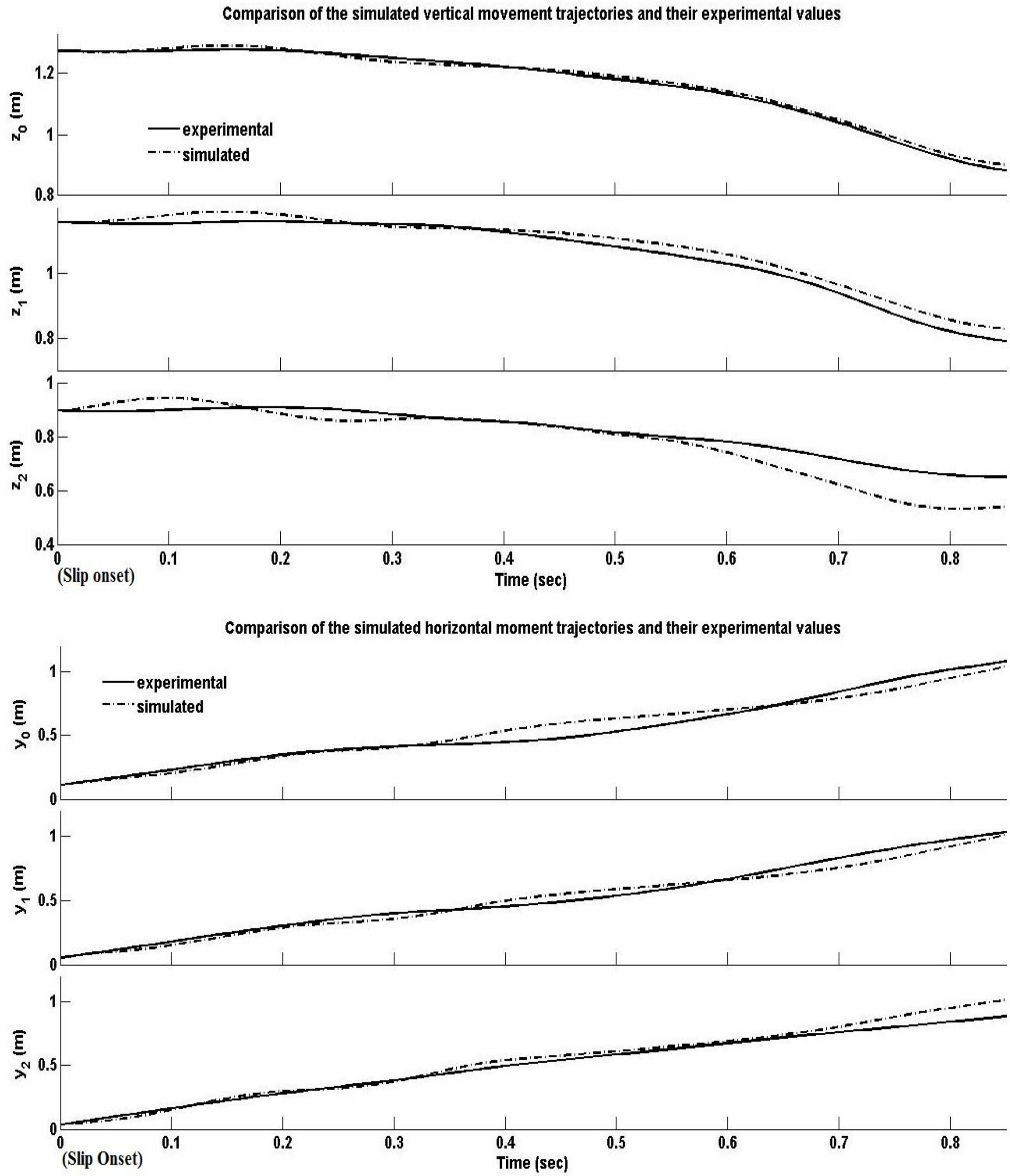


Figure 10: Comparison of the simulated moment trajectories and their experimental values for one subject. Time t=0sec indicates the instant of slip onset. Fall ended at 0.851 sec for this subject. The top figure shows the vertical displacement whereas the bottom figure shows horizontal displacement in the head, upper trunk and lower trunk.

