A COMPUTATIONAL FRAMEWORK FOR ANALYSIS AND SYNTHESIS OF CONTROL STRATEGIES IN HUMAN AUGMENTATION DEVICES

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1. ABSTRACT

Development of human augmentation devices for restoration or enhancement of motor function is an important area of research in helping humans with disabilities or compromised neuromuscular function. Although realization of human augmentation systems for day-to-day activities is still in its infancy, significant progress is under way toward meeting this objective. At Honda's Wako Research Center, a mechanically powered walking assist prototype system was recently unveiled. Two important and challenging question to consider in the implementation of the Honda prototype and similar human augmentation systems include: 1) analysis and monitoring of biomechanical as well as physiological quantities which cannot be readily measured and 2) the synthesis of an active control which can safely and effectively augment voluntary and involuntary control. In this paper, we present a computational framework whereby the analysis and synthesis problems are considered in a unified control theoretic formulation. By developing computational methods to study these issues, future performance and designs of human augmentation devices can be studied through simulation, without the risk and constraints imposed by hardware implementations of current technology.

2. INTRODUCTION

Effective usage of human assistive systems or augmentation devices for restoration or enhancement of motor function is an important area of research in rehabilitation and performance enhancement. A partial list of important and desired features of an effective assistive system include: (1) a decrease in energy rate and cost with respect to able-bodied subjects performing the same task. (2) minimum disruption and maximum comfort of normal activities when employing the assistive system and (3) practicality. The third requirement considers the ease of wearing such a device and its power consumption needs. These requirements and available technology have led to the development of externally powered orthoses and prostheses that interface directly or indirectly with the human neuromuscular system. Although significant progress has been made in meeting many of the requirements needed for development of practical human assist devices (Popovic 2000), realization of such systems for daily applications is still in its infancy. The complexity of the central nervous system (CNS) control and the interface between voluntary control and external artificial control are still challenging, unanswered questions.

At Honda's Wako Research Center, a mechanically powered walking assist prototype

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system was recently unveiled (Kato and Hirata, 2001}. The target application is to help the elderly and disabled people either execute daily tasks they could not previously perform, or use less physical exertion than they currently employ for these tasks. The tasks considered include walking, lifting, sitting/standing, and climbing stairs. Two important and challenging question to consider in the implementation of the Honda prototype and similar human augmentation systems include: 1) analysis and monitoring of biomechanical as well as physiological quantities which cannot be readily measured and 2) the synthesis of an active control which can safely and effectively augment voluntary control. By developing computational methods to study these issues, future performance of human augmentation devices can be studied through simulation, without the risk and constraints imposed by hardware implementations of current technology.
Simulation studies also enable us to estimate physiological quantities which cannot be easily measured, including muscle forces, joint forces, and energetics of motion. We can simulate effects of aging, predict muscular activity, estimate muscle fatigue and capacity, and detect potential dangerous physiological conditions. It should be mentioned that the exclusive use of simulation is not a substitute for eventual testing on live human subjects. However, an accurate subject-specific simulation allows control algorithms to be designed and refined for the walking assist device without injuring a human operator. This is especially relevant in our target user population because they already have existing health constraints.

In this paper, we present computational methods for analysis, and synthesis of human motion under partial assist from powered augmentation devices. The algorithms are integrated in a simulation platform to be used as a test-bed for prototyping, simulating, and verifying algorithms to control human motion under artificial control. The analysis and synthesis problems in human motion are formulated as a trajectory tracking control algorithm using inverse and forward models coupled by proportional and derivative feedback terms. A muscle force distribution and capacity module is used to monitor the computed joint torques in order to assess the physiological consequences of the artificial control, and if needed to make modifications. This framework allows us to verify robustness, stability, and performance of the controller, and to be able to quickly change parameters in a simulation environment. We can detect unsafe or harmful control strategies and study many different motions in a simulation environment.

3. THE SYSTEM MODEL

The system (or plant) refers to a dynamic model of the combined musculoskeletal and augmentation device system. The system may be designed having various degrees of complexity, depending on the requirements imposed by the study. Without loss of generality, we consider a simple planar biped system to illustrate the concepts (See Figure 1). The equations of motion are formulated in such a way to handle three phases of biped motion as shown in Figure 1b. They include single support, double support, and airborn. Let q be the coordinates corresponding to the rotational and translational degrees of freedom.

$$q = [x_3\ y_3\ \Theta_1\ \Theta_2\ \Theta_3\ \Theta_4\ \Theta_5]^T$$

Where (x3,y3) corresponds to the center of mass of the torso and the joint angles Θ are measured clockwise from the vertical.

