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Active Hip Orthosis for Assisting the Training of the Gait in Hemiplegics

Dejan B. Popović[1], Aleksandar Veg[2]

Abstract: The gait restoration is a significant component of the rehabilitation of patients with hemiplegia after central nervous system lesion. The typical symptom of the disability is the lack of hip extension during the stance phase and hip flexion during the swing phase of the affected leg; therefore, asymmetry and perturbed balance while walking. We developed a new modular orthosis for assisting hip flexion and extension during the therapeutic sessions of the gait in hemiplegic individuals based on thorough analysis of the gait characteristics of normal gait. The system is designed to fit into the powered postural controller Walkaround®. The system is developed based on the estimation of torque and power requirement for the gait at speeds up to 1.1 m/s. The model used for the simulation is a known double pendulum in the sagittal plane. The kinematics and dynamics used as the input for the simulation have been recorded in the gait laboratory with eight cameras and force platforms. The system needs to be clinically evaluated before it possibly can be turned into innovation.

Keywords: Hip orthosis, Modular orthosis, Walkaround®, Gait modeling, Hemiplegia.

1 Introduction

Hemiplegia is a disability which follows the central nervous system (CNS) and has several symptoms which decrease the quality of life. One of the signs is the reduced mobility. Recent research suggests that intensive exercise of a skill contributes to the recovery of function after CNS lesion since it adds to the cortical plasticity [1]. The most prominent mobility limitation is the drop foot which prevents normal swing of the leg required for the gait. The drop foot symptom can be eliminated by a mechanical ankle-foot orthosis or an electrical stimulation system applied to activate the ankle flexors when necessary [2]. Here, we address the most significant change of the motor systems that eventually will be minimized by an exoskeleton applied at the hip.

To elaborate the importance of the hip assistance, we review essential features of the gait [e.g., 3 – 5]. Most studies related to bipedal gait consider the

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basic operation (i.e., rhythmic displacement of body segment resulting in the progression over a level ground). The changes in speed of the gait, the direction of progression, and slope/stairs negotiations are considered as the variations from the basic pattern [6].

If the model of a human is reduced to a single rigid body, then the gait can be analyzed as a sequence of: 1) a pendulum-like swinging of the body about a point above the head, and 2) the inverted pendulum motion about the contact point of the sole with the ground (Fig. 1).

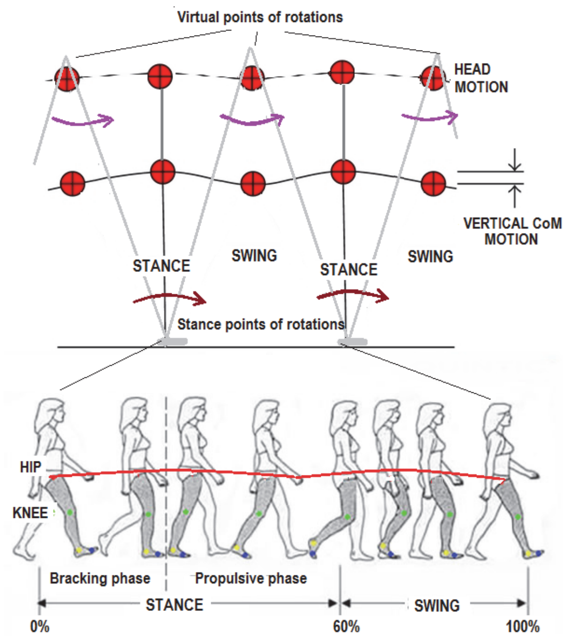

Fig. 1 – *The ballistic model of the gait.*

During the inverted pendulum phase of the gait, the potential energy is rising while the kinetic energy drops decreases. During the pendulum phase, the kinetic energy increases, while the potential energy declines. The early stance phase is characterized by the horizontal component of the ground reaction force opposite to the direction of the progression (slowing down phase). The late stance phase is characterized by the horizontal ground reaction force in the direction of the progression (acceleration phase) as shown in Fig. 2.

The muscles contract to push the body center of the mass against the gravity to compensate for the energy used (transformation into heat) during the slowing down of the center of mass and impact with the ground at the beginning of the gait cycle. In the pendulum-like phase of the gait cycle, the body swings with synchronous flexion of the swinging leg to ensure the ground clearance.

