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Investigating Human Balance Using a Robotic Motion Platform

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***Abstract*—We present the system design for a novel robotic balance simulator that enables the investigation of the balance mechanisms involved in natural human standing. Our system allows for complete control of task dynamics to mimic normal standing while avoiding the pitfalls associated with applying external perturbations. The system enables subjects to balance themselves according to a programmable physical model of an inverted pendulum. Subjects were able to balance the system, and results show that the load stiffness curves approximate those of normal human standing to within 20.1 ± 9.7% (S.D.). Differences were within the range expected from control loop delay, reduced ankle motion, and approximations inherent to the inverted pendulum model.**

I. Introduction

THE human body integrates sensory information from the vestibular (located in the inner ear), visual, and somatosensory (skin, muscle, and joint receptors) systems to maintain upright balance. Although normal balancing requires minimal voluntary effort by a healthy adult, this motor behaviour may pose significant challenges for elderly populations [1] or persons suffering from neuromuscular impairments, including incomplete spinal cord injury, stroke, or Parkinson's disease. Deficient balancing behaviour can limit a person's ability to stand upright as well as perform more complex tasks such as walking.

In spite of its prevalence, the method by which the human body integrates sensory information to produce an appropriate motor command for balance is not well understood [1]-[3]. Physiologists aim to understand exactly how sensorimotor integration works during balance control, and accurately characterize the balancing behaviour of those with balancing deficits. It is hoped such research will allow medical professionals to diagnose neurological impairments and design more effective rehabilitative exercises.

Traditionally, there have been two key methods of investigating human balance control: by measuring changes in system behaviour due to a controlled change in sensory feedback, and by measuring compensatory reflex response to an applied perturbation [3]-[5]. Both methods have several limitations. In the first case, adjustment of sensory inputs causes a change in the statistical properties of the task itself by changing a subject's perception of how the task operates. This has been shown to cause adaptation in sensorimotor control systems [4], [6], indicating a reorganization of the fundamental control scheme used to perform the task. In the second case, large or sudden perturbations of the balance system (e.g., a sudden push, or an auditory beep), may introduce large compensatory mechanisms that are not necessarily active during normal standing [5].

If a subject could perform a balance simulation task that engages the same neural pathways as during natural standing, one could better investigate the processes involved in normal balance. This paper presents the system design for a novel robotic device that simulates a natural standing task. We aim to develop a robotic simulator capable of adjusting balancing dynamics in order to avoid the confounding effects caused by a change in the statistical task environment [5], [7], [8]. Robotic simulation provides full control over dynamic balancing parameters, so that we can investigate sensorimotor integration under a variety of sensory inputs and balancing parameter configurations. The first goal of this project is to create a system that can mimic natural standing for any subject. This is achieved by matching the relationship between ankle torque and standing angle (i.e., the load stiffness curve) for a subject [5], [9].

The robotic balance simulator is also developed for rehabilitative applications. Current rehabilitative therapy often focuses on activating and strengthening the muscles involved in balance control. However, this style of treatment may fail to engage core neural processes that drive muscle behaviour. This is particularly relevant to balance behaviour which requires less cortical drive compared to voluntary activation of the same muscles [10]. Hence, traditional rehabilitation may be inadequate for relearning or improving balance control. The Lokomat® by Hocoma AG is an example of robotic assistive device that improves locomotion training for patients with spinal cord injuries by mimicking normal gait. Standing and walking, however, involve fundamentally different neural processes and at present there is no equivalent device that assists human balance training. One robotic balance device presented by Takahashi et al. [11] detects a subject's body position and automatically restores posture. However, this device does not require the subject to actively control their balance. In contrast, the robotic balance simulator presented herein can engage users in a customizable balancing task to facilitate

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proper motor learning.

In Section II we present the balance simulation control loop, which includes the inverted pendulum model used to replicate human balance physics. In Section III we describe the mechanical, electronic, and software components used to implement such a system. In Section IV we detail the experiments used to verify system capability, and present and discuss the results. In Section V and VI we conclude with a discussion of the future work related to the balance simulator, and review the advantages offered over previous balance investigation devices.

