Gait & Posture 30 (2009) 518–522

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Model-based assessments of the effects of age and ankle fatigue on the control of upright posture in humans

Xingda Qu [a,*], Maury A. Nussbaum [b], Michael L. Madigan [c]

[a] School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore
[b] Department of Industrial and Systems Engineering, Virginia Tech, Blacksburg, VA, USA
[c] Department of Engineering Science and Mechanics, Virginia Tech, Blacksburg, VA, USA

ARTICLE INFO

Article history:
Received 27 March 2009
Received in revised form 21 July 2009
Accepted 30 July 2009

Keywords:
Balance
Postural control
Age
Ankle fatigue
Model-based simulation

ABSTRACT

The aim of this study was to investigate how and why age and localized muscle fatigue affect postural control using model-based simulations. A balance control model, based on an optimal control strategy, was used to simulate trials of quiet upright stance both pre-fatigue and following induced ankle plantarflexor fatigue. Empirical data were obtained from an earlier study that included both younger and older participants. Effects of age and ankle fatigue were determined from center-of-pressure (COP) measures and fitted model parameters. Though some discrepancies existed, the simulated effects of age and ankle fatigue were consistent with experimental findings in terms of trends in COP-based measures with age and ankle fatigue. Changes in both COP-based measures and model parameters were used to infer potential underlying causal mechanisms for the observed effects of age and ankle fatigue. For example, the model-based simulations indicated that sensory delay time increased with age and ankle fatigue by 31.1% and 2.9%, respectively, suggesting a potentially important role for such delay in postural control and fall risks.

1. Introduction

Falls are common events in both daily and occupational activities. It was reported that more than 44% of all injuries in 2000 resulted from falls or fall-related incidents [1]. Many falls are subsequent to a "loss-of-balance". Hsiao and Simeonov [2] summarized a number of risk factors that could compromise balance and highlighted the need for better understanding the roles of each factor in the control of balance. Among these diverse risk factors, the present work emphasized two particular intrinsic factors: age and localized muscle fatigue (LMF). These factors are considered of importance since falls occur more frequently in older adults, and LMF is commonly experienced in both daily activities and occupational settings.

Both age and LMF affect the way in which humans control balance and posture, and the quality of the postural control mechanisms. For example, aging can decrease strength [3] and the speed of a response to loss of balance [4], both of which result in an increased fall risk [5]. LMF appears to challenge the postural control system, as evidenced by increased COP mean velocity during upright stance [6,7]. Fatigue induced in simulated occupational tasks, such as repetitive lifting [8] and overhead assembly work [9], has also been found to impair postural control.

As an adjunct to biomechanical and neurophysiological studies, mathematical models are often used to study postural control and balance [10,11]. Such models have the advantage of suggesting how the entire system functions and how individual components influence the overall system's response [12]. Thus, model-based simulation may provide some understanding of underlying postural control mechanisms.

We recently presented a balance control model, based on an optimal control strategy, which can simulate spontaneous sway behaviors during quiet upright stance [13]. In the current study, the purposes were to evaluate the ability of the model to simulate age and LMF effects, and to employ the model to investigate how and why age and LMF affect postural control. Based on prior evidence, it was hypothesized that the model would be able to accurately simulate postural changes caused by age and LMF, and that simulation results regarding postural control mechanisms would be consistent with previous findings.

2. Methods

2.1. Participants and experimental procedures

Experimental data were obtained from a prior study [14]. As such, the methods and procedures are only presented summarily here, and the reader is referred to the noted publication for details. Thirty-two individuals (16 males and 16 females)

* Corresponding author at: School of Mechanical and Aerospace Engineering, Nanyang Technological University, Blk N3, North Spine, Nanyang Avenue, Singapore 639798, Singapore. Tel.: +65 67904458; fax: +65 67924062.
E-mail address: xdqu@ntu.edu.sg (X. Qu).