Cappozzo [7] suggested that all parts of the body are engaged in the generation of what is called "normal gait." The head, neck, and trunk have to be taken into consideration, even though the task is to control the legs.

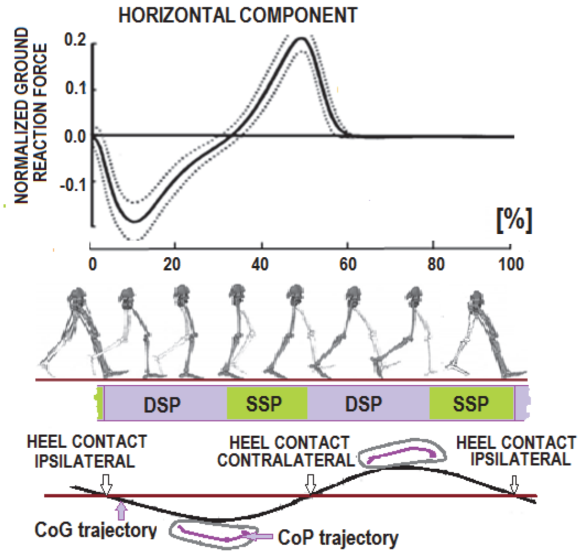

Fig. 2 – *Horizontal component of the ground reaction force in the sagittal plane for one gait cycle (top panel), and trajectories of the center of gravity (CoG) and center of pressure (CoP). The acronyms DSP and SSP are for the double and single support phases of the gait cycle.*

This greatly simplified model was developed based on the assumption that the bipedal gait is steadily recovery stability by having a ballistic behavior [8−14]. Electromyographic recordings (intramuscular electrodes) proved this assumption [8]. The muscles are active during the double support period (DSP) (Fig. 2). Mochon and McMahon [12] have simulated the ballistic movement of the swing leg for arbitrary chosen initial conditions for a ground level walking. Their model has three links, one for the stance leg and two for the thigh and shank of the swinging leg. The mass of all body parts above the hips was lumped as the point at the hip joint, and the masses of the lower limb segments were distributed more realistically. This model has been extended for flexion of the stance knee, and pelvic tilt [13]. They pointed out that the last two are necessary to obtain agreement between the predicted and observed vertical ground reaction forces. McGeer [10] has demonstrated, by both computer simulation and physical model construction, that some anthropomorphic legged mechanisms can exhibit stable, human-like gait on a range of slopes with no actuation and no control. A version of double-pendulum ("compass-gait") walker having point-feet has come from Goswami [15]. Garcia et al. [16] proposed an improved model with straight legs and point-feet which eliminated the constraints of the models described. Van der Linde [14] used a straight-legged walker with curved feet and variable joint stiffness as an internal energy supply to compensate for energy losses due to damping, friction, and collisions without destroying the passive ballistic walking cycle [9].

All of the mentioned models show the essential feature of the bipedal gait where the motion of the pelvic region is the source of human-like gait. The trunk of the body must move in a manner that guarantees that the CoP (the point at which the ground reaction force acts on the sole) and the CoG (projection of the center of mass to the ground) move synchronously from the ipsilateral to the contralateral sole to secure balance. Humans tend to minimize the acceleration of the head; hence, they control the movements of the pelvic region to satisfy the balance task.

The stroke patients often cannot assume the appropriate posture for the gait (Fig. 3, right panel).

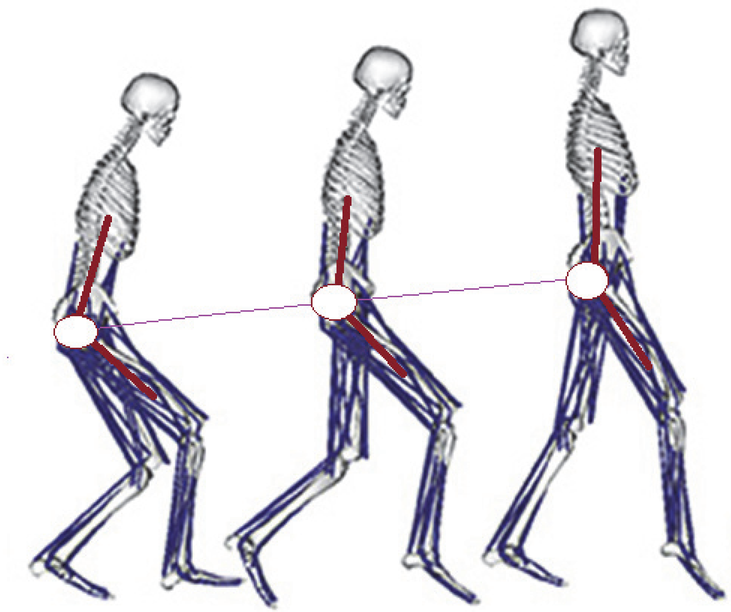

Fig. 3 – *The sketch of the crouched posture (left) and the progression of the recovery (middle and right).*