Table I

CORRELATION COEFFICIENT BETWEEN SIMULATED AND EXPERIMENTAL KINEMATICS

Subjects	Correlation coefficient (σ) for					
	Head segment (z-axis)	Upper trunk segment (z-axis)	Lower trunk segment (z-axis)	Head segment (y-axis)	Upper trunk segment (y-axis)	Lower trunk segment (y-axis)
1	0.998	0.995	0.984	0.977	0.989	0.996
2	0.989	0.965	0.982	0.982	0.986	0.928
3	0.987	0.951	0.963	0.912	0.926	0.897
4	0.978	0.986	0.982	0.996	0.996	0.994
5	0.951	0.932	0.977	0.961	0.973	0.991
6	0.938	0.926	0.868	0.951	0.977	0.967
7	0.917	0.902	0.912	0.921	0.922	0.852
8	0.984	0.990	0.948	0.966	0.965	0.916
9	0.986	0.993	0.968	0.997	0.996	0.977
10	0.884	0.852	0.872	0.918	0.911	0.859
11	0.845	0.810	0.906	0.935	0.942	0.960
12	0.788	0.732	0.872	0.801	0.792	0.860
13	0.932	0.867	0.825	0.874	0.928	0.969
14	0.966	0.950	0.911	0.984	0.964	0.865
15	0.797	0.751	0.882	0.988	0.993	0.994
Mean	0.857	0.836	0.840	0.944	0.951	0.935
Std. Deviation	0.322	0.331	0.339	0.054	0.053	0.056

Table II

ROOT MEAN SQUARE OF THE RESIDUAL ERROR BETWEEN OPTIMAL SIMULATED AND EXPERIMENTAL KINEMATICS

Subjects	Root Mean Square (RMS) of the residual error for (mm)					
	Head segment (z-axis)	Upper trunk segment (z-axis)	Lower trunk segment (z-axis)	Head segment (y-axis)	Upper trunk segment (y-axis)	Lower trunk segment (y-axis)
1	0.9	2.3	5.8	5.6	4.2	4.6
2	2.5	2.0	4.1	5.1	4.2	8.2
3	4.1	5.5	9.6	8.7	10.3	8.5
4	6.3	5.8	11.4	3.6	6.5	6.6
5	9.8	9.3	8.7	8.3	6.9	7.8
6	7.2	4.1	4.9	4.7	3.6	5.6
7	10.6	8.1	9.6	9.8	9.8	19.2
8	4.9	2.8	7.6	13.1	14.6	30.9
9	2.9	1.6	6.5	2.0	2.6	3.3
10	16.6	4.7	16.5	20.4	21.4	28.1
11	9.9	5.5	5.5	10.6	9.7	8.2
12	16.6	15.8	23.3	26.3	27.3	22.9
13	3.8	4.8	5.7	5.9	4.8	6.2
14	4.9	4.2	4.8	6.8	4.5	6.9
15	6.1	5.6	3.9	4.3	3.1	5.5
Mean	7.2	6.2	8.5	9.0	8.9	11.5
Std. Deviation	4.76	4.3	5.3	6.5	7.2	9.0

3.3 Intersegment Reaction Forces

In this study, intersegment forces refer to the forces acting between the head and the upper-trunk and between the upper-trunk and the lower-trunk. Forces typically acting on the human body during slip-initiated fall are compression, tension and shear force. Axial loading that produces a squeezing or crushing effect is called compressive force and axial loading in the direction opposite that of compression is called tensile force. Tension is the pulling force that stretches the object to which it is applied. Compressive and tensile forces are directed toward and away from an object. A third category of force termed as Shear Force, acts parallel or tangent to a plane passing through the object. Shear force tends to cause one part of the object to slide against, displace, or shear with another part of the object [17].

For each subject, intersegment forces were computed and mean +/- 1 standard deviation are shown in figure 11 and 12 for slip induced fall trials and for normal walking trials respectively. Also maximum reaction forces between each segment were computed for slip induced fall trials and normal walking trial and are shown in table III and IV. During slip induced fall trial, the maximum compressive force recorded for head - upper trunk segment was 226 N and for upper trunk - lower trunk was 1272 N. The maximum tensile force recorded for head – upper trunk was 329 N and upper trunk – lower trunk was 2115 N. The maximum Shear force computed anterior-posterior for head – upper trunk was 455 N and upper trunk – lower trunk was 1940 N. Comparison of maximum impact forces during slip induced fall and during normal walking was done using student t-test and is shown in Figure 13. There were significant difference between normal and slipping trials.