0966-6362/$ – see front matter © 2009 Elsevier B.V. All rights reserved.
doi:10.1016/j.gaitpost.2009.07.127

without self-reported injuries, illness, or musculoskeletal disorders participated in the study. Equal numbers of younger (18–24 years) and older (55–65 years) participants were included, intended as representative of those at the beginning and end of working life, and were balanced across genders.

Trials consisting of quiet, bilateral upright stance were conducted both prior to and after fatigue, which was induced in several muscle groups through repetitive, isokinetic, submaximal exercises. Each experimental session involved fatigue at only one joint, and a minimum of two days separated consecutive sessions to avoid residual effects of fatigue at other joints. Only pre- and post-fatigue trials involving fatigue at the ankle were used for three reasons. First, ankle control torque appears to be dominant in postural control during spontaneous sway [15]. Second, our existing balance control model [13] was used, in which the human body is represented by a single inverted pendulum. Thus, the control input to this model is the ankle control torque, and is likely to be affected by ankle fatigue. Third, ankle fatigue has been found to have substantial effects on postural control, as assessed by COP- and center-of-mass (COM)-based measures, and these effects have been evident in response to both bilateral [7,16] and unilateral exertions [14].

A commercial dynamometer (Biodex Medical Systems, Shirley, NY) was used to generate fatigue in the ankle plantarflexors through performance of repetitive (12/min), concentric isotonic exertions at 60% of individual maximum voluntary capacity. Exertions were terminated, and fatigue considered to have been induced, when participants could not complete a preset range of motion (45°) for three consecutive repetitions. The mean (SD) durations of the fatiguing ankle exercises were 26.3 (10.9) and 28.3 (13.9) min for the older and younger adults, respectively.

Three trials of quiet upright stance were conducted prior to the fatiguing exertions, one of which was randomly selected for analysis here. Several post-fatigue trials were performed, the first of which commenced 45 s after the fatiguing exertions and were studied here. During these trials, quiet upright stance was maintained for 75 s while participants stood as still as possible on a force platform (AMTI OR6-7-1000, Watertown, Massachusetts, USA), with barefoot, arms at sides, feet together, and eyes closed. Triaxial ground reaction forces and moments were sampled at 100 Hz, low-pass filtered (5 Hz cut-off, 2nd order, Butterworth) in software, and subsequently used to derive COP time series.

2.2. Model description

Our existing balance control model [13] was used to predict the effects of age and ankle fatigue under the conditions described above. In this model, the human body was represented by a single-segment inverted pendulum with feet during quiet upright stance, and the neural controller was assumed to be an optimal controller. Anthropometric measures required by the model were either directly measured or estimated (see [17]). The optimal neural controller generated ankle control torques, and was determined by an infinite-time linear quadratic regulator that minimizes a performance criterion (Eq. (1)) defined by several physical quantities relevant to sway:

$$J = \frac{1}{2}\int_0^{\infty}(w_1\hat{\theta}^2(t) + w_2\dot{\hat{\theta}}^2(t) + w_3T^2(t) + w_4\dot{T}^2(t) + w_5T''^2(t))\,dt \tag{1}$$

where $\hat{\theta}$ is the delayed ankle angle; T is the ankle torque; and w_1, w_2, w_3, w_4, and w_5 are weightings of the respective relevant physical quantities. Several model parameters cannot be specified *a priori*, specifically the weightings within the performance criterion (Eq. (1)), as well as random disturbance gain (k_n) and sensory delay time (τ_d). An optimization procedure was performed to determine the values of these unspecified model parameters, using the following cost function:

$$E = \sum_{i=1}^{N}\left(\frac{COPM_i - \hat{COPM}_i}{\hat{COPM}_i}\right)^2 \tag{2}$$

where $COPM_i$ and $\hat{COPM}_i$ are the ith COP-based measure from the simulation results and from the experimental results, respectively. Heuristic approaches (i.e. a genetic algorithm (GA) and simulated annealing (SA)) were implemented to determine model parameters [18]. Simulation results from this model have been reported to lead to the same general conclusions with experimental findings in terms of age-related changes in postural control [13], the effects of external loads [17], and the importance of passive and active control of quiet upright stance [19].