The crouched posture and crouched gait are the consequences of the lost control for full extension of the supporting leg during the stance phase and the constant pushing of the hip forward before the swing phase. The crouched posture prevents normal gait and transforms the dynamic process of optimized used of the gravity into the quasi-static ambulation. The crouched gait eliminates the periods during which the CoG is moving quickly, shortens the step length and reduces the number of steps per minute.

2 Materials and Methods

Veg and Popović [17] suggested a solution by introducing the walking assist termed Walkaround®. The Walkaround® is a powered walker providing the postural orientation to the trunk by a lumbar belt, preventing from the fall, and pushing the hips in the direction of the gait. The lumbar brace is a multi-segment chain of flat rigid plates interfacing the trunk via an inflatable lumbar cuff that fits the contours of the trunk and distributes the contact forces under the complete cuff. The system allows the gait at rates up to 1.1 m/s. However, the first Walkaround® does not provide the relative motion of the pelvic region with respect the frame which is the prerequisite for the normal gait. This motivated the integration of an actuation of the lumbar belt (Fig. 4) for moving the pelvic region [18].

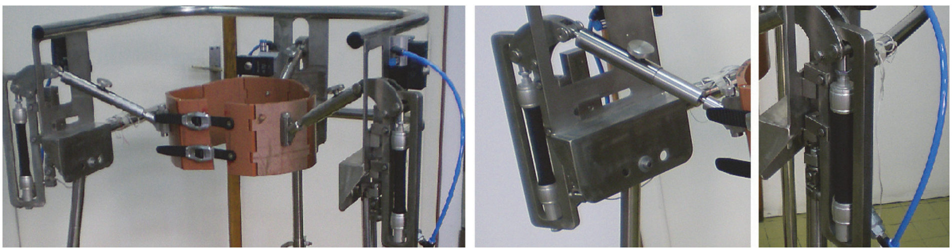

Fig. 4 – *The actuated lumbar belt of the Walkaround® (left). The rigid plates are connected with hinges, and they form a belt (left panel). Visco-elastic mechanisms (middle panel) are driven by FESTO linear actuators (right panel).*

The operation of three actuators acting in three directions (one from the front and two from the lateral back sites) is controlled by a microcontroller PIC 18F452L from Microchip©. The lengths sensors signals provide data for sensors driven control of valves (lengths of the actuators). The force sensors integrated into the frame of the Walkaround® in series with the actuators provide signals that guaranty that the forces are within safety limits. The desired butterfly pattern of pelvic region is shown in Fig. 5.

The operation of the system was tested in three hemiplegic subjects. The motion of the center of mass was within the 2.1 cm in the vertical direction, and the forward/backward movement was about 2.9 cm. The controlled movement of the pelvic region increased the symmetry of gait. The ratio of double support phase vs. the stride cycle reached 0.9 ± 0.1 compared to free walking.

