DESIGN AND MODEL A NOVEL ANKLE FOOT ORTHOSIS

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ABSTRACT
A lower limb orthosis is a type of external supporting devices that are used to support the patients who have problems due to trauma, incomplete spinal cord injury, stroke, etc. in the process of treatment and recovery. In the three articulations of lower extremity, ankle joint bears more weight than the other joints. Almost the current active, passive ankle orthosis systems focus on controlling dorsiflexion, plantar flexion around ankle joint or both of them. In reality, however, the locomotion of human is continuous movement of Center of Pressure (CoP) points underneath the foot. This paper suggests a novel ankle orthosis system based on CoP. This new model can control dorsiflexion, plantarflextion to prevent foot from foot slap and foot drop effect for patients who have trouble with their ankle. This paper will demonstrate the design and the model of the novel orthosis system.

Keywords: Ankle Foot Orthosis (AFO), Center of Pressure, Design, foot slap, foot drop.

1. INTRODUCTION

Locomotion is a fundamental movement of human and has reciprocal influences with the other organs in the human body. For those who have gait impairment have to face significant health consequences, including loss of bone mineral content, bedridden situation, increased incidence of urinary tract infection, muscle spasticity, impaired lymphatic and vascular circulation, impaired digestive operation, and reduced respiratory and cardiovascular capacities (Phillips and et al. 1987). The physical therapy for these gait impairments varies depending on the cause of gait disorders. Prevalent causes leading gait impairments are neurological disorder and injuries, such as trauma, spinal cord injuries, multiple sclerosis, and Parkinson's disease.

Robotic therapy is more and more used in gait rehabilitation after stroke because of having a number of advantages. These advantages include reducing the physical burden for the therapist, ability to provide a large volume of movement in a safe environment, enhancing capacities of monitor performance of the lower limb during rehabilitation.

Currently, the AFO can be divided into two main groups as either passive of active AFO. The first type, passive AFO (Becker Orhopedic 2009, Kitaoka and et al. 2006, Novachec and et al. 2007), is used when human subject transfer force to move the ankle joint. On the contrary, for the other type of AFO, the actuators generate torque to control the system. The passive AFO, which is further classified by material, has some virtues such as: light weight, economy, and ability to provide assistance stiffness from some few Newton meters up to 20 N.m of resistive torque. However, the main essential point needs to be improved for this type of system is control capacity. The control of the passive AFO depends on the activation of spring, valves, or switches in open-loop as the user walk (Shorter and et al. 2013). Besides, the passive AFO has another limitation that is impossible to generate propulsion force during terminal stance phase. In order to improve the robustness capacity of controller, overcome the drawback points of the passive AFO, the active AFO(s) was developed using different external force sources to control ankle articulation. These energy sources include Magneto Rheological (MR) damper (Naito and et al. 2009), Series Elastic Actuator (SEA) (Svensson and Holmberg, 2008), McKibben pneumatic actuator (Ferris, Czerniecki and Hannaford 2005) and etc. The active AFO(s), which improve the weak points of the passive AFO, however, still contain some limitations. For the AFO system that uses the MR damper, they are capable of controlling foot motion, but they are impossible to generate the propulsive force in the push-off phase. The AFO developed by a group of MIT using SEA (Hwang, Kim and Kim 2007) can assist patients with both plantarflexion and dosirflexion. However, this system did not fully support when push-off phase. It only reduced the impedance of the AFO to allow full plantarflexion movement.

In this paper, we present a novel 3D model of active AFO that can support patients with both plantarflexion and dosirflexion movements, prevent the weak ankle foot from both foot drop and foot slap. Moreover, the new system is also capable of supporting the foot to generate propulsive force at terminal stance phase.

The structure of the remaining paper is described as follows. The paper starts with the introduction of a normal walking cycle. After that, the mechanical design as well as 3D model of the new AFO is shown. The controller selection and controllable feasibility of the

AFO are presented in the followed section. Finally, the conclusion and future works are demonstrated.