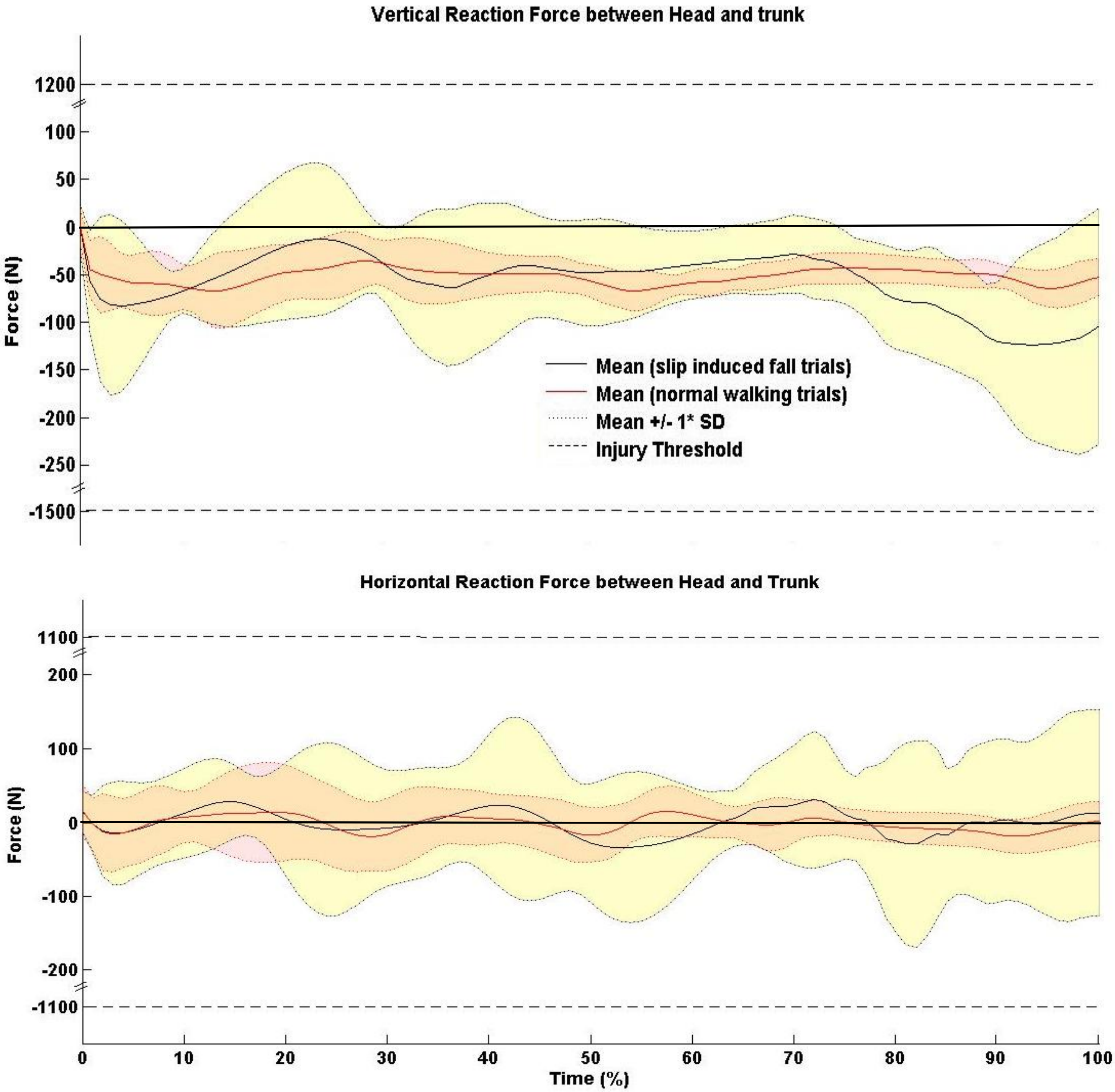