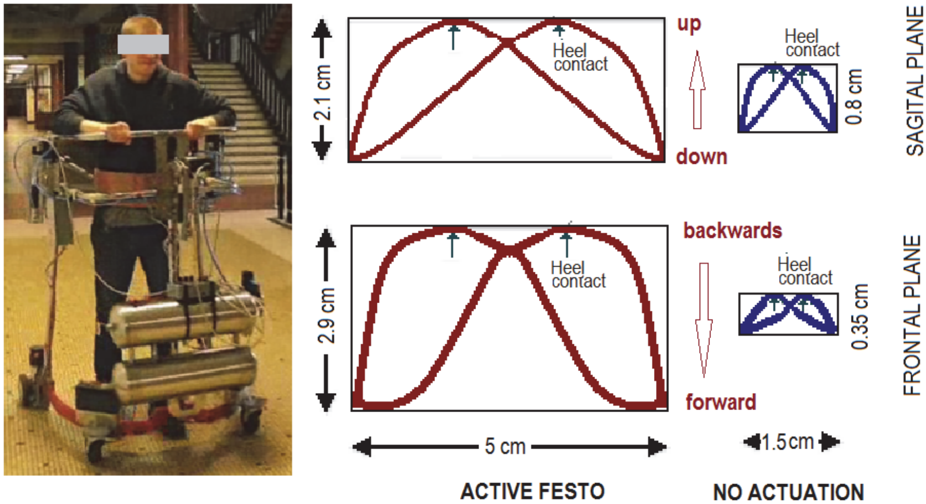

Fig. 5 – *Left panel is the photograph of the prototype of the system. Right, panels show the trajectories that were selected for targets (one gait cycle). Modified from [18] with permission.*

However, the design with linear pneumatic actuators was not accepted well enough by the clinicians and patients for several reasons: the noise, size of the pneumatic tanks, limited distance on a single charge were stated as technical problems, but the incentives were the perturbations that were not integrated appropriately to fit the biological timings and amplitudes of motions. The acceptance can be summed as the opinion from all subjects who tested the system: "I do not feel safe when walking."

This led to the development of an alternative assistance that provides the control of the hip joint by a light modular orthosis. The orthosis is connected to the belt and have telescopic sliders as the interface with the thigh of the patient. The assumed system is shown in Fig. 6.

This design followed the original development of the self-fitting modular orthosis (SFMO) many years ago. The SFMO was used with some success in several rehabilitation institutions around the world [19 – 21] and led to many other exoskeleton developments. The new orthosis has a little friction bar that slides in the slider being fixed softly to the body (e.g., tight trousers) [21].

The development of the modular orthosis followed conventional methods that use the simulation of the gait to estimate the characteristics of the optimal actuation. We used the simulation package developed in our earlier work [22]. The method used experimental data collected in the Gait Laboratory. The gait kinematics and dynamics for the simulation were recorded from a healthy subject (H = 1.73 m, m = 72 kg) in the Human Performance Laboratory at the

Center for Sensory-Motor Interaction, Aalborg, Denmark. We used the optoelectronic system (8 ProReflex MCU240 cameras, Qualisys AB, Sweden) and a force plate (OR6-5, Advanced Mechanical Technology, US). Motion capture and force plate data were acquired synchronously with the sampling frequency of 100 Hz using a standard desktop computer. Reflective markers were placed on the anatomical landmarks of the subject's pelvis and right leg, and rigid clusters of markers were strapped to the thigh and shank. The subject was first recorded in the anatomical position (standing straight, palms forward) for the purpose of calibration. After the calibration trial, the subject was instructed to walk normally in a straight line along the lab walkway. Only those trials in which the subject stepped on the force plate with his right foot were considered for the simulation. The recorded marker trajectories were labeled and gap filled using the Qualisys Track Manager and then exported into the Visual 3D (C-Motion, US). The absolute segment angles (pelvis, thigh, shank, and foot) in the sagittal plane were computed from the marker trajectories and low pass filtered using a second-order zero phase Butterworth filter with a cutoff frequency of 6Hz. Angular accelerations of the segments were obtained by filtering (2nd order, zero phase, Butterworth, 6 Hz) and differentiating two times the acquired segment angles by using the third order spline interpolation. Vertical and horizontal ground reaction forces and the trajectory of the center of pressure were estimated from the force plate data.