II. Balance System Model

The robotic balance simulation system is designed to replicate the physical sensation of human balance in the anteroposterior pitch (forward-backward) direction. Prior work has shown that the balancing physics of human standing are closely modeled as an inverted pendulum [5], [9], [12]. An inverted pendulum (see Fig. 1b) is an unstable mechanical system in which a near-vertical rod (assumed to have zero mass) has a rotational pin joint at its bottom end and a large mass at its top. A balancing task is simulated by having test subjects experience inverted pendulum-like dynamics, and attempt to balance the system by modulating their ankle torque.

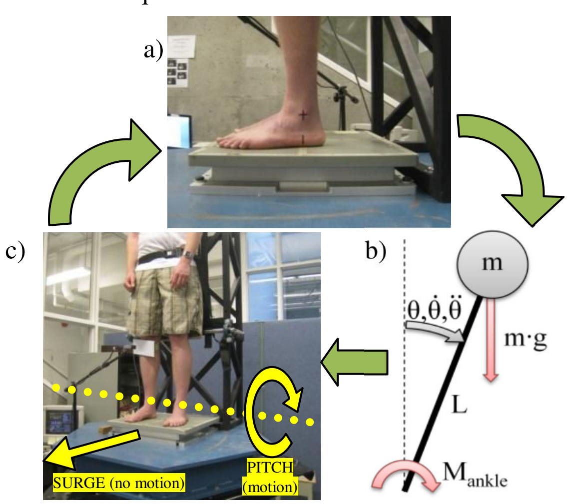

Fig. 1. Control loop for balance simulator (clockwise from top): a) forceplate measures ankle torque applied by subject; b) a computer calculates motion and position of an inverted pendulum with the same torque applied; c) the motion platform moves in pitch direction to match the position of the inverted pendulum; control loop repeats as subject moves with platform and adjusts ankle torque.

As shown in Fig. 2, a subject stands on top of a motion platform and is securely fastened to an adjustable backplate rigidly mounted on the platform. A forceplate beneath the subject's feet measures the torque applied by both ankles. The ankle torque information is sent to a computer that interprets the torque information as if it were applied to an inverted pendulum. The computer calculates the rotational motion of such a system, and commands the motion platform to move in response to this torque. The subject experiences the motion of the platform, and reacts by adjusting ankle torque in order to restore the system to unstable equilibrium. Fig. 1 schematically shows the complete control loop.


Fig. 2. Test subject standing on balance simulator.

Fig. 3 shows a block diagram representation of the inverted pendulum model implemented in software.

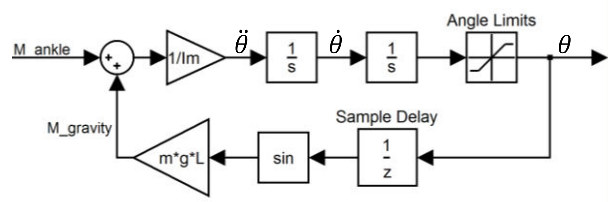

Fig. 3. Inverted pendulum system implemented in software.

Parameters m and L are the mass and length, respectively, of the desired pendulum model, and g is the gravitational constant 9.81 m/s^2. For the experiments performed, m and L were the subject's mass and distance from ankle to hip (approximate centre of mass). $I_m=m*k^2$ is the mass moment of inertia, where k is the radius of gyration. For an inverted pendulum, the radius of gyration is equal to the centre of mass length (i.e., $k=L$). For the purposes of this work we also make this assumption, which is consistent with the methodology of [3]-[5], [7], [8], [12], and [13]. Further work is required to revise this estimation for a human subject. M_{ankle} is the externally applied torque, equal to the torque applied by a subject's ankles. $M_{gravity}$ is the moment applied due to gravity. As shown in Fig. 3, the acceleration of the inverted pendulum is calculated as

$$\ddot{\theta} = \frac{(M_{grav}+M_{ankle})}{I_m} = \frac{(m_m g L_m sin\theta+M_{ankle})}{I_m} \quad (1)$$

and angular position (θ) is derived by integrating twice. The rotation axis is programmatically set to pass through the subject's ankles. Pitch angle is constrained to stay safely within configurable limits.

III. System Setup & Design

A. Mechanical Equipment and Instrumentation

The robotic balance simulation system is composed of a

6-axis motion platform, forceplate, and multiple driving computers. A laser analog sensor is used to measure pitch angle during experiments.