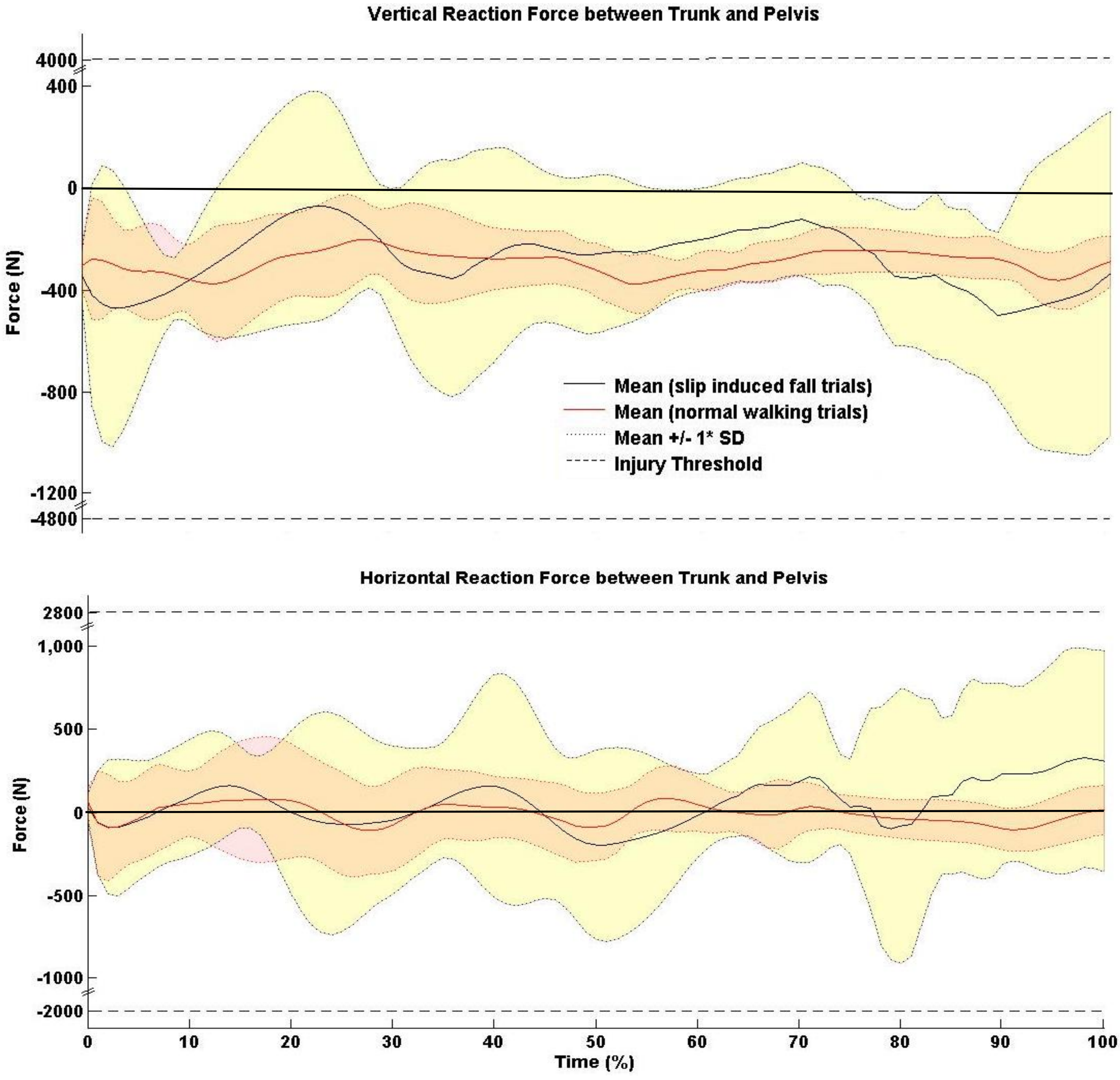


