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Gait stability assessment within a patient-cooperative lower limb exoskeleton

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Abstract: Gait stability and user safety are crucial aspects in order to use assistive exoskeletons not only for rehabilitation therapy but also for mobility support in daily life. For this reason, we present a gait stability assessment approach based on the Center of Mass and the Zero Moment Point. The designed balance observer successfully detects falls in the forward and backward directions as they occur. In the future, gait stabilizing controls will be designed based on this approach.

I. Introduction

Active exoskeletons are increasingly being used in rehabilitation therapy. Within this therapy, gait safety and fall prevention are typically ensured by therapists, handles on treadmills, or support structures on the ceiling. However, outside of this controlled environment, e.g. during gait assistance in daily life, the risk of falling increases significantly. Most available exoskeletons, therefore, require the support of crutches or other mobile walking aids to ensure balance. Although the stability of human gait, on the one hand, and of bipedal robots, on the other hand, has been studied widely, there is little literature on analyzing and controlling the gait stability of coupled human-robot systems. The research that does exist on this topic focuses mainly on '*exoskeleton in charge*' systems [1]–[3], where the free movements of the patient are neglected. Nevertheless, risk management, inherent stability, and fall interception in assistive exoskeletons are of great importance to minimize the risk of patient injury. Additionally, Shirota et al. compared balance assessment capabilities for various rehabilitation robots, including exoskeletons, and concluded that '*robotic devices are promising and can become useful and relevant tools for assessment of balance in patients with neurological disorders, both in research and in clinical use*.' [4].

For this reason, we present a real-time capable approach to assess gait stability by detecting different kinds of falls while walking with a 6 Degrees of Freedom (DoF) exoskeleton. The gait stability analysis constitutes a preliminary work for gait stabilizing control methods.

II. Material and methods

II.I. Modeling of User and Exoskeleton

We developed the gait stability assessment approach for a generic exoskeleton by introducing a Denavit Hartenberg model with 18 DoF (**Fig. 1**, **left**) representing the main joints of the human's lower body. However, the approach was implemented and tested for the Lower Limb Exoskeleton with Serial Elastic Actuators (L^2Exo-SE, **Fig. 1**, **right**) [5]. The L^2Exo-SE provides unilateral assistance for hip and knee flexion/extension, e.g. to freely assist the gait movement of hemiplegic patients. Additionally, both legs are equipped with high-resolution encoders in the ankle, hip, and knee joints. Moreover, inertial measurement units (IMUs) are available on both thighs and the upper body using a chest strap.

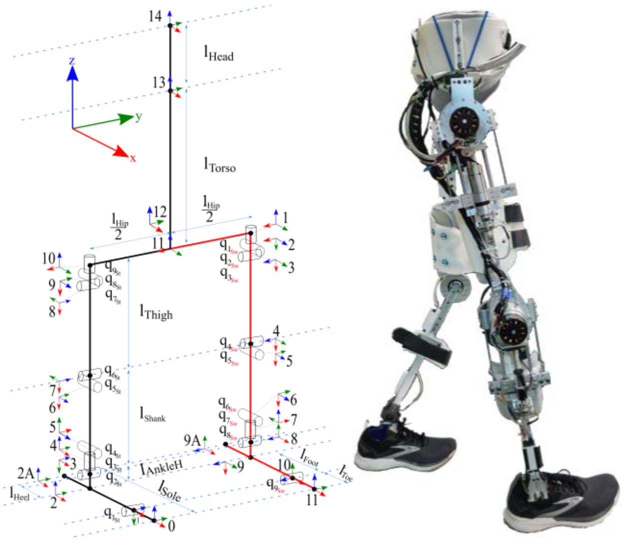

Fig. 1: Left: DH Model with 18 DoF. Right: Lower Limb Exoskeleton with serial elastic actuators and 6 DoF.

The base frame of the derived DH model changes during the gait and is defined by the toes of the foot with the highest ground reaction forces (**Fig. 1**, **left**, black leg). The other leg is considered to be in the swing phase (**Fig. 1**, **left**, red leg). Due to the constraints of the exoskeleton, we can consider certain DoFs of the generic DH model as fixed, leaving 12 DoFs. The remaining DoFs are: 2x toe deflection angle, 2x Hip/Knee/Ankle flexion/extension, 2x hip rotation, and adduction/abduction.

II.II. Assessing Gait Stability

Based on the derived DH model we were able to derive each segment's position ${}^0\boldsymbol{p}_i = {}^0(x,y,z)_i^T$ with respect to the base frame. Subsequently, the total center of mass of the exoskeleton-human system can be derived to:

$$ {}^0\boldsymbol{p}_{CoM} = \sum_{i=1}^{n} \frac{m_i \cdot {}^0\boldsymbol{p}_i}{m_{Total}} \qquad (1)$$

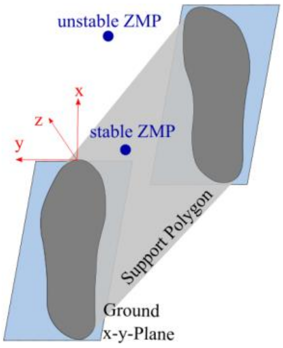

Fig. 2.: Visualization of the support polygon (SP) and a stable and unstable ZMP.