A MOOG 6DOF2000E motion base capable of 500 deg/s^2 acceleration actuates rotational motion in the anteroposterior pitch direction. The motion base uses 6 ball-screw belt-driven linear actuators with an onboard computer and position control loop. The motion platform has a rigid back support that can be adjusted to rest against all subjects in their normal standing posture (see Fig. 2).

All real-time computations, data acquisition, and communication with the motion base are performed by a National Instruments PXI-8196 embedded controller and PXI-6289 DAQ board, with BNC-2090A connector block and PXI-1031 chassis. Data communication occurs over a 100 Mbit/s dedicated network connected to the motion base, PXI terminal, and an additional host PC terminal. Reaction forces and moments beneath a subject's feet are recorded using an AMTI OR6-7-1000 6-axis forceplate with AMTI MSA-6 amplifier unit. An A-Tech LM100 Laser Analog Sensor (70 µm resolution) measures balancing angle.

B. Software Driver and Data Compensation

The driver software manages data acquisition, computation of balancing task physics, and 60 Hz data communication with the motion base. Software is written in the LabVIEW 8.5 graphical programming environment.

1) Forceplate Data Compensation

The forceplate measures forces and torques applied by a test subject's feet. Since the forceplate is fixed to the motion platform it is also affected by platform acceleration and displacement. Fig. 4 shows the relevant geometry, motion, forces, and moments affecting forceplate measurements. The moment applied at the ankle is computed from an analysis of forceplate mechanics.

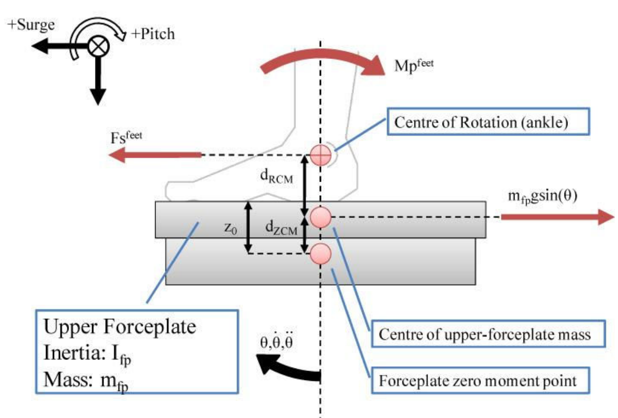

Fig. 4. Relevant geometry, motion, forces, and moments affecting the forceplate.

With the effective zero moment point (ZMP) of the forceplate located below the top surface of the plate, forces acting in the surge direction (see Fig. 1c) generate a measurable moment in the pitch direction. Surge forces occur due to three different effects:

1. a subject's feet applying shear stress forward or backward,
2. a component of gravity pulling on the upper forceplate at non-zero pitch angles, and
3. a reaction force due to linear acceleration of the upper forceplate mass.

Equation (2) relates these components to the measured force in the surge direction.

$$F_S^{measured} = F_S^{feet} - m_{fp}a - m_{fp}gsin(\theta) \tag{2}$$

Each of the surge force components acts at a particular distance from the ZMP, generating a moment in the pitch direction. In addition to the surge force effects, the measured pitch moment is also a function of:

1. the moment applied by a subject's feet, and
2. the reaction moment due to rotational acceleration of the upper forceplate mass.

These forces and moments are related according to (3):

$$\begin{aligned} M_P^{measured} = &-F_S^{feet} * (d_{ZCM} + d_{RCM}) \\ &+ \left(m_{fp}a + m_{fp}gsin(\theta)\right) * d_{ZCM} + M_P^{feet} \\ &- I_{fp}\ddot{\theta} \end{aligned} \tag{3}$$