2. A NORMAL GAIT CYCLE

A normal level-walking gait cycle is defined as starting with the heel strike of one foot and terminating at the next heel strike of the same foot (Inman, Ralston, and Tood 1981, Perry 1992). This cycle is subdivided into two sub cycles as the stance phase (about 60% of a gait cycle) and swing phase (about 40% of a gait cycle) as in Figure 1. Whereas the swing phase (SW) is a portion of the gait cycle when the foot is off the ground, the stance phase (ST) is from heel-strike when the heel contacts with the ground to toe-off when the same foot leave the ground. According to the ref(s) (Palmer 2002, Gate 2004) the ST phase can be divided into three sub-subphases: Controlled PlantarFlexion (CP), Controlled Dosirflexion (CD) and Powered Plantarflexion (PP) (Berniker and Herr 2008).

- The CP begins at heel-strike and ends at the foot-flat. In this process the heel and forefoot initially contact with the ground.
- The CD begins at the foot - flat and continues until the ankle reaches a state of maximum dorsiflexion.
- The PP begins after CD and finishes at the instant of toe-off.

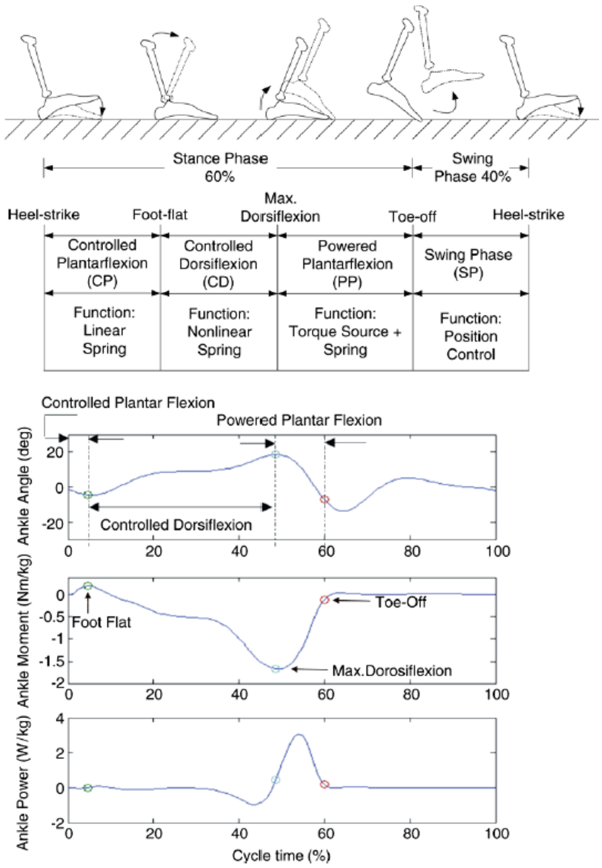


Figure 1: Normal Ankle biomechanics when level walking (Berniker and Herr 2008).

3. MECHANICAL STRUCTURE OF THE NOVEL AFO

The idea to develop the new active AFO is generated from the movement of the CoP during locomotion. The CoP (gait line) is the average vector of all force that acts on the bottom of the normal foot as it goes through the stance phase. This line goes from the lateral heel side of medial forefoot and leave at the toe of the foot. This movement goes very fast from the rearfoot to forefoot (Grundy, Tosh and McLeish RD 1975)

We developed a novel active AFO to control two important points of CoP line: the lateral rearfoot and medial forefoot point by using one DC Servomotor. The Table 1 expresses the control role of the each point when implementing to control the AFO. In the gait cycle mentioned in the Table1 the SW phase is more specifically divided into Initial SW (ISW), Medium SW (MSW) and Terminal SW (TSW).

