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Dynamic Analysis of a Stewart Platform for Lower Human Limb Rehabilitation Using Cad Tools

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Abstract

The following article shows the analysis made to a scale prototype of a Stewart platform, which is intended to be used in the rehabilitation of children with locomotion problems due to cerebral palsy. The platform was built with servo linear actuators, and the kinetic analysis executed in the "SolidWorks" software, indicated each actuator's required speed profile to follow a specific trajectory with the Stewart platform´s final effector. The force profiles for each actuator were also obtained, in order to verify that the movements obtained with inverse kinematics could be achieved by the actuators used in the project's final assembly.

Keywords: Stewart, Rehabilitation, Kinematics, Dynamics, CAD, Motion Analysis.

INTRODUCTION

Currently, a health problem that affects a large group of the population is infantile cerebral palsy, requiring the design of machines and systems that allow this sector of the population to have a life within the parameters considered normal. Infantile Cerebral Palsy (ICP) was described in 1889 by Olsen as a non-progressive neurological injury that occurs during pregnancy or at birth, with predominantly motion affectation [1].

IPC is described as a postural and motor disorder caused by a non-progressive aggression on a not-fully developed brain and is the most frequent cause of pediatric motor disability [2], with spastic diparesis and spastic hemiparesis being the most common types. Depending on the severity of the affectation, the individual will suffer locomotive system´s functional limitations ranging from posture alterations, weakness or muscle tone increase, muscle spasms, sensory and reflex decrease, orthopedic malformations, lack of limb control and lack of motor coordination, to name a few. Within the integral treatment for this disorder, functional and motor rehabilitation therapies acquire great relevance to improve quality of life in level I, II and III GMFCS scale patients, for whom, there is a favorable ambulation prognosis [3].

This condition is related to injuries on brain's high motor neurons, being a central nervous system's chronic disability characterized by an aberrant control of movement and posture, which appears at an early age and not as a result of a progressive neurological disease [4].

World's ICP incidence has been calculated from 2 to 2.5 per thousand live newborns [5]. In the United States, there are around ten thousand new cases every year, with this condition being frequent both in preterm newborns and full 9-month newborns [6]. In this regard, ICP is one of the main neurological pathologies that appear in Latin American children, and it is estimated that there are approximately twelve thousand new cases every year year in Mexico. Depending on its severity, PCI has been clinically classified as mild (without limitations in daily activities), moderate (difficulties in daily tasks, need for means of assistance or support) and major, profound or severe (with great limitations for performing daily activities) [7].

The most common cerebral palsy types (Fig. 1), are: Athetoid or atetoxic, with 5 to 10% of patients, which present involuntary movements that lead to abnormal postural control and the presence of "dance-like" movement while walking; the step pattern is high in flexion and low during the support phase, with internal rotation, flexed hip and hyper-extended lumbar spine; ataxic, with 5 to 10% of patients with awkward but voluntary movements and a coordination failure causing balance alterations, and a walking motion is characterized by a lack of coordination; Spastic, where 70 to 80% of the cases are found, shows a muscle tone increase leading patients to adopt abnormal postures and develop knife-type hypertonia, however, they can execute voluntary movements. In this type of paralysis, walkin motion presents patterns associated with increased muscle tone, producing spasticity that will cause incorrect limb support while standing and walking, such as the patient swinging the affected leg in circles (hemiplegia), "Scissor-like" walking (diplegia) and gait development with less affectation than patients with diplegia.

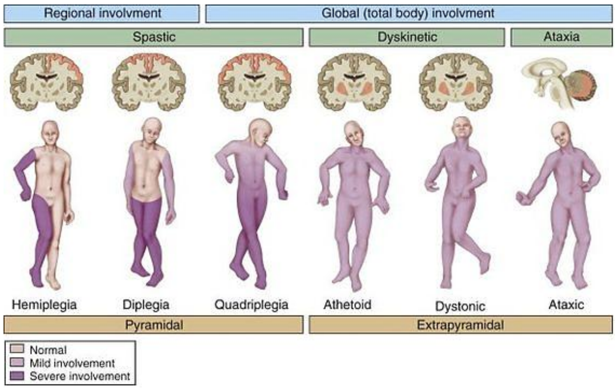

Figure 1. Types of cerebral palsy and affected brain regions

Locomotion or walking simulators, seek to give a feeling of real walk motion and have been object of research due to potential applications in rehabilitation of people with lower limb disabilities. Current gait simulation systems have been classified by Hollerbach [7] into three categories: on-site walking devices, treadmills, and foot platforms.

On-site walking devices

The general idea of these devices is to allow the person to walk naturally, without achieving any displacement, while the sensors in the device take the required measurements. The system captures the person´s walking movements, which are replicated by a virtual representation of the person. A system of this type is developed by Templeman [8], where magnetic followers are placed on the person's thighs. The movement's direction and speed are calculated from the movements of the person's knee. The orientation of the person's body is calculated through sensors arranged along the waist. Force sensors arranged on the feet soles detect the steps with greater accuracy (Fig. 2a).
Other works on similar devices were made by Lampton [9], which placed magnetic sensors on the ankles, and by Iwata [10], through the usage of low friction shoes and the anchoring of the person to the device's frame with a strap that ran around the waist (Fig. 2b and 2c).