By commanding the motion platform along sine wave trajectories in the pitch axis, with different subjects fastened in place, we found that the moment applied by the feet was a function of both ankle torque and a gravity moment due to approximately 2% of the subject's weight involuntarily applied to the forceplate. We saw no measurable effect due to the subject's rotational inertia (i.e., the backplate and straps bore all the reaction moment due to the subject's inertia). Therefore, the moment applied by the feet on the forceplate was estimated as:

$$M_P^{feet} = -M_{ankle} + 0.02m_mgL_m\sin(\theta). \tag{4}$$

The distance of the ZMP below the top surface of the forceplate, z_0, is a calibrated value provided by the manufacturer. The mass of the upper forceplate, m_{fp}, was measured by zeroing the forceplate, holding it upside down, and dividing the measured vertical force by $2g$. The distance, d_{ZCM}, from the centre of mass of the upper forceplate to the ZMP was measured by moving the motion platform (with unloaded forceplate on top) to various static pitch angles and measuring the generated pitch moment. Linear acceleration of the forceplate was computed as:

$$a = \ddot{\theta} * d_{RCM}. \tag{5}$$

The distance, d_{RCM}, from the centre of rotation to the centre of mass of the upper forceplate is a function of the subject's measured ankle height (as shown in Table I).

PARAMETERS FOR FORCEPLATE DATA COMPENSATION

m_{fp}	=	15.1 kg
I_{fp}	=	0.23 kg·m^2
z_0	=	0.043 m
d_{ZCM}	=	0.030 m

Combining (2) through (5) leads to the implicit solution for M_{ankle} that is applied to the pendulum model as shown in Fig. 3. For trials where the motion platform and forceplate remain in a fixed horizontal position (relative to ground), all acceleration and angle terms are zero, and (3) simplifies to

$$M_P^{measured} = -F_S^{measured} * z_0 - M_{ankle}. \tag{6}$$

2) Data Flow & System Delay

Forceplate data are acquired at 2 kHz and filtered using a second-order, low-pass Butterworth filter with 5 Hz cutoff. The data are down-sampled to match the control rate determined by the motion platform (nominally 60 Hz). The delay between position command and position feedback (for motion within actuator velocity and acceleration limits) was measured to be 7 sample periods, or 117 ms. The delay term shown in Fig. 3 causes a software computational delay of one sample period, or 17 ms. The delay due to forceplate input filtering was measured to be 3 sample periods, or 50 ms. Summing all components, the aggregate delay of the balance control loop was found to be 183 ms.

IV. EXPERIMENTS

Experiments were performed to assess the ability of the balance simulation system to replicate load stiffness curves for natural standing.

A. Methodology

Six healthy male subjects participated in this study. Measured physical data are presented in Table I. Each subject's centre of mass (CoM) was approximated to be located at the anterior superior iliac spine.

All subjects were barefoot during the experiment. Pitch angle was measured using a laser distance sensor sampled at 60 Hz (see Fig. 5), and computed according to (7):

$$\theta = -\tan^{-1}\left(\frac{x_{ankle} - x_{\text{laser}}}{x_{\text{leg}}}\right) \tag{7}$$

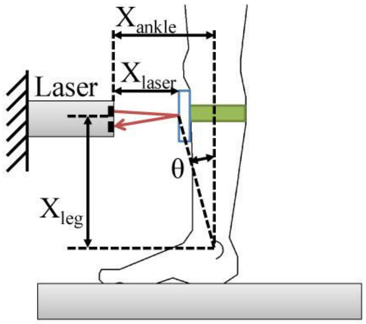

Fig. 5. Angle measurement using laser distance sensor.

During experiments, a laser-reflective white surface was fixed below the subject's kneecap using an elastic strap, and the laser was positioned approximately 7cm horizontally away from the surface.

TABLE I: PHYSICAL DATA FOR TEST SUBJECTS

Subject ID	Age	Mass [kg]	Length Ankle-CoM [m]	d_{RCM} [m]
A	27	59.98	0.815	0.078
B	24	76.94	0.940	0.098
C	32	77.25	0.905	0.083
D	24	82.09	0.920	0.093
E	24	72.70	1.000	0.093
F	23	71.77	0.905	0.088

1) Load Stiffness during Natural Balancing

Subjects were instructed to stand still in a normal, relaxed position with the motion platform stationary and the backplate moved out of contact with their body. The forceplate was zeroed with the subject standing in this relaxed position. The angle measured at this time was used as the zero-reference position. Subjects were then instructed to sway forward and backward within a comfortable, self-determined range without lifting toes or heels from the forceplate. The torque versus angle relationship (i.e., load stiffness of the human body) was recorded over 5 full periods of sway.