		Stance phase			Swing phase		
		CP 5%	CD 40 %	PP 15 %	ISW 10 %	MSW 15 %	TSW 15 %
Controlled part	Rearfoot point	○	○	×	×	×	×
	Forefoot point	×	×	○	○	○	○

Table 1: Controlling role of rearfoot and forefoot point.

In order to separate one torque source from a DC Servomotor into two different sources to control two points, a three bevel gear mechanism that are perpendicularly assembled is used as in Figure 3.a. These three bevel gears include a main bevel gear connected to the motor, a rearfoot bevel gear that has a clearance fit on the rearfoot shaft to control rearfoot point, and a forefoot bevel gear which also clearance fit on the forefoot shaft to control forefoot point. Moreover, it is necessary to design some mechanism to change the torque flow from the rearfoot shaft to the forefoot shaft and vice versa. These mechanisms are summarized as in Figure 2:

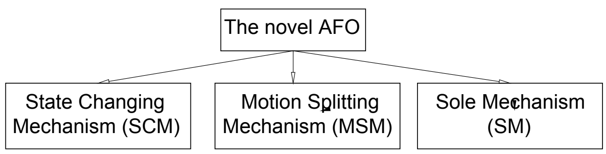
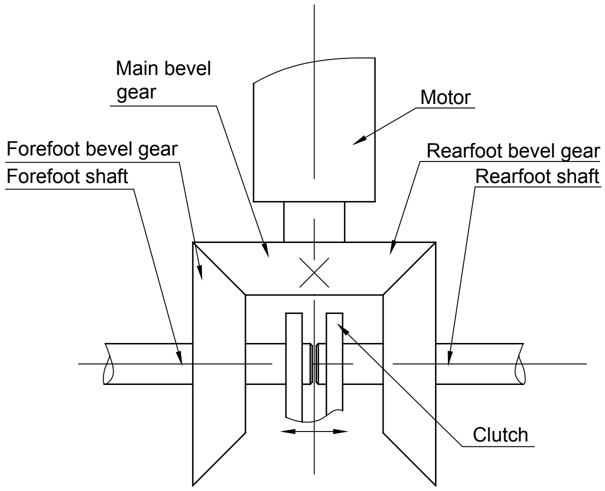


Figure 2: Mechanisms of the AFO.

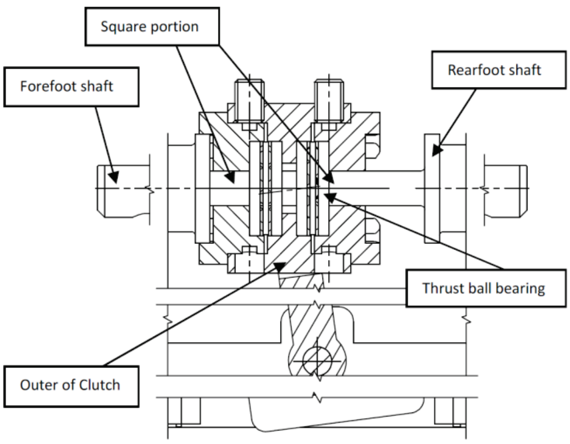
3.1. Motion Splitting Mechanism

The mission of the MSM is to change the torque flow from the DC Servomotor into two different shafts: the rearfoot shaft and forefoot shaft.

After designing and choosing, the final structure of the MSM is shown in the Figure 3. The torque flow is transferred as follows: Motor -> Main bevel gear -> Rearfoot (forefoot) bevel gear -> Rearfoot (Forefoot) shaft -> Rearfoot (Forefoot) point on the sole.



a) Bevel gears structures.



b) Motion Splitting Mechanism.

Figure 3: The bevel gears structure and Motion Splitting Mechanism.

The system uses clutches to transfer the energy from the rearfoot/forefoot bevel gear to the rearfoot/forefoot clutch shaft, respectively. Because the shaft's velocity is not high, then in order to transmit the energy from rearfoot/forefoot clutch to rearfoot/forefoot shaft the system uses a square section shaft portion as well as a square section hole on the clutches as in Figure 3.b.