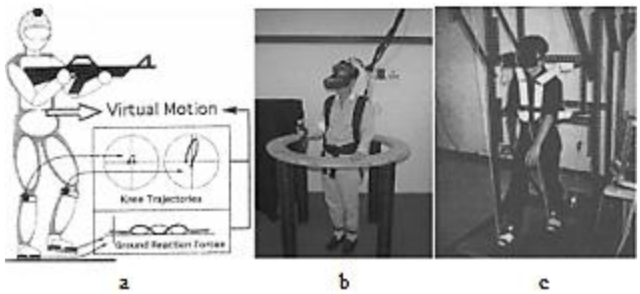

Figure 2: a) Immersion system's Simplified model b) The Fully Immersive Team Training. c) Early state of the Virtual Perambulator.

Treadmills

An example for this type of devices used in rehabilitation was developed at the University of Utah, and was called Sarcos Treadport [11,12]. The differential factor for this treadmill compared to its commercial counterparts, was the ascending surface simulation possibilites with inclinations beyond 5 ° (Fig. 3).

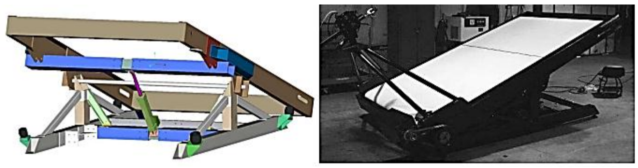

Figure 3. Design and real model of the Sarcos Treadport project.

Iwata also worked with treadmills to develop systems that would simulate human walking. His work focused on developing a treadmill that allowed a person to walk in any direction on a 2D plane [13,14]. This development, called the TORUS treadmill (Fig. 4), used 10 conveyor belts to achieve movement in any direction.

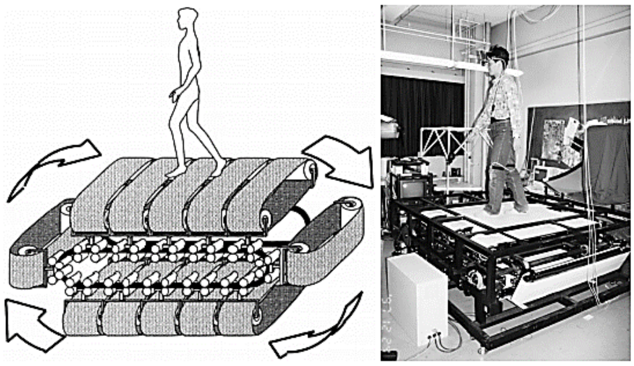

Figure 4. Design and real model for the TORUS system.

Miyasato developed a treadmill called ATLAS (Fig. 5). This treadmill had panels, which could be raised or lowered, in order to simulate irregular surfaces [15, 16].

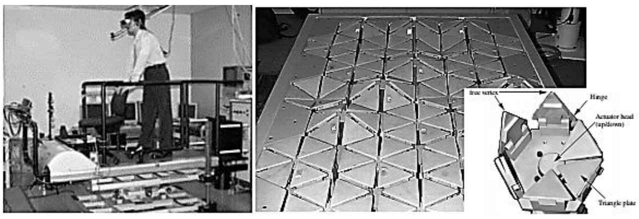

Figure 5. ATLAS system along with the mobile panels system.

Foot platforms

Iwata developed two generations of foot platforms called Gait Master and Gait Master_2 [17]. The first one (Fig. 6a), consisted of two platforms with three FD mounted on a rotating plate. These platforms followed feet´s movement through inputs given by sensors located on the person´s legs, thus providing an infinite virtual walk. The second generation in this line of platforms (Fig. 6b) was made with two movement platforms with two DOF. These platforms were operated by rotational motors and chains. Similarly, this system generated the feet's movement path thru sensors located on the person's legs.

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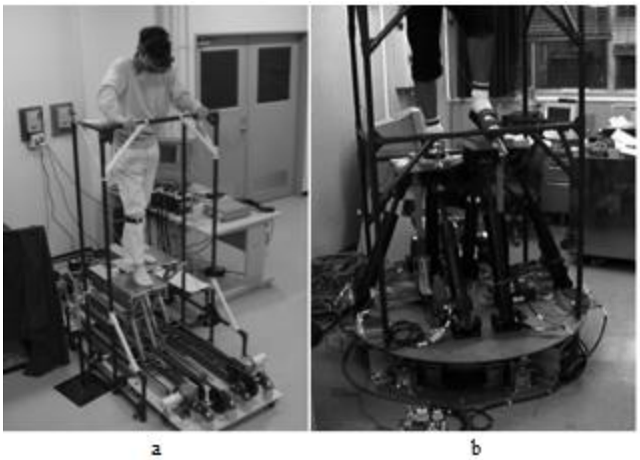

Figure 6. a) Gait Master system 1. b) Gait Master 2.