2) Load Stiffness during Balance Simulation

The subject was securely fastened to the backplate using a seatbelt-type strap placed around the chest and waist (Fig. 2). Subjects were instructed to keep their feet planted in the same location immediately following the natural balance experiment. In this experiment, load stiffness was determined with the balance simulator engaged and the subject actively controlling the position of the motion platform as described in Section II.

Prior to measuring load stiffness, subjects were given up to 15 minutes to familiarize themselves with balancing on the simulator. Subjects were instructed to balance normally with no data being recorded, until they were comfortable with the task and able to balance the motion platform without hitting angle limits (6° anterior, 3° posterior, from vertical) for at least 30 seconds.

Load stiffness was determined from the torque versus angle relationship recorded as subjects balanced on the simulator. Subjects were instructed to rotate the motion platform forward and backward in a slow controlled manner for 5 full periods of (pitch) rotation.

B. Load Stiffness Results

Raw data for the best and worst matched load stiffness (between natural balance and balance simulation) are presented in Fig. 6. Subjects A and F balanced a load that was respectively 1.3 Nm/deg (14.9%) and 4.0 Nm/deg (35.4%) stiffer during the balance simulation than during natural balance. Both subjects showed greater variability

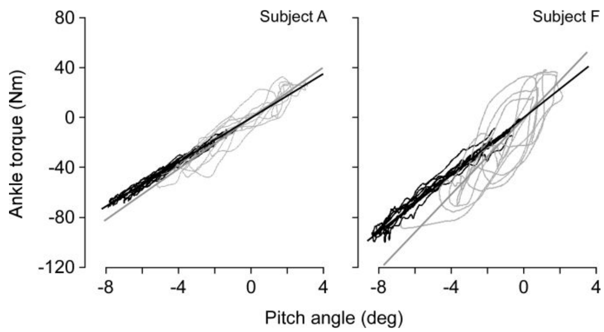

Fig. 6. Raw data for subjects A and F. Load stiffness during natural balance is shown in black and during balance simulation is shown in grey.

during the balance simulation. Table II summarizes the raw load stiffness results obtained for all subjects. As described in the previous section, the zero angle corresponds to the relaxed standing angle of each subject. Torque values plotted in Fig. 6 are the negative of M_{ankle} (Fig. 3), such that load stiffness values presented in Table II and Fig. 7 are positive.

Across subjects, the normalized mean load stiffness was 12.9 ± 0.4 Nm/deg for natural balance and 15.4 ± 1.1 Nm/deg for the balance simulation condition (Fig. 7), a difference of 2.6 ± 1.2 Nm/deg (20.1 ± 9.7%).

TABLE II: SUMMARIZED LOAD STIFFNESS RESULTS

	Natural Balance		Balance Simulation	
Test Subject	**Load Stiffness (Nm/deg)**	**R^2**	**Load Stiffness (Nm/deg)**	**R^2**
A	8.8	0.98	10.2	0.92
B	13.2	0.98	17.0	0.75
C	13.2	0.99	15.1	0.73
D	13.1	0.97	15.4	0.70
E	13.4	0.97	14.7	0.70
F	11.3	0.97	15.3	0.77

C. Discussion

Subjects were able to balance successfully on this simulator based on an inverted pendulum model, with similar load stiffness curves. We believe the differences between the balance simulation results and the baseline natural standing results may be due to reduced ankle proprioception, passive stiffness effects, control loop delay, and approximations in the underlying physical model. Since the relative angle between feet and legs was held constant, ankle proprioception was limited to force feedback from muscle contraction and skin sensation, rather than muscle spindle afferent signals that code for muscle length and velocity. The reduction of sensory feedback could cause greater variability, as suggested by [7]. Moreover, during normal standing, ankle motion stretches the calf muscle-tendon unit to produce passive stiffness [9], [14], [15], thereby reducing the amount of torque that must be actively produced by the ankle in order to overcome gravity. In this study, passive stiffness had no effect during the balance simulation as the ankle was held in a fixed position. Therefore, larger motor units may have been recruited and muscle firing frequency may have increased in order to generate higher force output. This would lead to a reduction in fine motor control and may account for some of the increased variability observed in this study.