3.2. Changing State Mechanism

In the operation of the MSM, in order to change from controlling the rearfoot shaft to controlling the forefoot shaft and vice versa, it is necessary to have an external mechanism to alter the clutches states. The states' clutches are ON/OFF forefoot/rearfoot clutch on the clutch portion of the bevel gear.

The current model used two different Electromagnets to control the ON/OFF states for the two clutches. From the 3D model expressed in Figure 4, in order to close the forefoot clutch (on the left side) the system's controller has to turn the rearfoot electromagnet off and turn the forefoot electromagnet on to generate the pull force. This force makes the level rod rotate around O point and pull the clutch move to the left side.

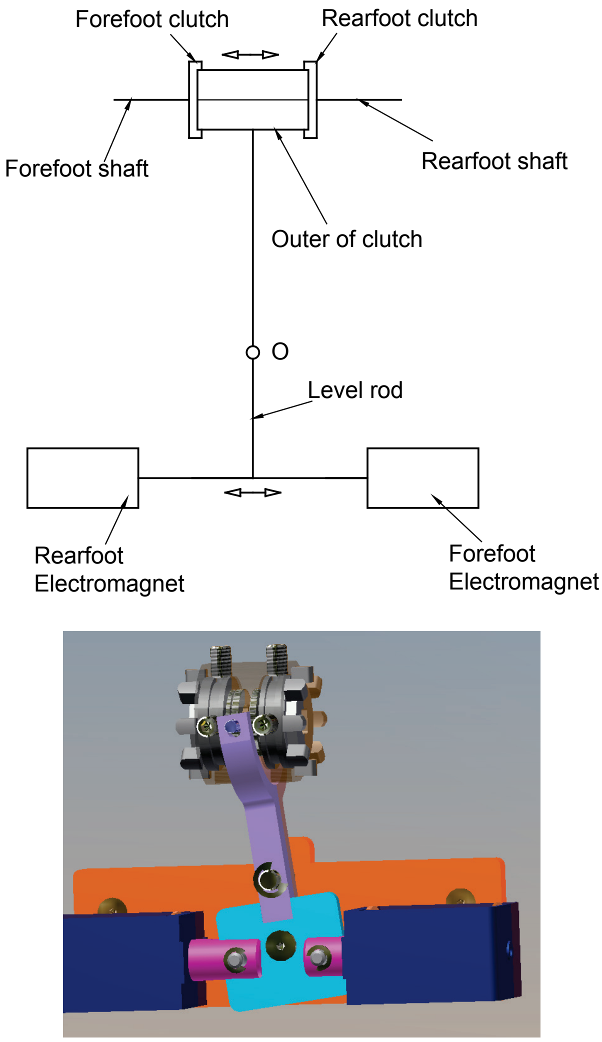


Figure 4: The changing State Mechanism.

3.3. The sole mechanism of the AFO

The sole of the AFO has roles of transferring the controlled forces from the motor to the ground to rotate the ankle joint at stance phase and to rotate ankle articulation only at swing phase. In the current design, the model used the hinge structure for the AFO's sole mechanism. Each of rearfoot and forefoot sole portions has two plates which are upper and lower plate as in Figure 5.

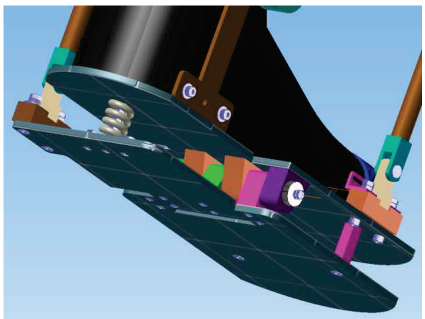


Figure 5: AFO's sole structure.

At the initial contact time, the impact force increases very fast and can be harmful for the pathologic ankle foot as well as for the mechanical parts and motor system. To reduce this effect, a compression spring is assembled between the upper and the lower plate of the rearfoot.