Figure 11: Inter-segmental Vertical and Horizontal forces between head – upper trunk for 15 subjects. Mean along with ± 1 standard deviation for 15 subjects were calculated for slip induced fall trials and normal walking trials. For the vertical forces, the positive value indicates the tensile force acting in that segment pair whereas negative values indicate compressive forces. The horizontal forces represent the shear force acting in anterior-posterior direction.



<u>Figure 12</u>: Inter-segmental Vertical and Horizontal forces between upper trunk – lower trunk for 15 subjects. Mean along with ±1 standard deviation for 15 subjects were calculated for slip induced fall trials and normal walking trials. For the vertical forces, the positive value indicates the tensile force acting in that segment pair whereas negative values indicate compressive forces. The horizontal forces represent the shear force acting in anterior-posterior direction.

Table III

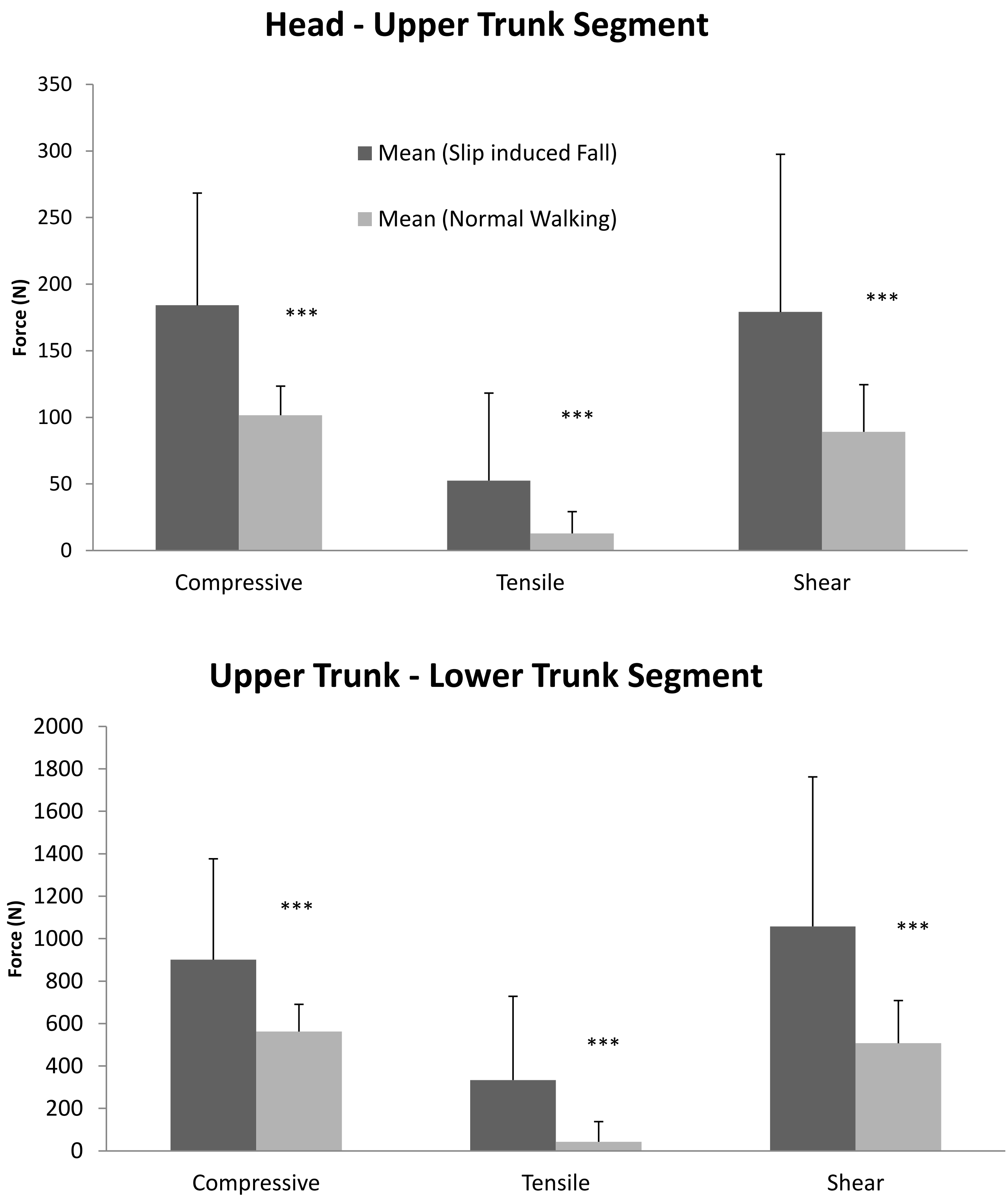
MAXIMUM REACTION FORCE BETWEEN EACH SEGMENT PAIR OF THE MODEL DURING NORMAL WALKING (NO SLIP INDUCED)