The greater variability in load stiffness during the balance simulation may also be due to the 183 ms control loop delay described in Section II. The presence of loops in the load stiffness data (Fig. 6) suggests hysteresis is occurring; Delayed motion feedback could cause subjects to perpetually overshoot the torque required to hold the motion platform at a desired angle, leading to oscillatory corrective behaviour instead of controlled linear motion.

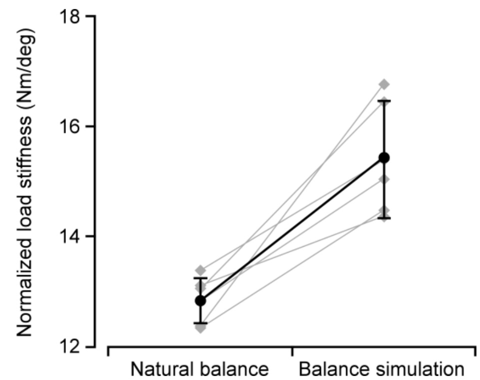

Fig. 7. Normalized load stiffness results for all 6 subjects (grey) with group data shown as mean ±S.D. (black). Data was normalized to each subject's predicted load stiffness: *mgL* (in Nm/deg), where *m* = subject mass, *L* = ankle to centre of mass length, and *g* = gravitational acceleration (9.81 m/s^2); Normalized data was then multiplied by the group mean load stiffness from the natural balance trials.

As described in Section II, the inertia of the physical model was calculated as $I_m=m*k^2$ with the radius of gyration, k, equal to the length between the subject's ankles and approximate centre of mass, L. Since the calculation for mass moment of inertia integrates the squared distance from rotation point to mass distributed along a person's height, the true radius of gyration is likely greater than L. Prior investigation using an inverted pendulum model found that it was valid for standing motions with low accelerations where inertial effects could be neglected [9]. This research also showed similarly large load stiffness variability when subjects balanced a real inverted pendulum. Therefore, when calculating inertia of the underlying model, the true radius of gyration calculated for a given subject may produce a more linear load stiffness curve. If the inertia were increased, as would be the case for $k>L$, accelerations would be reduced and standing sway would be expected to decrease.

V. FUTURE WORK

A. Balance Simulation Development

Continued development of the balance simulation system includes reducing the overall control loop delay and incorporating passive ankle stiffness and viscous damping.

We also plan to investigate the validity of the inverted pendulum model for describing human balance dynamics. By analyzing torque versus angle data for natural human standing we plan to identify the transfer function describing the human balance plant, and determine empirical values for inertia, viscous damping, and static load stiffness. A mechanical system identification of the MOOG motion platform will be used as a basis for feedforward predictive model control in order to reduce the effective delay from the motion platform.

B. Future Experiments

Future experiments will involve a wider range of subjects of different ages and genders. We also plan to utilize additional sensory input instrumentation to further study the human balance system:

1. Electrodes can be fixed behind a subject's ears in order to electrically stimulate the vestibular system and provide a pure vestibular error. Vestibular stimulation can be provided using galvanic or stochastic currents [16], [17].
2. A 160-degree parabolic display (Elumens VisionStation) can be used to control a subject's visual input. An Ascension Technologies Flock of Birds® sensor can track a subject's head and the visual display can be adjusted to simulate the effect of an immersive 3-dimensional environment.
3. A motorized foot stabilization platform can be added to adjust foot orientation and control ankle proprioception.

The 6-axis robot allows for multi-axis balance motions, including the ability to select and change sway direction as desired. Future tests may utilize this capability to study human balance control under unnatural or changing task mechanics; e.g., ankle torque in the anteroposterior (pitch) direction could be used to control balance motions in the mediolateral (roll) direction.

VI. Conclusion

In this paper we report on a robotic simulation system developed for investigating human balance control. The system enables subjects to balance according to the physics of an inverted pendulum model with configurable physical parameters. To the best of our knowledge, this is the first system to enable true simulation of a balancing task that allows researchers to study the mechanisms of human balance control in a configurable task environment without applying external perturbation. We have achieved the first goal in development of the simulator by showing that test subjects with different physical parameters can balance on the system and generate a load stiffness curve that approximates natural standing. Passive stiffness, reduced ankle proprioception, system delay, and approximations in the balance simulation model may account for the increased variability and load stiffness observed during balance simulations. Future development will focus on reducing control loop delay and improving the system model used to govern balance dynamics.

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