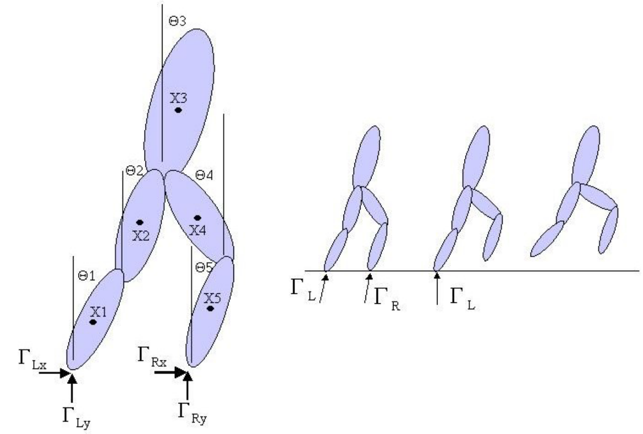


Figure 1 A biped system having five degrees of freedom in the sagittal plane with intermittent ground contact during double support, single support, and air-born phase.

The system is actuated by voluntary control from the muscles and artificial control from the augmentation device. The total torque applied at the joints are the combined torque from the muscles (τ_m) and the assist actuators (τ_a),

$$\tau = \tau_a + \tau_m$$

Let *C(q)* represent the foot-floor contact constraints and $\Gamma = [\Gamma_L \;\; \Gamma_R]^T$ be the vector corresponding to the ground reaction forces under the left and right feet. The equations of motion of the system are given by,

$$J(q)\ddot{q} + B(q,\dot{q})\dot{q} + G(q) + T_{ad} = \frac{\partial C^T}{\partial q}\Gamma' + D\tau$$

where J, B, and G correspond to the inertia, coriolis and centripetal torques, and gravitational terms, respectively. The vector T_{ad} models the augmentation device dynamics and the constant matrix D characterizes the torque coupling effects at the joints. The ground reaction forces may be expressed as a function of the state and inputs by (Hemami 1980),

$$\Gamma = \left(\frac{\partial C}{\partial q} J(q)^{-1} \frac{\partial C^T}{\partial q}\right)^{-1} \left(-\frac{\partial}{\partial q}\left(\frac{\partial C}{\partial q}\dot{q}\right)\dot{q} + \frac{\partial C}{\partial q} J(q)^{-1}\big(B(q,\dot{q})\dot{q} + G(q) + T_{ad} - D\tau\big)\right)$$

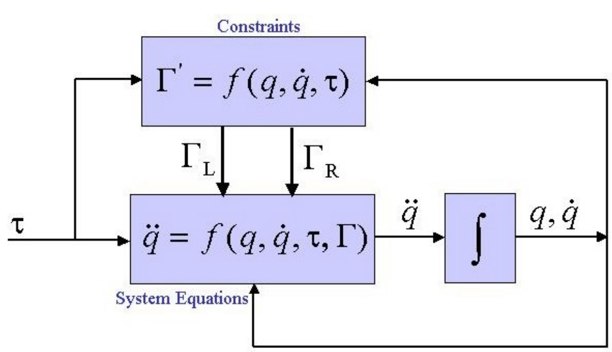


Figure 2: System Model Description with intermittent contact of left and right feet with the ground.

3. THE INTERNAL (INVERSE) MODEL

It has been demonstrated that the behavior of the human body when coupled with a novel mechanical system is very similar to the behavior that results when the controller relies on an internal model. One such internal model is thought to be a forward model, a term used to describe the computations involved in predicting sensory consequences of a motor command. There are a number of studies that have suggested that a forward model may be used by the human central nervous system (CNS) to estimate sensory consequences of motor actions (Wolpert et al., 1993: Flanagan and Wing,1997}. This theory is easily understood when considering transmission delays inherent in the sensory-motor loop Although a forward model is particularly relevant to feedback control of time delayed systems, an inverse model is sometimes considered to predict the motor commands that are appropriate for a desired behavior (Atkeson, 1989; Kawato, 1989; Shadmehr, 1990; Gomi and Kawato, 1992).