Subjects	Max. Vertical Force (N)				Max. Horizontal Force (N)	
	Compressive		Tensile		Shear	
	Head- Upper trunk	Upper-Lower trunk	Head- Upper trunk	Upper-Lower trunk	Head- Upper trunk	Upper-Lower trunk
1	103.55	584.56	14.62	11.69	192.05	1079.58
2	76.40	435.67	0	0	68.24	383.76
3	102.16	574.34	4.39	27.13	56.22	317.08
4	79.50	454.29	5.37	0	75.55	466.93
5	108.17	610.32	19.17	23.71	94.83	526.94
6	84.03	477.03	0	0	45.76	197.04
7	96.85	546.86	20.11	0	113.31	644.48
8	162.36	916.41	43.91	242.98	60.80	353.23
9	105.01	599.15	0	0	75.31	436.98
10	122.45	693.65	53.88	305.59	64.79	373.49
11	73.06	404.41	14.09	0	110.76	623.86
12	93.16	509.15	0	0	105.71	591.50
13	109.59	594.36	12.23	12.67	94.29	538.97
14	111.66	605.68	0	0	104.63	569.11
15	98.66	428.39	5.89	16.87	75.48	501.36
Mean	101.65	562.28	12.91	42.71	89.18	506.95
Std. Deviation	21.87	128.38	16.36	95.22	35.43	201.45

Table IV

MAXIMUM REACTION FORCE BETWEEN EACH SEGMENT PAIR OF THE MODEL DURING SLIP-RELATED FALLS IN GAIT

Subjects	Max. Vertical Force (N)				Max. Horizontal Force (N)	
	Compressive		Tensile		Shear	
	Head- Upper trunk	Upper-Lower trunk	Head- Upper trunk	Upper-Lower trunk	Head- Upper trunk	Upper-Lower trunk
1	120.26	568.81	13.80	0	87.78	491.28
2	79.04	411.15	0	310.05	105.44	810.22
3	175.76	715.64	19.71	120.44	130.45	626.53
4	130.27	672.14	24.27	135.01	118.15	782.18
5	165.89	648.07	5.16	8.79	88.49	329.21
6	170.94	962.31	75.16	422.56	97.15	892.15
7	162.05	932.72	162.61	1116.62	234.87	1333.89
8	115.02	560.35	28.67	0	455.44	2743.43
9	226.33	786.43	13.31	343.50	157.14	1023.76
10	374.08	2115.48	226.37	1272.83	341.57	1941.89
11	124.08	701.58	82.10	460.13	326.17	1836.10
12	283.31	1602.11	85.91	508.10	265.98	1646.54
13	178.98	631.49	3.81	0	72.76	275.53
14	329.42	1534.13	47.33	298.63	142.99	526.23
15	126.35	666.99	0	8.77	62.74	607.08
Mean	184.12	900.63	52.55	333.69	179.14	1057.73
Std. Deviation	84.36	475.81	65.73	394.72	118.46	704.72



<u>Figure 13</u>: Comparison of maximum inter-segmental forces for 15 subjects during slip induced fall trials and normal walking trial. ***P < 0.001 indicates significant change.

3.4 Injury Criteria

Injuries due to slip-related falls in gait can be produced by two main mechanisms. Firstly by direct impact, where the rapid acceleration of that particular body part's mass produces the inertia loads resulting in a direct injury. Secondly, by indirect loading, where the loads are generated not by the inertia of that body part, but, by the acceleration forces of the whole body transferring loads through that body part. In order to evaluate the severity of a particular type of injury an injury criterion is used with tolerance levels to determine the actual severity level. Definitions of both injury criterion and tolerance levels are given overleaf [18].

Table IV

HUMAN BODY INJURY CRITERIA

Neck/Cervical Spine Injury Criteria	Compressive Force (N)	Tensile Force (N)	Shear Force (N)
Soft Tissue Injury	1500 – 4000	1200 – 3300	1100 – 3300
Severe Injury	> 4000	> 3300	> 3300

Thoracic-Lumbar Injury Criteria	Compressive Force (N)	Tensile Force (N)	Shear Force (N)
Soft Tissue Injury	No data	No data	1290 – 2000
Severe Injury	> 4800	> 4000	2000 – 2800
IV Disc	12000	2800	No data