In summary, after design the new active AFO which has some proposed mechanisms, the final 3D model is presented in the Figure 6. It is also noted that in this model we also attached a potentiometer to measure the ankle angle. Besides, the system has two hinge mechanisms to hang the system on the foot's shank. By using the hinge structure the system can increase the force transferring area between the AFO and human shank even though the dimension of the shank is changed with different patients.

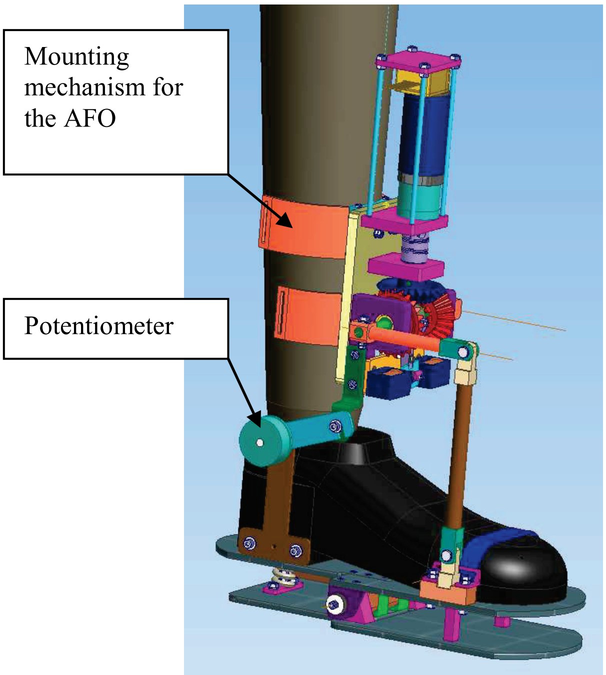


Figure 6: 3D model of the new AFO which is also attached the mounting mechanism and potentiometer.

4. FEASIBILITY ANALYSIS

In order to analysis the possibility of controlling of the system, the system was assumed to be controlled by using a position controller.

According to the above discussion, the system uses the state controller to detect the phase of gait and control them. These states include CP, CD, PP and Swing state as in Table 1. Following is the feasibility analysis of controlling for each of the state:

4.1. Controlled Plantarflexion and Controlled Dorsiflexion

According to Table 1 and Figure 1 these sub-subphases start from the initial contact and ends at the maximum dorsiflexion angle. The system used the rearfoot point to control this stage. As in Figure 7, during this stage the rearfoot part of the AFO always contacts with the ground. Moreover, in the sagittal plane, there are rotational joints of A, B, C, and D.

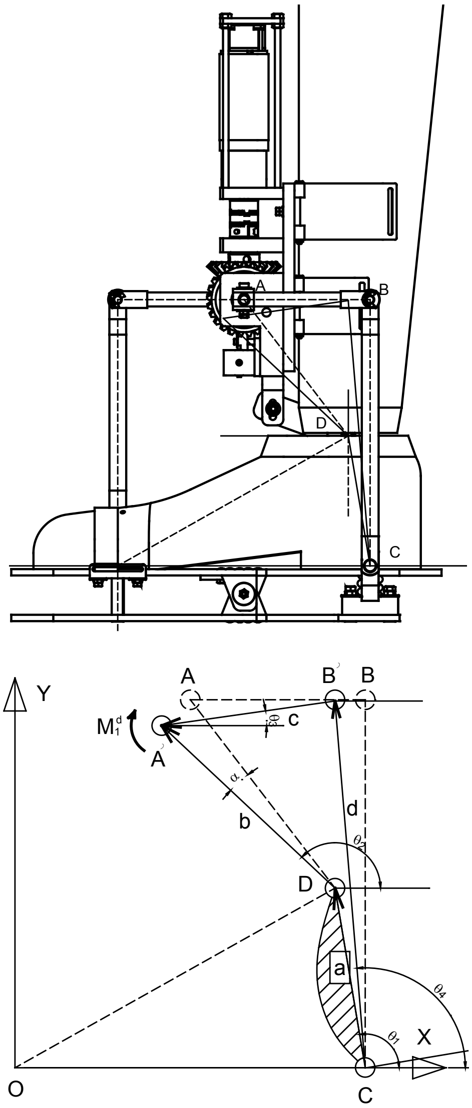


Figure 7: Positon controlling model at the 1st stage.