Inverse models are generally not considered for control of time delayed systems since the controller would seem to not have the ability to respond to the error and results in instability. However, it is plausible that local or intrinsic feedback mechanisms in conjunction with an inverse model can function to stabilize a system with latencies. Local feedback with stabilizing characteristics is believed to exist in humans in the form of viscoelastic properties of muscles and spinal reflex loop. The concept of an inverse model is also attractive for analysis problems of biomechanical quantities, whereby internal loads are estimated from kinesiological measurements. The approach adopted here in developing a computational model of human sensorimotor control is based on the concept of an inverse model coupled with nonlinear feedback (Figure 3). This mechanism is compelling from the standpoint of biomechanical analysis of human motion as well as the synthesis of artificial control. Let q_d represent the desired kinematics, obtained from motion capture data. The following control law, when applied to the system equations, will result in a simulated response that will track and reproduce the desired kinematic data,

$$D\,\tau_{\mathrm{m}} = J(q)\ddot{q}^* + B(q,\dot{q})\dot{q} + G(q) + T_{ad} - \frac{\partial C^T}{\partial q}\Gamma' - D\,\tau_{\mathrm{a}}$$

Where,

$$\ddot{q}^* = a\,\ddot{q}_d + K_p(q_d - q) + K_v(\dot{q}_d - \dot{q})$$

The diagonal matrices Kp and Kv represent the position and velocity feedback gains, respectively. The eigenvalues of the closed loop system are related to the feedback gains by the following,

$$K_{\mathrm{p}} = -(\lambda_1 + \lambda_2)$$
$$K_v = \lambda_1\lambda_2$$

A critically damped response to the tracking error can be achieved by specifying the eigenvalues to be equal, real, and negative. The parameter a is constant and set to 0 or 1, depending on the severity of noise in the measurements. If the desired trajectories are obtained from noisy motion capture measurements, it may be appropriate to set $a = 0$ and to specify the eigenvalues to be large and negative, This way, tracking is achieved without the need to compute unreliable accelerations from noisy kinematic data.

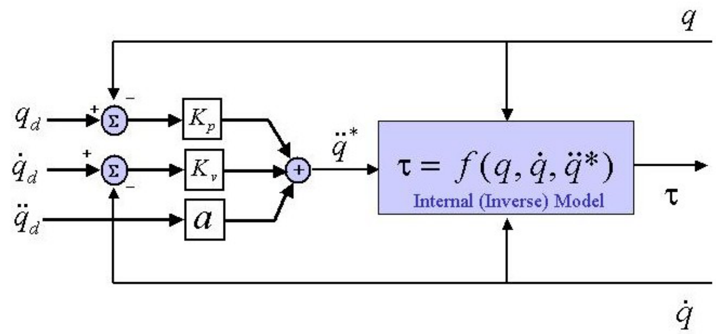


Figure 3: Inverse dynamics model with position and velocity feedback for calculation of torques that when applied to a system model, will track and reproduce the desired kinematic data.

3. MUSCLE FORCE & MUSCLE CAPACITY

The muscle force and muscle capacity module is currently implemented as a separate module and not integrated in the forward path in the closed loop system (Hungspreugs et. al 2000). The underlying concepts are presented below.

The relationship between the net muscular moment τ_m and the muscle forces F_m is given by,

$$D\tau_m = -\frac{\partial L^T}{\partial q} F_m$$

where $\partial L^T / \partial q$ is an $(n \times m)$ muscle moment arm matrix. Since the number of muscles (m) exceeds the degrees of freedom (n), the computation of the muscle actuator's excitation inputs (and the resulting forces) from an inverse dynamics computation amounts to solving a problem that is inherently ill-posed. Static, nonlinear optimization has been used extensively to predict the individual muscle forces to produce the required torque. There are several compelling reasons for using static optimization to predict the individual muscle forces: first, static, non-linear optimization techniques have well developed theoretical foundations. With the advance of commercial software for solving general, constrained, multi-variable non-linear optimization problems, it is now possible to solve sophisticated problems numerically in relatively short time. Second, the notion that muscle forces are controlled in some way to optimize physiological criteria has great intuitive appeal. It has been shown that for motions like walking, static optimization yields very similar results to dynamic optimization (Anderson and Pandy, 2001).

A muscle force and muscle capacity module takes the computed torques from the inverse model (denoted by $D\tau'$) as inputs and calculates the muscle forces based on a static optimization criterion (Figure 4). The muscle forces are compared with physiological capacity of the muscle in the muscle capacity module. The maximum force limits can be ascertained from the well-studied force-length-velocity relationship of muscle (Zajac 1989). In addition, the muscle forces with and without the assist torque are compared in order to assess whether the assist torque control has helped (improved efficiency) or hindered the motion. If the assist torque control hinders motion, the muscle forces are adjusted and feasible joint torques are computed. A poorly designed assist control would then result in $D\tau' \neq D\tau$, resulting in a response that would not track the desired response. If the assist torques are well designed, $D\tau' = D\tau$ and the resulting motion would track the desired motion.