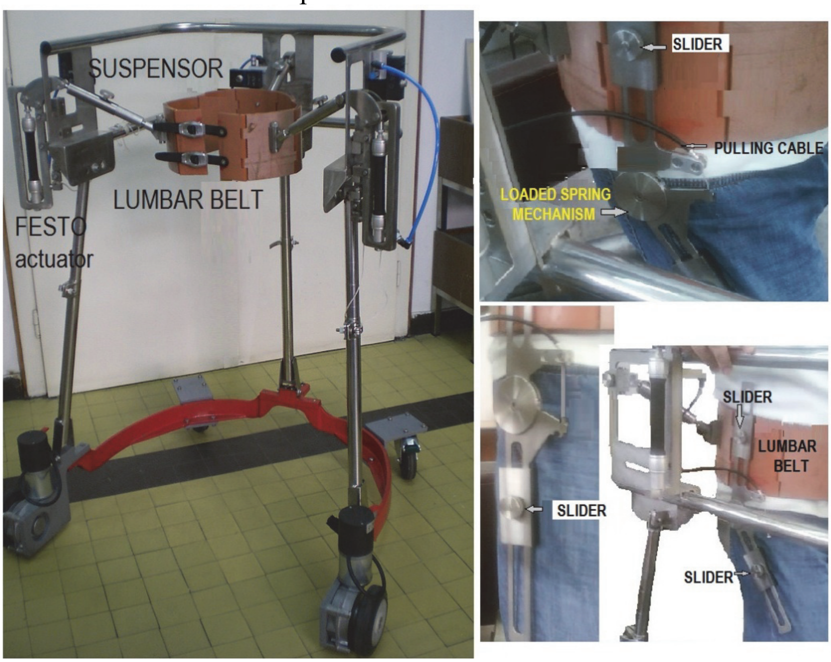

Fig. 6 – *The prototype of the modular hip orthosis fitted into the Walkaround® developed for Tecnalia Inspiring Business, San Sebastian, Spain.*

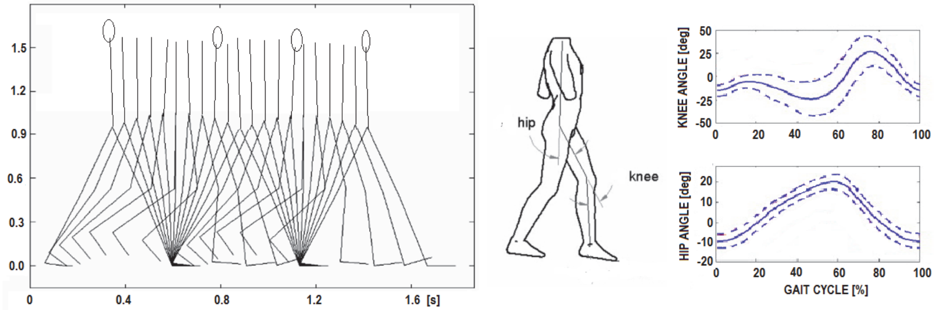

Fig. 7 – *Stick diagram and hip and knee joints during the gait at* 0.6 m/s *recorded in the gait laboratory with the Qualisis AB, Sweden at Aalborg University, Denmark.*

The simulation uses the model of the single leg with two degrees of freedom controlled by actuators. The leg is assumed to be a double, planar pendulum with a moving hanging point. The remaining part of the body was represented by the single force acting at the hanging point (hip joint) and the hip joint torque. The model does not include the ankle and phalangeal joints. The kinetic energy E of the leg is (1):

$$\begin{aligned}2E = 2E_{BP} &+ m_T((\dot{x}_H + d_T\dot{\varphi}_T\cos\varphi_T)^2 + (\dot{y}_H + d_T\dot{\varphi}_T\sin\varphi_T)^2) + \\ &+ J_{CT}\dot{\varphi}_t^2 + m_S((\dot{x}_H + L_T\dot{\varphi}_T\cos\varphi_T + d_S\dot{\varphi}_S\cos\varphi_S)^2 + \\ &+ (\dot{y}_H + L_T\dot{\varphi}_T\sin\varphi_T + d_S\dot{\varphi}_S\sin\varphi_S)^2) + J_{CS}\dot{\varphi}_S^2,\end{aligned} \tag{1}$$

where BP stands for the biological part of the system, H for the hip, T for the thigh, S for the shank, C_T and C_S are the centers of the mass of the thigh and shank respectively, m is the mass, L is the length, d is the distance from the proximal joint to the mass center and J is the moment of inertia for the segment, relative to the corresponding axes through the center of mass perpendicular to the sagittal plane. Applying the Lagrange principle, we obtain the second-order nonlinear, time variable equations of motion in the form (2):

$$\begin{aligned} A_{11}\dot{x}_2 + A_{12}\cos(x_3 - x_1)\dot{x}_4 &= F_1 + u_1, \\ A_{21}\dot{x}_4 + A_{22}\cos(x_3 - x_1)\dot{x}_2 &= F_2 + u_2 - u_1, \end{aligned} \tag{2}$$

where the state and control vectors are defined by (3):

$$\begin{gathered} x_1 = \varphi_S, \quad x_2 = \dot{x}_1, \quad x_3 = \varphi_T, \quad x_4 = \dot{x}_3; \qquad x = (x_1, x_2, x_3, x_4), \\ u_1 = M_K, \quad u_2 = M_H; \qquad u = (u_1, u_2). \end{gathered} \tag{3}$$