HUMAN WALKING

Biomechanics is the study of body movements and its effects, considering kinematic aspects (movements of body segments without considering forces that generate them [18]), kinetics (description of the internal forces from body segments) and dynamics (movements considering forces acting in body segments).

Walking is the most used locomotion method by humans to move from one place to another [19], with lower limbs and trunk's rhythmic alternating movements, which determine the forward displacement of the center of gravity with low effort and minimum energy consumption in a period of time called the walking cycle. Thus, the walking cycle (Fig. 7) is the time in which two successive identical events of the same foot occur, ie, the walking cycle begins when a foot is in contact with the ground and ends with the next contact of the same foot on the ground (stride); this cycle is divided into three important phases:

The stance phase is the time during which the reference foot is in contact with the ground, transferring the body's weight from one limb to the other, while the contralateral foot is oscillating.

The balancing phase, corresponds to the toes taking off from the ground until the next heel contact, determining 40% of the cycle.

In the double stance phase, both feet are in contact with the ground, this means, the reference foot makes contact with the ground by the heel, slowing the body's acceleration forward, while the contralateral foot takes off from the ground. This phase represents 20% of the total cycle [20, 21, 22].

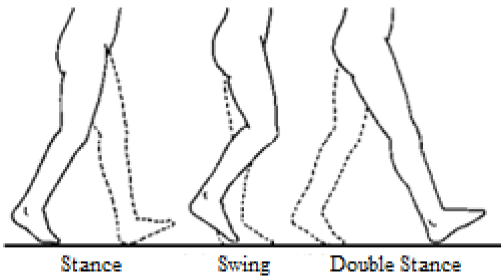

Figure 7. Phases in human walking.

On the other hand, the step length or stride length is the linear distance between two equal and successive events from the same limb (Fig. 8) and is composed of two phases: stance phase and oscillation phase. At a certain speed determined by the subject, each foot is out of phase compared to the other by 50% of the walking cycle, which allows that in a certain lapse, both feet are in contact with the ground, so that at a higher speed, the double stance phase is shorter and disappears when running [23,24].

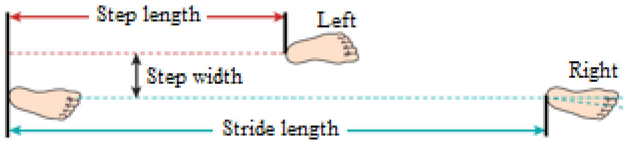

Figure 8. Human's stride length

Step length and step frequency allow walking speed's calculation, since a change in some of these variables or both, will automatically generate a change in speed, understanding step frequency as the number of steps per time unit, determining pace and walking speed. Its variation depends on lower limb length and body weight; increasing cadence or step frequency leads to an increase in step width and a decrease in step angle [25].

Bipedal gait is one of the most difficult processes to perform for infants during motor development stage. The ability to walk in children with CP increases their quality of life. Therefore, the required training for an appropriate walking skills development from an early age will give children a favorable future in terms of the level of custody they will need, and above all, the independence they may have during their adult life.

The trajectory followed by a specific point located on the foot can be found by simplifying the human leg's mechanism to a pair of articulated links (Figure 9). With the link length data (L_1 is the length from the hip to the knee, and L_2 is the length from the knee to the ankle), and the relative joint positions (q_1 is the angle between the femur and the hip, and q_2 is the angle between the calf and the femur), we obtain the two equations that indicate position P_x(1) and P_y(2) for the point located on the foot. The trajectory obtained mathematically, is verified with a simulation made in OpenSim.

$$P_x = L_2 \cos(q_1 + q_2) + L_1 \cos(q_1) \quad (1)$$

$$P_y = L_2 \sin(q_1 + q_2) + L_1 \sin(q_1) \quad (2)$$

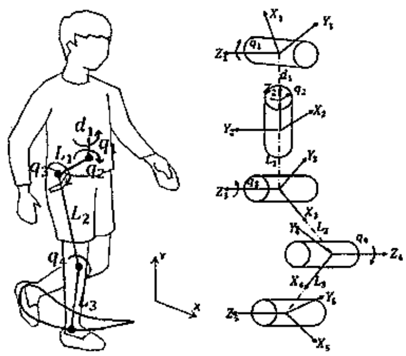

Figure 9. Simplified model of the lower limb and ankle trajectory during the walking process

STEWART PLATFORM

There is a huge diversity of mechanisms that allow human walking to be replicated, and even more so when the followed trajectory should only be replicated in position, and not in orientation. Experimentally, and due to the ease of its construction, a Stewart platform was proposed as one of the possible mechanisms to replicate human walking. Since this platform was already implemented, it was used as a rehabilitation mechanism to follow the ankle´s path on a scale prototyped foot. Initially, the 6 FD on the platform are not required, but for later trajectories in which foot orientation needs to be controlled, it can be done independently, and thus make the best use of the platform.

The experimental platform (Fig. 10) has a base and an upper plate made of aluminum. The platform is operated by Firgelli brand linear servomotors, with a 100mm stroke length, 16 mm/s maximum speed and 6N force. A scale model of a human leg was placed on the platform, to evaluate each joint's relative movements.