4. DISCUSSION

The purpose of this study was to examine the impact force between body segments resulted from safety harness during falling. Modeling approach was used to calculate the contact force between each pair of body segments due to the difficulty to directly measure them. In this study, human upper body was modeled as a mechanical system with masses and connecting springs and dampers. The inter-segmental forces were then obtained by driving the model based on forward dynamic computer simulation. The experimental error was successfully reduced by optimization routine in each individualized forward-dynamics model, such that the simulated movements of body segments could closely reproduce the originally recorded motion. Results indicated that our methodology could successfully and precisely reproduce the experimental kinematics. The mean RMS and the mean Correlation coefficient (σ) between optimal simulated kinematics and experimental data are 0.0855 and 0.894 respectively.

Forces acting between the head and the upper-trunk and between the upper-trunk and the lower-trunk were computed for slip-induced fall trials and normal walking trials. Comparison was drawn using student t-test and results clearly indicated that there was significant difference between the inter-segmental forces during slip-induced fall trials and during normal walking. Inter-segmental forces during slip-induced fall trials were then compared to human injury threshold. Results showed that the impact compressive or tensile force would not cause any injury to either bone or soft tissue. However, for two subjects relatively higher forces were noted in the lower back or thoracic-lumbar region. The compressive forces were around 1200N for both the subjects and tensile forces were 1534N and 2110N. Forces of this magnitude would not cause any severe injury. But it is not sure whether such forces can cause any soft tissue injury in the thoracic-lumbar region because of the lack of injury threshold data. For the shear forces in the neck or the cervical spine region, the magnitude of forces were way below the threshold thereby ruling out the chances of injury in this region. However the shear forces recorded for the thoracic-lumbar region for 5 subjects indicate soft tissue injury. Analyzing the results, a general trend was observed which indicated relatively higher reaction forces in the thoracic-lumbar region as compared to neck or cervical spine region. This could be explained by three factors: 1) when a person slips and is supported by the harness entire body mass is acting on lower back segment, 2) gravitation force acting on lower back segment is greater than that acting on the neck or cervical region and 3) the energy is absorbed by tissues as it travels from the lower back region to the head region.

Summarizing the results, model predicted that no severe injury would have occurred. This was supported by the participant's follow-up in which none of the subjects reported issues concerning injury. Our study is the first one to quantitatively examine the impact force between body segments during slip-induced fall. This study could advance our understanding in human injury during slip-related falls in gait. Also importantly this study could provide theoretical guidance in designing safety harness.

REFERENCES

[1] Bhatt T, Wening JD, Pai YC. Influence of gait speed on stability: recovery from anterior slips and compensatory stepping. Gait Posture. 2005;21:146-56.

[2] Pai YC, Bhatt T, Wang E, Espy D, Pavol M. Inoculation against falls: Rapid adaptation by young and older adults to repeated slips during daily activities. Arch Phys Med Rehabil. 2010;91:452-9.

[3] Pai YC, Bhatt T. Repeated slip training: An emerging paradigm for prevention of slip-related falls in older adults. Physical Therapy 2007;87:1478-91.

[4] Yang F, Pai YC. Automatic Recognition of falls in gait-slip training: Harness Load cell based criteria. J Biomech. 2011;44:2243-9.

[5] Mosley R, Quick N, Seidlitz P. (2006). Slips, Trips and Falls: Resident Care Facilities. Available:http://www.illinoisosha.com/PDF/Poster_SlipsTripsFalls.pdf. Last accessed 29 March 2012.

[6] Yang F, Pai YC. Reactive control and its operation limits in responding to a novel slip in gait. Ann Biomed Eng. 2010;38:3246-56.

[7] Bhatt T, Espy D, Yang F, Pai YC. Dynamic gait stability, clinical correlates, and prognosis of falls among community-dwelling older adults. Arch Phys Med Rehabil. 2011;92:799-805.

[8] Yang F, Bhatt T, Pai YC. Limits of recovery against slip-induced falls while walking. J Biomech. 2011;44:2607-13.

[9] Piazza SJ. Muscle-driven forward dynamic simulations for the study of normal and pathological gait. J Neuroeng Rehabil. 2006;3:5.

[10] Liu W, Nigg BM**.** A mechanical model to determine the influence of masses and mass distribution on the impact force during running. J Biomech. 2000;33:219-24.