The notations used in (2) are the following:

$$
\begin{aligned}
&A_{11} = J_{CS} + m_S d_S^2 \quad \wedge \quad A_{12} = m_S d_S L_T, \\
&A_{21} = J_{CT} + m_S L_T^2 + m_T d_T^2 \quad \wedge \quad A_{22} = m_S d_S L_T, \\
&F_1 = A_3 x_4^2 \ \sin\ (x_3 - x_1) + A_4 \ddot{x}_H \sin\ x_1 + A_5(\ddot{y}_H - g)\ \cos\ x_1 - M_K^r, \\
&F_2 = B_3 x_3^2 \ \sin\ (x_3 - x_1) + B_4 \ddot{x}_H \sin\ x_1 + B_5(\ddot{y}_H - g)\ \cos\ x_3 + M_H^r, \\
&A_3 = J_{CS} + m_S d_S^2, \quad A_4 = m_s d_S, \quad A_5 = A_4, \\
&B_3 = -m_S d_S L_T, \quad B_4 = m_s L_T + m_T d_T, \quad B_5 = A_4.
\end{aligned}
\tag{4}
$$

The M_{rK} and M_{rH} are the resistive torques at the hip and the knee ($J = K$). The air resistance, damping and elastic properties of the joints are represented by the resistive torque. Equation (4) considers the ground reaction forces during the stance phase (the forces are zero during the swing phase). The order of the differential equations (2) can be reduced since they do not include the time in the explicit form. The first order differential equations are (5):

$$
\begin{aligned}
\dot{x}_1 &= x_2, \\
\dot{x}_2 &= P_1 + G_1 u_1 + G_2 u_2, \\
\dot{x}_3 &= x_4, \\
\dot{x}_4 &= P_2 + G_3 u_1 + G_4 u_2,
\end{aligned}
\tag{5}
$$

where the terms are:

$$
\begin{aligned}
P_1 &= [A_{21} F_1 - A_{12} F_2 \cos(x_3 - x_1)] / C, \\
P_2 &= [A_{11} F_2 - A_{22} F_1 \cos(x_3 - x_1)] / C, \\
G_1 &= [A_{21} + A_{12} \cos(x_3 - x_1)] / C, \\
G_2 &= A_{12} \cos(x_3 - x_1) / C, \\
G_3 &= [A_{11} + A_{22} \cos(x_3 - x_1)] / C\ , \\
G_4 &= A_{11} / C, \\
C &= A_{11} A_{21} + A_{12} A_{22} \cos^2 (x_3 - x_1).
\end{aligned}
\tag{6}
$$

The values of the inertial parameters in Table 1 are from the literature [24].

Table 1

Body parameters used for simulation.

	J_C [kgm^2]	L [m]	d [m]	m [kg]	k_i, $I = 1, 2, 3, 4$ [Nm]
Shank (*S*)	0.062	0.54	0.25	5.05	0.01
Thigh (*T*)	0.084	0.44	0.25	7.1	

The simulation of the model (6) with the input (Fig. 7, right panels) results with the net joint torque shown in Fig. 8.