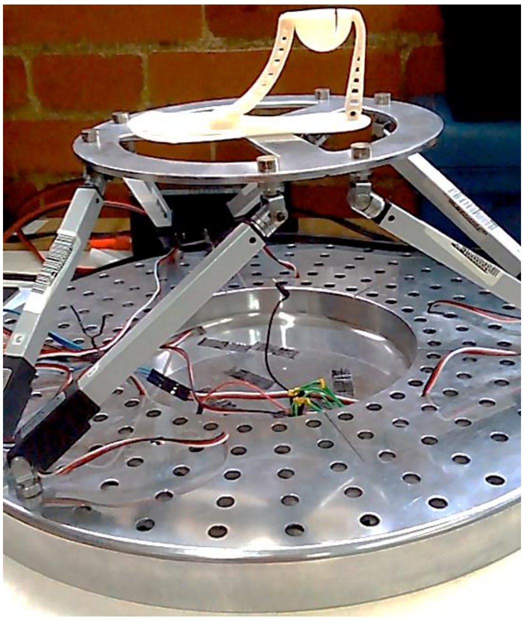

Figure 10. Experimental Stewart platform

Kinematic and Dynamic Analysis

The platform's positions, speeds and forces analysis was executed in the SolidWorks Motion module. Initially, an inverse kinematic analysis was carried out, in order to know the relative positions for each motor, as a function of time, when the platform was commanded to follow the walking trajectory.

This calculation can also be done by programming the Stewart platform's inverse kinematics equations [Eq (3) - (8)] in a computer routine, taking into account the input parameters, which depend on the model's geometry (Figure 11).

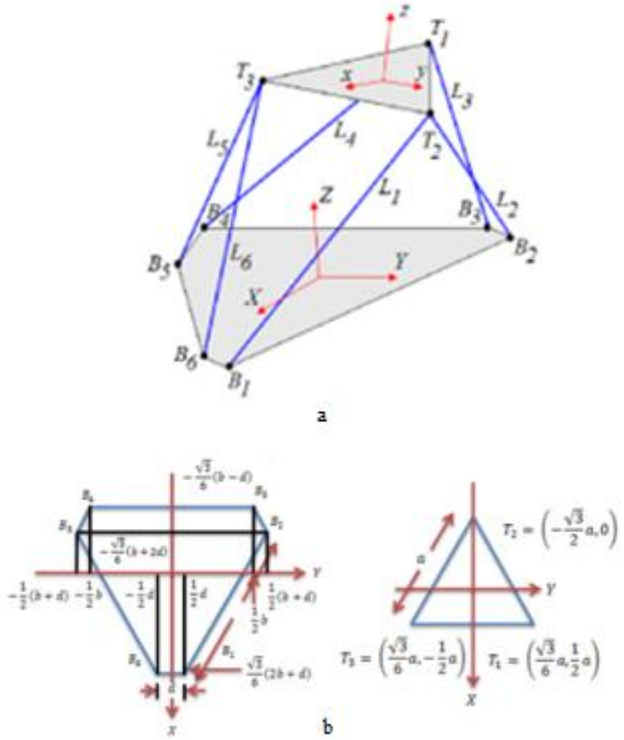

Figure 11. a) Coordinate systems arrangement. b) geometric parameters.

$$L_1 = \sqrt{\left(X_{T1} - \frac{d}{2\sqrt{3}} - \frac{b}{\sqrt{3}}\right)^2 + \left(Y_{T1} - \frac{d}{2}\right)^2 + Z_{T1}^2} \quad (3)$$

$$L_2 = \sqrt{\left(X_{T1} - \frac{d}{2\sqrt{3}} + \frac{b}{2\sqrt{3}}\right)^2 + \left(Y_{T1} - \frac{d}{2} - \frac{b}{2}\right)^2 + Z_{T1}^2} \quad (4)$$

$$L_3 = \sqrt{\left(X_{T2} + \frac{d}{\sqrt{3}} + \frac{b}{2\sqrt{3}}\right)^2 + \left(Y_{T2} - \frac{b}{2}\right)^2 + Z_{T2}^2} \quad (5)$$

$$L_4 = \sqrt{\left(X_{T2} + \frac{d}{\sqrt{3}} + \frac{b}{2\sqrt{3}}\right)^2 + \left(Y_{T2} + \frac{b}{2}\right)^2 + Z_{T2}^2} \quad (6)$$

$$L_5 = \sqrt{\left(X_{T3} - \frac{d}{2\sqrt{3}} \mp \frac{b}{2\sqrt{3}}\right)^2 + \left(Y_{T3} + \frac{d}{2} + \frac{b}{2}\right)^2 + Z_{T3}^2} \quad (7)$$

$$L_6 = \sqrt{\left(X_{T3} - \frac{d}{2\sqrt{3}} - \frac{b}{\sqrt{3}}\right)^2 + \left(Y_{T3} + \frac{d}{2}\right)^2 + Z_{T3}^2} \quad (8)$$

The trajectory followed by the platform's end point was graphed with a spline, trying to replicate the trajectory obtained mathematically. Since the trajectory's curve only indicates position, that is, it does not take time into account, the speed's function for the follower point is unkknown for this trajectory. For this reason, the point followed the trajectory with an almost linear speed profile, except for the start and end points, in which an acceleration and deceleration value was included (Figure 12).