2.3. Dependent COP-based measures

A diverse set of measures was desired to represent different aspects of postural control. Using time-domain COP-based measures, numerous studies have demonstrated that aging and LMF increase postural sway [7,20]. Frequency-domain COP-based measures can reflect altered periodicity which is an indicator of reduced physiological functional ability [21] possibly caused by aging and LMF. Statistical mechanics measures have been proposed [22] to account for the dynamic characteristics of the postural control system [23] and were used to identify differences with respect to age and LMF [7,23]. From such observations, we earlier selected eight COP-based measures (including two time-domain, two frequency-domain, and four statistical mechanics measures) to provide a diverse representation of postural control [17]. This same set of measures was used here to investigate the effects of age and ankle fatigue (Table 1). When calculating these measures, the initial 10 s and last five seconds of each trial were removed.

Table 1
Glossary of COP-based dependent measures.

Acronym	Description	Unit
RMS	Root mean square distance	mm
MV	Mean velocity	mm/s
CFREQ	Centroidal frequency	Hz
FREQD	Frequency dispersion	–
TT	Transition time	s
TA	Transition amplitude	mm^2
H_S	Short-term scaling exponent	–
H_L	Long-term scaling exponent	–

2.4. Model simulation and analysis

Sixty-four experimental trials of upright stance were simulated using our balance control model (one pre-fatigue and one post-fatigue trials for each of the 32 participants). Each simulation trial was 75 s in duration, with the initial 10 s and last five seconds removed. Simulations involved several steps (see [13] for a detailed description): 1) a set of initial values was assigned to the unspecified model parameters; 2) the human body dynamics, sensory systems, and optimal neural controller were determined; 3) postural sway was simulated; 4) an optimization procedure was used to optimize the cost function through iterative adaptation of the model parameters. Upon completion of the optimization, simulated COP time series and model parameters were retained for subsequent analyses.

Two-way repeated measures analyses of variance (ANOVA) were used to identify effects of age and ankle fatigue ($p < 0.05$) on the control of upright posture. Specifically, for each of the experimental and simulated COP-based measures of postural sway, and model parameters, ANOVA was performed once with age (young vs. old) and ankle fatigue (pre-fatigue vs. post-fatigue) as independent variables. Identified effects were used to posit how and why age and LMF affect postural control. For example, if sensory delay is found to increase with aging, it can be inferred that changes in postural control due to age may be partly caused by an increased sensory delay. To evaluate the performance of the model, model-based predictions and experimental results were compared to identify whether trends in COP-based measures with respect to age and fatigue status were consistent between experimental and simulation results.

3. Results

3.1. Age and ankle fatigue effects on COP-based measures

For the experimental results, significant main effects of age were found on MV, CFREQ, FREQD, TT, and H_L (Table 2). More specifically, MV and CFREQ increased, while FREQD, TT, and H_L decreased with age (Fig. 1). For the simulated measures, a similar pattern of age effects was found with the exception that the effect on MV was not significant ($p = 0.065$). Ankle fatigue led to significant changes only in experimental MV, which increased post-fatigue. A significant age × ankle fatigue interaction effect was found only for experimental H_S. When ankle fatigue was induced, experimental H_S decreased in the older group (from 0.829 to 0.801), but increased among younger participants (from 0.795 to 0.810). In contrast to the effects on the experimental COP-based measures, only simulated H_s significantly changed with ankle fatigue.

3.2. Comparison between experimental and simulated measures

Comparisons of the influence of age and ankle fatigue on the experimental and simulated measures indicated that almost all the simulated (i.e. model-based) trends were consistent with experimental findings (Fig. 1). Clear discrepancies were only evident in the effects of age on TA, and the effects of ankle fatigue on CFREQ and H_S. Most of the significant age and ankle fatigue effects were simulated by the model, as summarized above (Table 2), with exceptions found for MV and H_S. More specifically, the simulated data did not reveal the main effects caused by age and ankle fatigue on MV, and the age × ankle fatigue interaction effect on H_S.