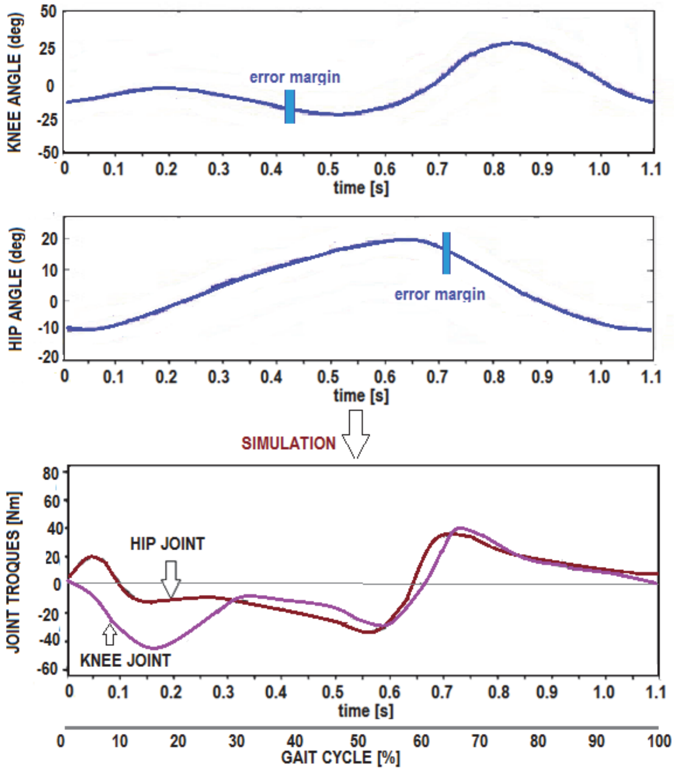

Fig. 8 – *The knee and hip joint torques required for the target gait shown in the upper panels. The error margin shows the region that has been determined when analyzing the ten consecutive strides. The horizontal axis shows the one gait cycle in time and %.*

3 Results and Discussion

The results of simulation directly provide information about the characteristics of the drive required in the hip joint modular orthosis. The joint torque needed for the target kinematics is within $M = \pm 25$ Nm. The orthosis needs to allow angular rate in the hip joints up to $\omega = 4$ rad/s. By analyzing in parallel the time course of the torque and the joint angular rate, we estimated that the power of the actuator at the output is $P \geq 30$ W.

The actual actuation unit of the hip module is based on the linear actuator for the flexion, and the loaded spring for the extension. The spring constant can be set individually. The complete control is with the linear actuator pulling the thigh segment into the flexion (Fig. 9). The actuator is active during the initial stance phase and at the end of the stance phase and beginning of the swing phase. The spring is being loaded during these periods, and it is released. The hold intervals are periods when the actuator is locked.

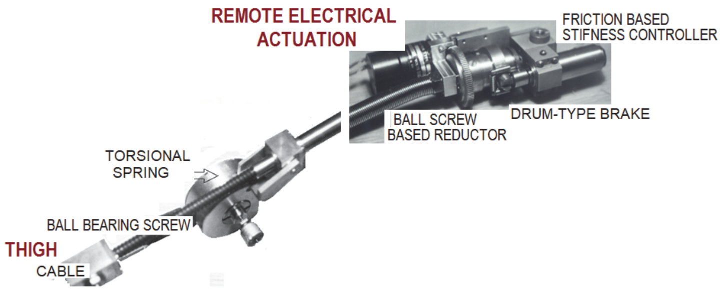

Fig. 9 – *The model of the new modular hip orthosis (replacing the FESTO) for the assistance of gait. The gearing uses the ball screw allowing back driving.*

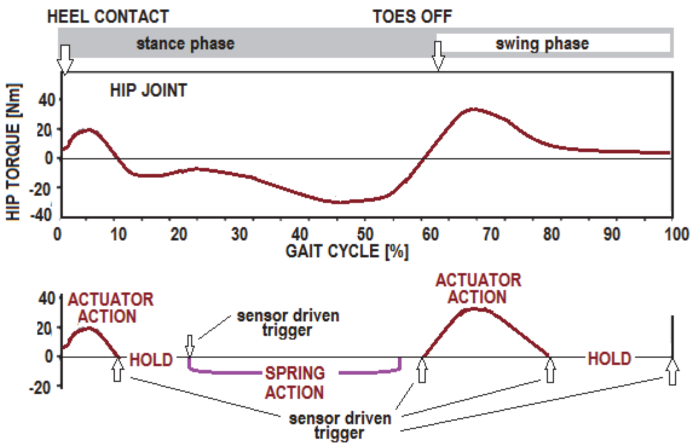

Fig. 10 – *The parallel presentation of the estimated hip torque and the joint torque generated by the actuators. The bell-shaped profiles are the consequence of the mechanical design (the actuator control signal is either on or off).*

Fig. 10 includes arrows indicating the on-off trigger. The signals are coming from the inertial sensors mounted on the leg assessing the kinematics (not described in this presentation).

Future works. The next phase of our research is to test the operation of the system in the clinical setting and potentially replace the current actuators with McKibben artificial muscles [25]. This development will follow only once when and if clinical tests provide results that the proposed innovation is of interest for clinicians and patients.

4 Acknowledgment

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