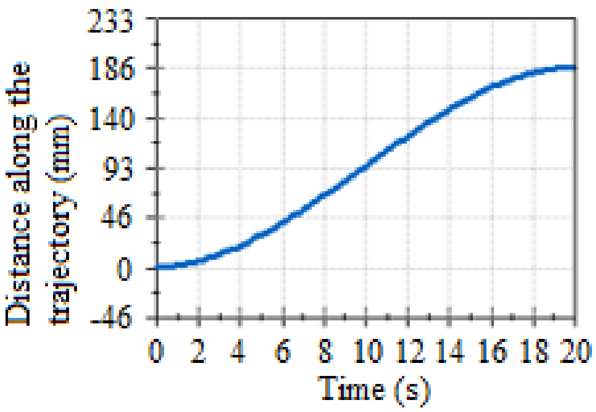

Figure 12. Fixed point's Displacement on the platform's mobile base on the ankle's path.

Since the the platform's movement is symmetrical, the actuators work in a similar way in pairs. The actuators were labeled (Figure 13), in order to identify the vectorial quantities of each one of them.

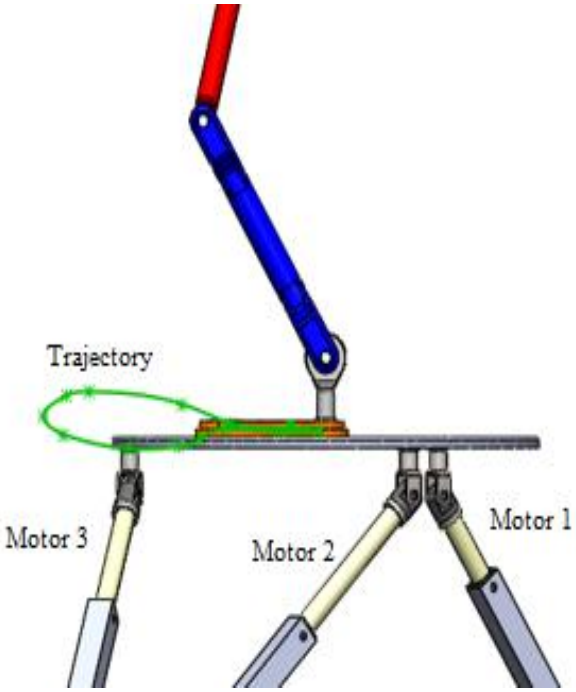

Figure 13. Motor´s Distribution and trajectory´s location.

A relative motion sensor was placed in each actuator, to see the position's behavior in terms of time for each one of them. Three curves were obtained (Figure 14), and the data was saved in a text file, to be used later in the inverse kinematics calculations.

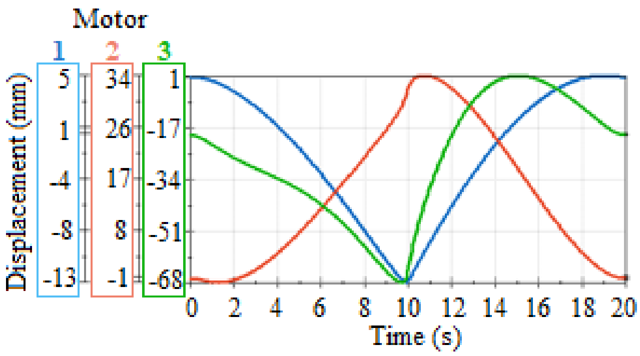

Figure 14. Motor's displacement curves.

With the curves indicating the position of each actuator as a function of time, a new motion analysis was carried out, in which the position profiles were loaded into the actuators, to verify that the platform's movement continued to replicate the ankle's trajectory. This analysis, which is nothing more than the system's direct kinematics, was applied not only to verify the tracking of the trajectory, but it was also used for the system's dynamic calculation.

The virtual platform used for movement analysis, had materials applied in the different pieces composing it. In the movement analysis, gravity was applied, in order to take body forces into account. It is also necessary to take inertial forces into account, due to the component's accelerations. The system also took the scale human leg system's mass into account, which is placed on the Stewart platform's upper plate (Figure 15).

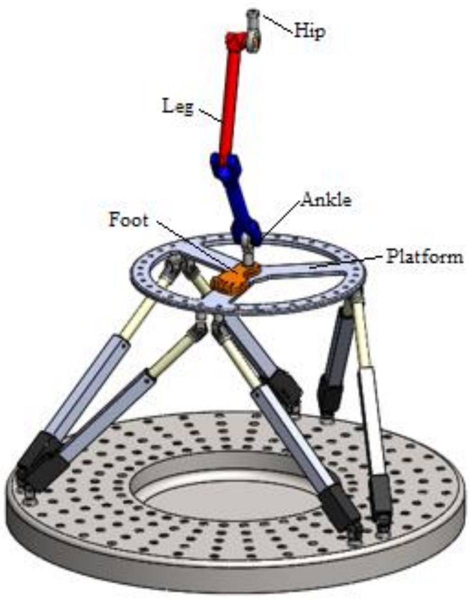

Figure 15. Lower limb anchored to the platform.