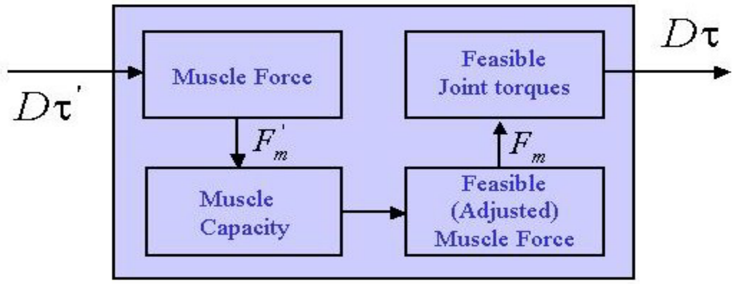
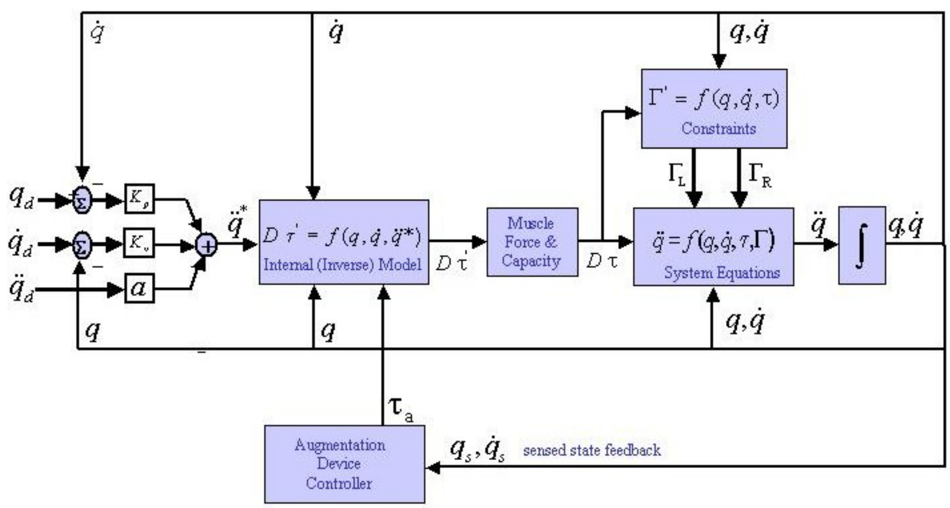


Figure 4 The Mucle Force and Muscle Capacity Module

4. INTEGRATION OF MODULES

The block-diagram of the integrated modules described in this paper system is shown in Figure 5. The Augmentation device controller is presumed to have as inputs the sensed states and output the assist torques. The overall framework is very general and enables flexible design of the augmentation device control signals. The details of such designs are beyond the scope of this paper.



Figrue 5. The block-diagram of the integrated simulation system.

5. SIMULATIONS

A very simple simulation of the tracking system is carried out to illustrate some of the concepts proposed in this paper. Due to space limitations, we present only a simulation illustrating the tracking characteristics of the proposed method without acceleration estimates of the reference trajectory. In particular, we simulated the double support phase of the biped system during a squatting maneuver. The results are illustrated in Figures 6-8. In Figure 6, the desired and simulated joint trajectories illustrate the effectiveness of the tracking procedure. These results were obtained by setting a=0, i.e. no acceleration estimates were used as inputs to the inverse model. The corresponding joint torques and ground reaction forces are depicted in Figure 7 and Figure 8, respectively.