Because during the 1st stage, the rearfoot plate always contacts with the ground, the CD link is fixed to the floor. As a result, the ABCD linkage is equal to four-bar mechanism. It is necessary to find out the angle relation between the rearfoot shaft angle θ_3 and the angle at ankle θ_2.

Using complex number to solve this problem:

$$\overrightarrow{CD} + \overrightarrow{DA} - \overrightarrow{CB} - \overrightarrow{BA} = 0 \qquad (1)$$

$$a.e^{j\theta_1} + b.e^{j\theta_2} - c.e^{j(180+\theta_3)} - d.e^{j\theta_4} = 0$$

After deploying the above mathematic model, we have:

$$K_1.\cos\beta_2 + K_2.\cos\beta_3 + K_3 = \cos(\beta_2 - \beta_3)$$

where

$$K_1 = -\frac{a}{c}; K_2 = -\frac{a}{b} \text{ and } K_3 = \frac{-a^2 - b^2 - c^2 + d^2}{2.b.c}$$

$$\beta_2 = \theta_1 - \theta_2; \beta_3 = \theta_1 - \theta_3;$$

Then

$$K_1.\cos\beta_2 + K_2.\cos\beta_3 + K_3 = \cos\beta_2.\cos\beta_3 + \sin\beta_2.\sin\beta_3 \quad (2)$$

Besides,

$$\sin\beta_3 = \frac{2.\tan(\frac{1}{2}\beta_3)}{[1+\tan^2(\frac{1}{2}\beta_3)]} \; and \; \cos\beta_3 = \frac{1-\tan^2(\frac{1}{2}\beta_3)}{1+\tan^2(\frac{1}{2}\beta_3)}$$

Replacing to the above functions (2):

$$A.\tan^2(\frac{1}{2}\beta_3) + B.\tan(\frac{1}{2}\beta_3) + C = 0 \quad (3)$$

The function (3) has two roots as:

$$\theta_3 = \theta_1 - \beta_3 = \theta_1 - 2.\tan^{-1}[\frac{-B \pm \sqrt{B^2 - 4.A.C}}{2.A}] \quad (4)$$

The roots (4) of angle θ_3 can be calculated by the geometric dimensions of the system as: a, b, c, d and ankle angle. There is a notice that the absolute value of the angle θ_3 is less than 90 degree, then in the two roots (4) there is only one value that is sufficient the condition. As a result, it is possible to control the CP and CD of the first stage.

4.2. Powered Plantarflexion and Swing phase

The system uses the forefoot point to control PP and SW sub-phase. During the PP duration, the forefoot plate always contacts with the ground and the rearfoot plate does not. However, the angle between the rearfoot and the ground at instant time is able to be calculated based on the values of hip, knee and ankle value. According to Figure 8

$$\gamma = h + k + a \quad (5)$$

Where: h, k and a are the angle value of hip, knee and ankle joint, respectively.

As a result, at instant time, the CO linkage can be recognized as a fixed linkage. Then, the ABCO is equal to four-bar structure and it is similar as the calculation aforementioned. The relationship between the rearfoot shaft and ankle value as follow:

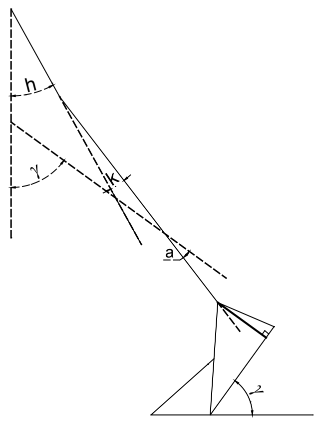


Figure 8: Value between the rearfoot plate and ground.