The measurements made in the direct kinematic analysis (Figure 16) showed the forces and speed profiles required for each engine, in order to comply with the movements involved in tracking the ankle's trajectory.

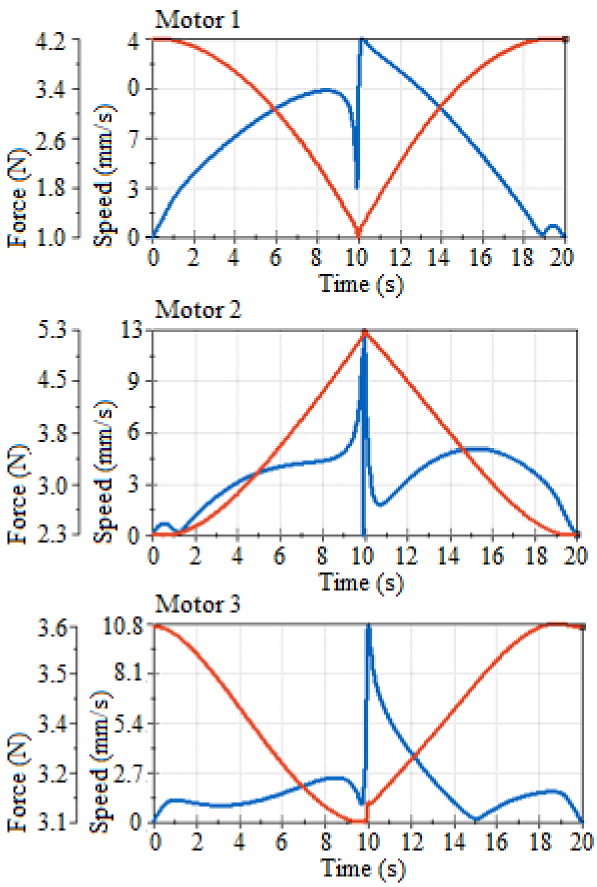

Figure 16. Speed and force curves for each of the engines.

RESULTS

Once the actuator's positions were calculated as a time function, and since they implied exact position data sending for each corresponding time, a function to connect those points was calculated, in order to be able to predict any point of the trajectory at any moment. With the MatLab CurveFitting package, the points obtained in SolidWorks, which had been stored in an Excel file, were interpolated. When interpolated, the fourier function gave better results (9), taking the interpolation for the 3 engines to 8 term polynomials.

$$f(x) = a_0 + \sum_{k=1}^{n} a_k \cos(kxw) + \sum_{k=1}^{n} b_k \sin(kxw) \qquad (9)$$

The curves obtained with the Fourier polynomials showed small errors during sudden direction changes. These errors are shown next to each curve's graph (Fig 17, 18 and 19), superimposed to the point cloud obtained from the data generated by SolidWorks. For engine 1, the curve deviated a maximum of 1.9mm, for engine 2, the deviation was 0.8, and for engine 3, it was 1mm.

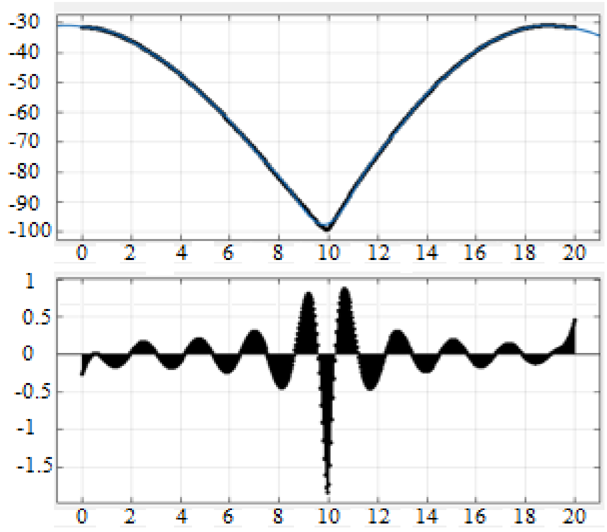

Figure 17. Curve for motor 1 and its deviation.

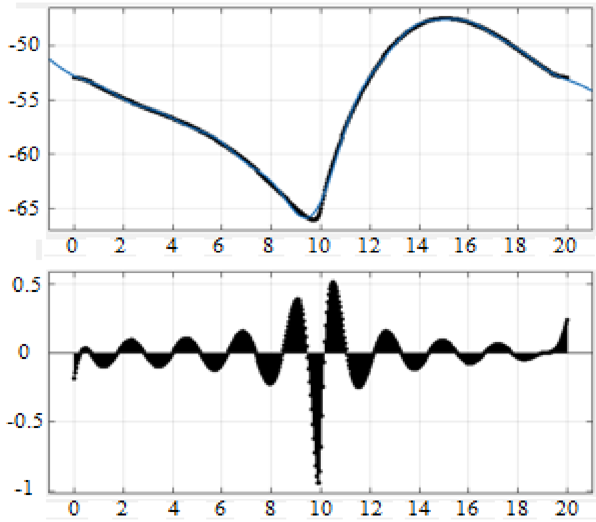

Figure 18. Curve for motor 2, and its deviation.