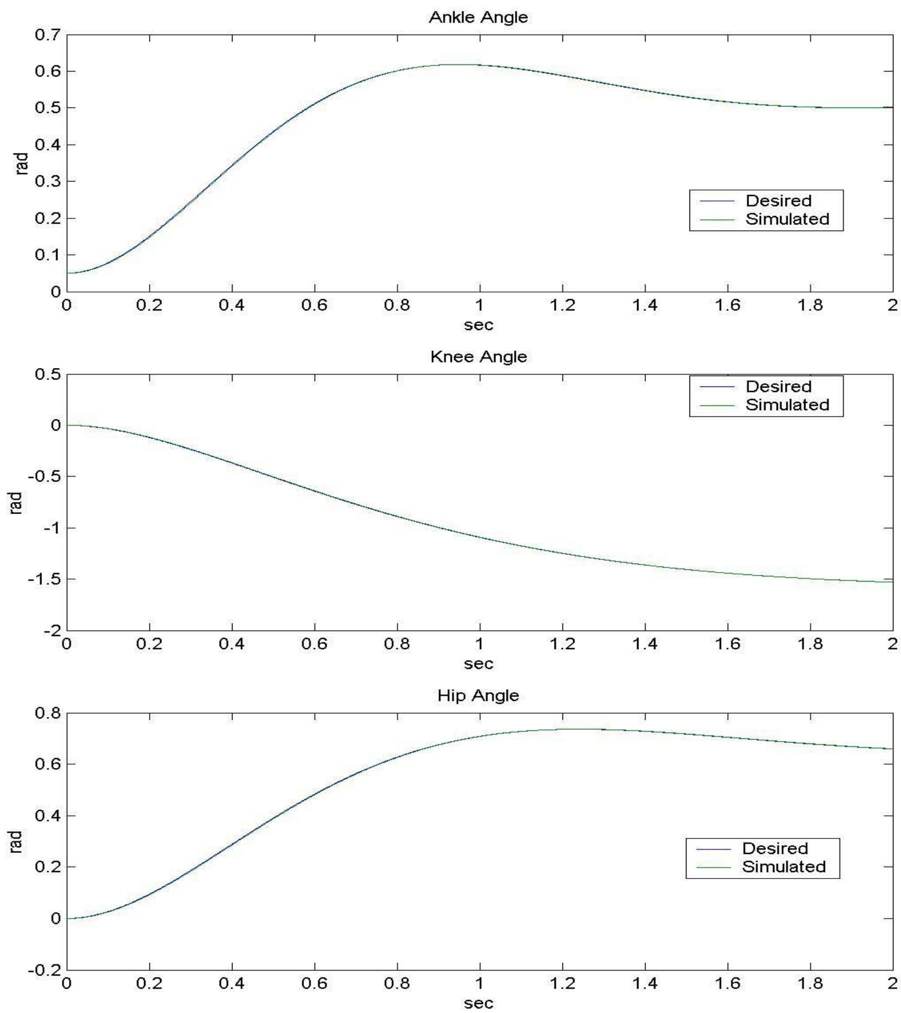


Figure 6 Simulation of the joint angles during a squatting maneuver without employing the desired accelerations (a=0). The result illustrate nearly perfect tracking of desired kinematic trajectories.