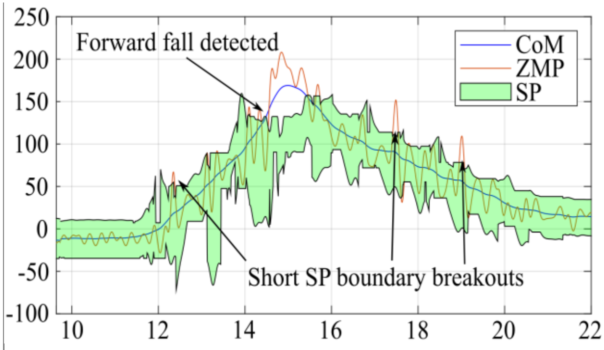

Fig. 3: Derived CoM, ZMP and SP area in x direction. t<12s: Standing; t=[12-14.45]s walking forward; t=[14.45-15.5]s; falling forward; t >15.5s walking backward.

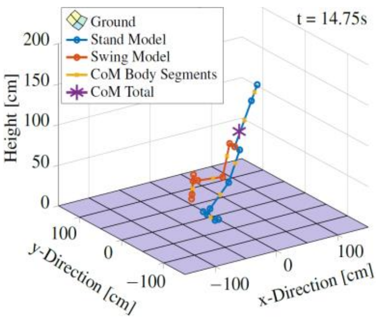

Fig. 4: Pose of falling test person derived from exoskeleton sensors.

Where m_i and m_{Total} describes the individual segment masses and the total mass, respectively. The CoM can be used to describe the **static stability** of the system. Static stability is achieved when the vertical projection of the CoM on the ground is within the so-called base of support [6]. The BoS is defined by the surface of the foot in the single stance phase and by the polygon between both feet (support polygon, SP) in the double stance phase. The support polygon is visualized in **Fig. 2**.

To determine the **dynamic stability**, we utilized a well-known technique invented for the control of humanoid robots: The Zero Moment Point (ZMP). The ZMP describes the point on the ground under the foot about which the sum of all torques exerted by a biped robot is zero. It can be calculated by

$$\begin{pmatrix} x \\ y \end{pmatrix}_{ZMP} = \begin{pmatrix} x \\ y \end{pmatrix}_{CoM} - \frac{z_{CoM}}{g} \begin{pmatrix} \ddot{x} \\ \ddot{y} \end{pmatrix}_{CoM} \tag{2}$$

With g being the earth's gravitational constant and z_{CoM} the height of the CoM [7]. If the ZMP is inside the SP, the dynamic stability of the system is guaranteed (see example ZMP in **Fig. 2)**.

II.III. Data Acquisition and Processing

The angular positions required to determine the CoM in eq. (1) were obtained by the angular encoders and the IMUs of the L^{2}Exo-SE. The toe joint deflection angle is estimated based on the kinematic chain and assuming an even floor. To derive the second derivative of the CoM in the x and y direction in eq. (2) we used discrete derivatives with additional low-pass filters (IIR, 3rd Order). The cut-off frequency ($f_c = 2.5$ Hz) was selected to incorporate up to the 7th harmonic frequency of the leg segment movements during a gait velocity of 0.5 km/h. The CoM, SP, and ZMP were calculated on a real-time computing unit at a sampling frequency of 200 Hz.

III. Results and Discussion

To validate the developed balance observer, a test subject (male, 30 years, 170cm, 66 kg) walked with the L^{2}Exo-SE in transparency mode (no active assistance) for three steps and then fell forward/backward. The test person was caught in the fall by two assistants. Afterward, the test subject was walking a few steps backward. The calculated CoM, ZMP, and SP in the x-direction are shown in **Fig. 3** for the forward fall test. It can be seen that at the time of the forward fall (at 14.45 s), both the CoM and the ZMP leave the range of the SP in the positive x-direction indicating the fall in the forward direction. The pose of the falling subject at time 14.75s derived from the angular sensors is shown in **Fig. 4.** Both forward and backward falls can be successfully detected in the course of the falling.

However, the calculation of CoM's acceleration utilizing the second derivative and low pass filtering leads to a trade-off between robustness and detection time. In this study, the walking speed of 0.5 km/h was relatively low allowing for a low cut-off frequency. Nevertheless, high angular accelerations due to non-perfect alignment between the user and exoskeleton lead to peaks in the ZMP estimation, and thus to brief exits from the SP region even for the stable gait condition (see **Fig. 3**).

Despite the limitations, we demonstrated that the ZMP can in principle be used for gait stability assessment in assistive exoskeletons with free subject movements.

IV. Conclusion

In this study, we utilized the CoM and ZMP to assess static and dynamic gait stability within an assistive exoskeleton. The CoM is calculated based on an 18 DoF (12 DoF with restricted movement) DH model where the base frame is changed in accordance with the standing leg. The ZMP was derived from the CoM. The presented approach detects both forward and backward falls as they occur. In the future, we would like to enable detection for higher gait speeds as well as design control approaches for gait stabilization based on our balance observer.

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