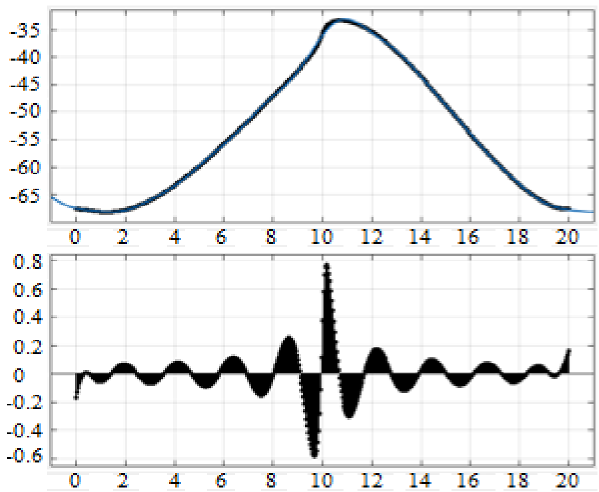

Figure 19. Curve for motor 3, and its deviation.

The curves obtained were used to move the Stewart platform (Fig. 20), supported by the Simmechanics tool. Within the platform's scheme, the Fourier polynomials were placed as input parameters in the prismatic joints that joined the stems of the actuators with its cylinders. These polynomials were entered as a function in ".m" extension file, which is called by the platform during the simulation (Figure 21).

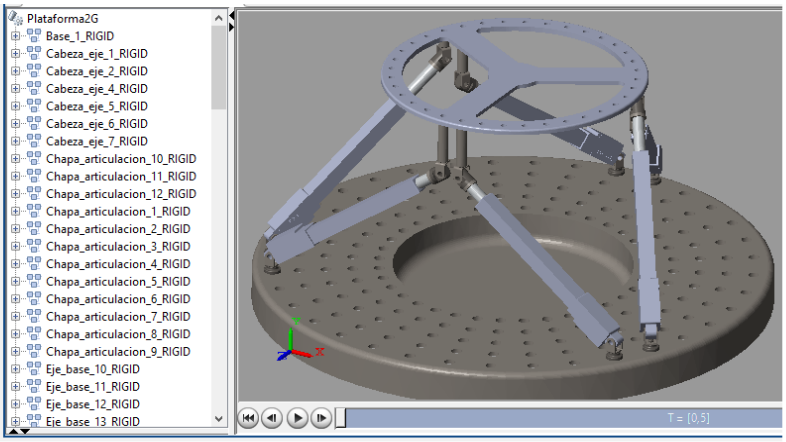

Figure 20. Stewart's platform during movement simulation in Simmechanics.