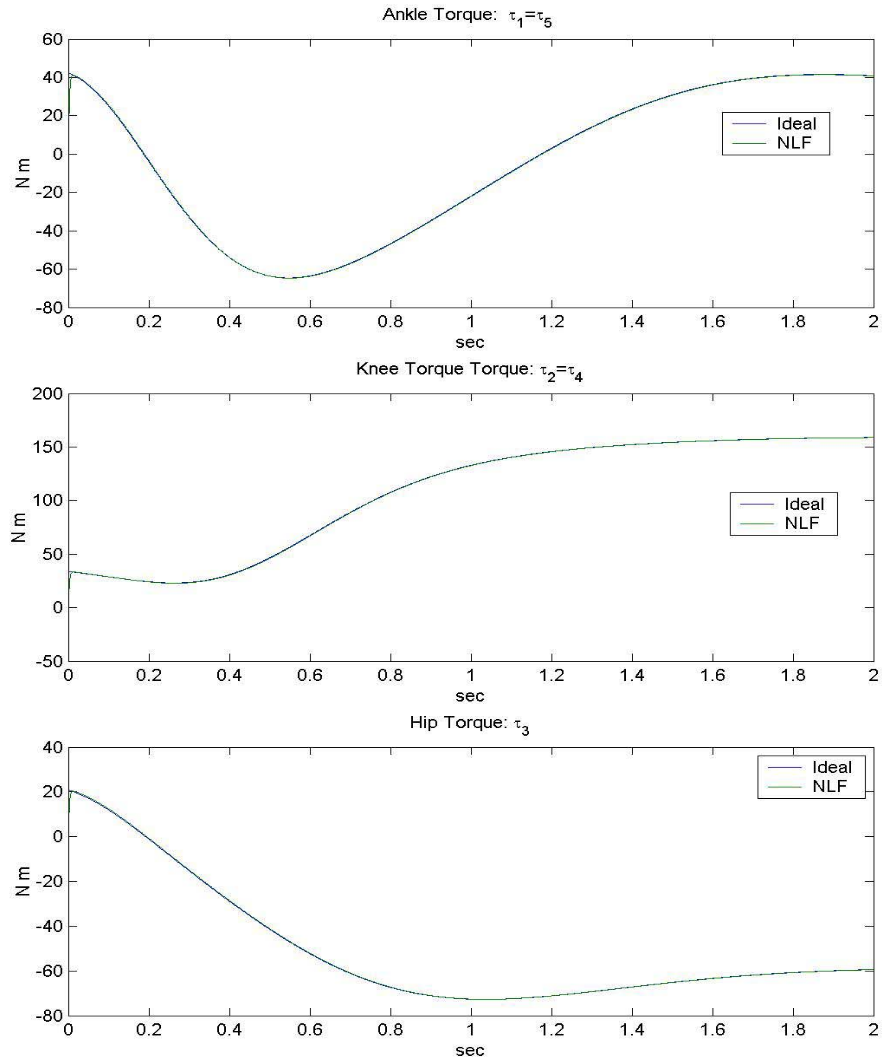


Figure 7 Simulation of the joint torques during a squatting maneuver without employing the desired accelerations (a=0). The proposed method using nonlinear feedback (NLF) produces nearly identical joint torque estimates as compared to the ground truth (ideal) joint torques obtained by a noise free inverse dynamics computation.

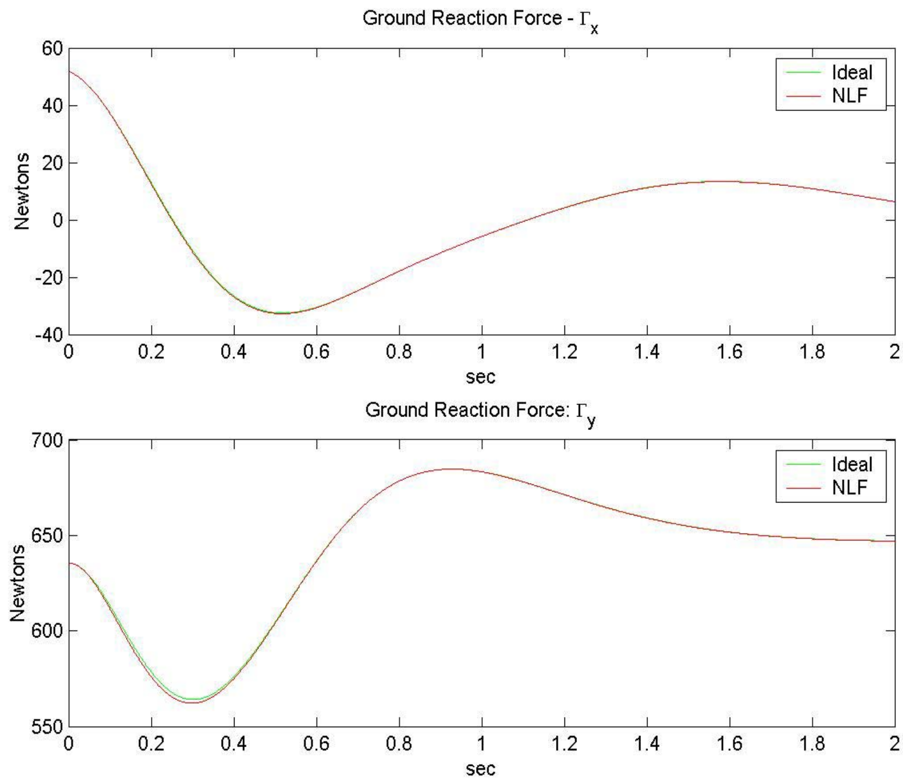


Figure 8 Simulation of the horizontal and vertical ground reaction forces during a squatting maneuver without employing the desired accelerations (a=0). The proposed method using nonlinear feedback (NLF) produces nearly identical ground reaction estimates as compared to the ground truth (ideal) ground reaction forces obtained by an Iterative Newton Euler inverse dynamics procedure.

6. CONCLUSION

A computational framework for analysis and synthesis of control strategies in human augmentation devices has been presented. Simulations of a squatting maneuver demonstrated the viability of the tracking methods proposed in this paper. The most notable feature of the approach is that acceleration estimates of kinematic data, which is often corrupted by noise, is not required to produce a forward simulation that will track and reproduce the desired kinematic data.

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