[11] Nikooyan AA, Zadpoor AA. Mass-spring-damper modelling of the human body to study running and hopping: an overview. Proc Inst Mech Eng H. 2011;225:1121-35.

[12] Thelen DG, Anderson FC, Delp SL. Generating dynamic simulations of movement using computed muscle control. J Biomech. 2003;36:321-28

[13] Clark F. (1998). The simulated annealing algorithm - a brief overview. Available: http://www.clarkfrancis.com/docs/sim_ann.html. Last accessed 29 March 2012.

[14] Simulated Annealing. Available: http://www.mathworks.com/discovery/simulated-annealing.html. Last accessed 29 March 2012.

[15] Thompson David M. (2002). Ground Reaction Force. Available: http://moon.ouhsc.edu/dthompso/gait/kinetics/GRFBKGND.HTM. Last accessed 29 March 2012.

[16] Tea Marasovic, Mojmil Cecic, Vlasta Zanchi. Analysis and Interpretation of Ground Reaction Forces in Normal Gait. WSEAS Transaction on Systems, No. 8, 2008; 1105-1114.

[17] Anderson Marcia K.: Fundamentals of Sports Injury Management. Lippincott Williams & Wilkins; 2Rev Ed edition (2002).

[18] Injury Mechanism and Injury Criteria. Available: http://www.eurailsafe.net/subsites/operas/HTML/Section3Menu.htm. Last accessed 1 May 2012.

[19] Jason Brick: Safety Harness Injuries. Available: http://www.livestrong.com/article/152090-safety-harness-injuries/. Last accessed 1May 2012.

[20] Guide Fisher: Safety Harness Injuries. Available: http://www.ehow.com/about_6605769_safety-harness-injuries.html. Last accessed 1May 2012.

APPENDIX

PERMISSION LETTER TO REPRINT COPYRIGHT MATERIAL

08/01/2012

Dr. Fang Yang

I am writing to request permission to use the “Schematic illustration of the experimental setup for inducing unannounced slips in gait” from your publication (‘Limits of recovery against slip-induced falls while walking’ J Biomech. 2011; 44:2607-13) in my thesis. This material will appear as originally published. Unless you request otherwise, I will use the conventional style of the Graduate College of the University of Illinois at Chicago as acknowledgment.

A copy of this letter is included for your records. Thank you for your kind consideration of this request.

Sincerely,
Kunal Jariwala

The above request is approved.
Approved by:____ Date:________________

Additional remarks may be included here for clarity

KUNAL JARIWALA

EDUCATION:

- Masters of Science in Bioengineering — GPA: 3.25/4.0
  University of Illinois-Chicago (UIC); Chicago, IL — Aug 2010 – May 2012

  Thesis: 'Impact Force on Human Body during Slip-related Falls in Gait'
  Advisor: Dr. Clive Pai

- Bachelors of Engineering in Biomedical Engineering — (First Class with Distinction)
  University of Mumbai; Mumbai, India — Aug 2006 – May 2010

  Final Year Project: 'Low Cost Foot Pressure Sensing System'
  Advisor: Dr. Manali J. Godse

RESEARCH EXPERIENCE:

- Thesis, University of Illinois-Chicago, IL — Sept 2010 – May 2012
  - Modeled human body as 3-segment spring-damper-mass model
  - Performed computer simulation and dynamic optimization to calculate reaction forces between each body segment during fall
- Research Intern, Stony Brook University, NY — May 2011 – August 2011
  - Conducted Clinical trials for Motion Analysis of Football Player during offensive linemen's tackle
  - Performed data reduction and statistical analysis to generate summaries and reports
  - Publications: Abstract for poster presentation to 'American Physical Therapy Association'

PROFESSIONAL EXPERIENCE:

- Coop, Biomedicon Systems (I) Pvt. Ltd., India — June 2009 – January 2010
  - Executed instrument field service functions & regulated instrumentation repairs
  - Managed on-site installation at domestic sites as well as extended technical phone support and coordinated with client to resolve problems

CERTIFICATION:

- CITI Certified Researcher trained in Responsible Conduct of Research (RCR), Human Subjects Protection and HIPAA

AWARDS:

- Recipient of prestigious JRD Tata scholarship Aug 2009
- 2004 Bronze medal & Certificate of Merit, State Board S.S.C. Examination
- State Talent Search Candidate (Maharashtra, India) in 2002