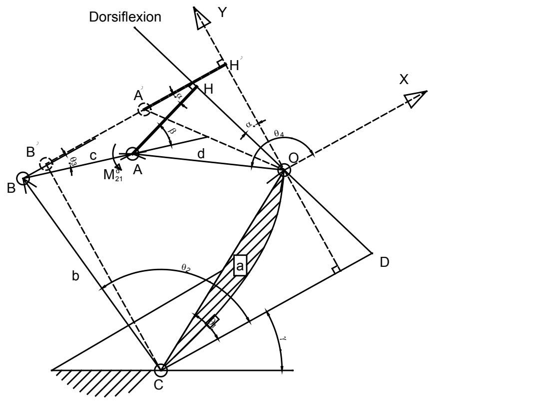


Figure 9: Position control model for PP

$$\beta = \alpha + \theta_3 = \alpha - \theta_1 + 2.\tan^{-1}[\frac{-B \pm \sqrt{B^2 - 4.A.C}}{2.A}] \quad (5)$$

Where:

$$A = (1+K_2).\cos\beta_4 - K_1 + K_3$$
$$B = -2.\sin\beta_4$$
$$C = (K_2 - 1)\cos\beta_4 + K_1 + K_3$$
$$K_1 = -\frac{a}{d}; K_2 = \frac{a}{c} \; and \; K_3 = \frac{a^2 + c^2 + d^2 - b^2}{2.c.d}$$

For the Swing phase, if the coordination is attached to the shank the AO of the ABCO mechanism is fixed. The relationship between controlled angle of the rearfoot shaft and the ankle angle values.The relationship between the rearfoot shaft ankle and the ankle angle is described

$$\beta = \alpha + \theta_3 = \alpha + \theta_1 - 2.\tan^{-1}[\frac{-B \pm \sqrt{B^2 - 4.A.C}}{2.A}] \quad (6)$$

Where

$$A = (1+K_2).\cos\beta_4 + K_3 - K_1$$
$$B = -2.\sin\beta_4$$
$$C = K_1 + K_3 - (1-K_2)\cos\beta_4$$
$$K_1 = -\frac{a}{d}; K_2 = \frac{a}{c} \; and \; K_3 = \frac{a^2 + c^2 + d^2 - b^2}{2.c.d}$$

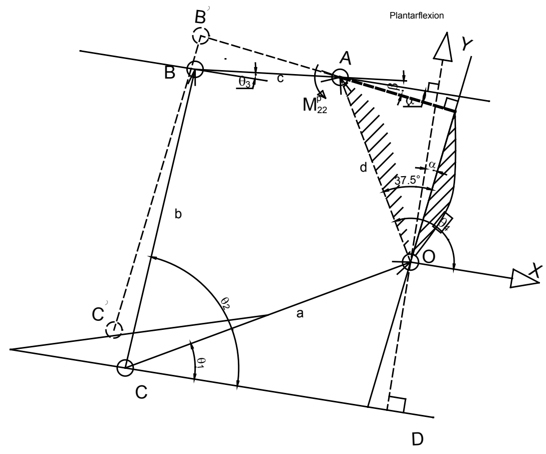


Figure 10: Position control model for the SW.

5. CONCLUSION

The paper presented the mechanical design, 3D model of a novel ankle foot orthosis. This model could help patient with ankle impairment prevent from foot drop, foot slap and improve capacity of dorsiflexion and plantarflexion. Moreover, the feasibility of the new model was also analyzed and discussed to ensure the controllable ability of the system.

In the future, the first prototype of the novel AFO will be produced and make some experiments to evaluate the system. Furthermore, in this prototype is also attached some force sensors to sense the impact force underneath of the sole to support the controller of the system.

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