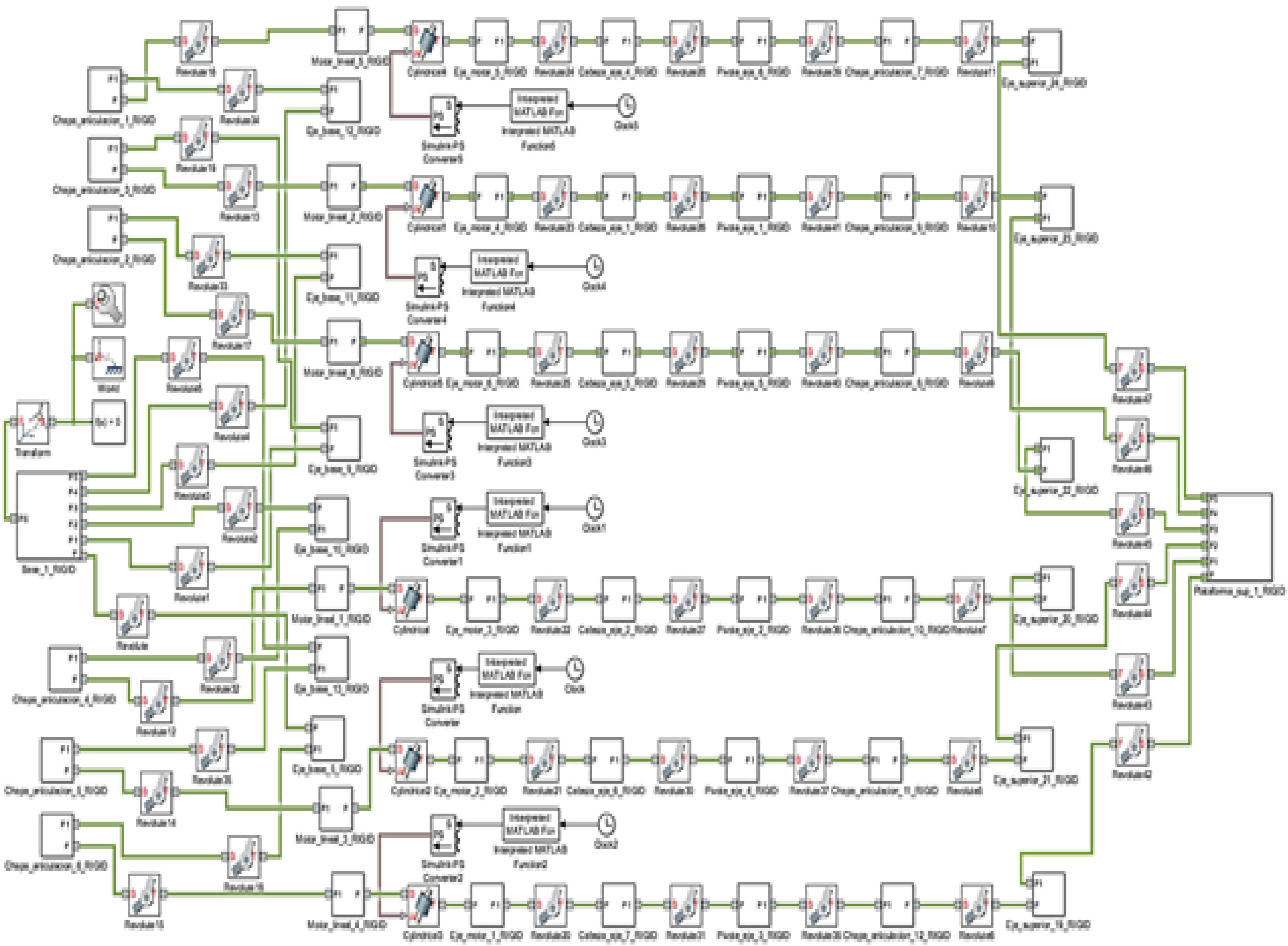

Figure 21. Block diagram for the Stewart platform with movement input functions.

For the tests on the experimental platform, all 6 linear actuators were connected to a controller, which had all 3 Fourier polynomials programmed, so that it could send the corresponding position to each motor, regardless of the time. Curves or polynomials were only valid to perform the movement in 20 seconds, because if its needed to be done in more or less time, different polynomials are required.

The platform's movement during the final tests was continuous and followed the desired trajectory (Fig. 22), performing the movement within the desired time. This is because, during the movement analysis, it was verified that force and speed parameters required for each actuator were within their deliverable range.

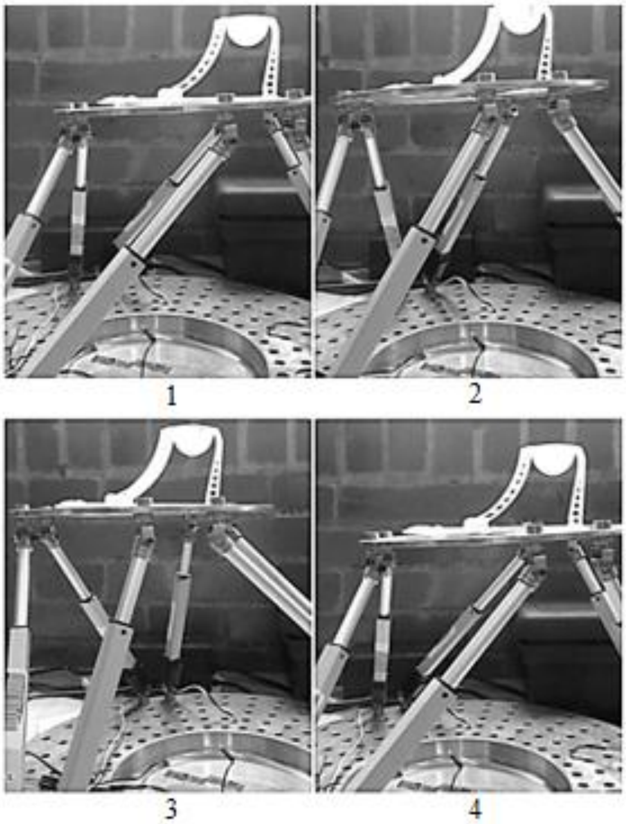

Figure 22. Stewart platform during the trajectory execution from initial point 1, until final point 4.

CONCLUSIONS

Motor's Speed and force data, obtained by movement analysis, allowed to verify that the actuators mounted on the platform had the required mechanical capacities to reach the requested positions. The simulation showed smooth and perfectly synchronized movements, however, it is necessary to generate the curve as a time function, since using a data chart like the one SolidWorks delivers, would force the controller to send an exact data in a certain time.

Since an already built experimental platform was used, it was impossible to have control over some aspects. However, for real-scale tests, it is necessary to implement position sensors that close the control loop, since the tests shown in this document were in an open loop, implying a high risk for a real situation.

The experimental platform followed the trajectory, partly because of the low load it had to move and the long time it took to perform the movement, but it is clear that, for faster movements, the actuator's mechanical requirements can increase exponentially, and the controller's Response time needs to be shorter.

ACKNOWLEDGEMENT

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