Visión Electrónica Vol. 17 No. 1 (2023) • january-june • P.P. 152-160 • ISSN 1909-9746 • ISSN-E 2248-4728 • Bogotá (Colombia)-

UNIVERSIDAD DISTRITAL
FRANCISCO JOSÉ DE CALDAS

Visión Electrónica

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VISIÓN ELECTRÓNICA

A CASE-STUDY VISION

Serial Elastic Actuator Model for Ankle-Foot Prosthesis Matlab Simulation

Simulación del modelo de actuador serial elástico para prótesis Tobillo-Pie en Matlab

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INFORMACIÓN DEL ARTÍCULO

Historia del artículo:
Enviado: 04/12/2022
Recibido: 20/12/2022
Aceptado: 16/02/2023

Keywords:
Serial Elastic Actuator (SEA)
Ankle Foot Prosthesis
Gait Cycle

Palabras clave:
Actuador elástico en serie (SEA)
Prótesis de tobillo y pie
Ciclo de marcha

ABSTRACT

The ankle - foot set plays a very important role for human displacement, such as walking or running, giving vertical support and propulsion to the human walking progression by using the muscle extension and contraction. Many designs have been developed to replicate the function of normal gait, lost by injuries or diseases affecting the limb below the knee [1]. Motor rehabilitation has become a field of growing interest, due to the large number of cases of people with injuries or mutilation in its members or in other cases by cerebrovascular accidents and spinal cord damage that cause paralysis or any kind of disability. [2], [3]. This paper shows the process to get the model of SEA mechanism in Matlab, linking VR-World of Simulink from 3D Solidworks Model to test the model and finally checking the characteristic curves of normal gait to 1.5 m/s with this SEA prosthesis.

RESUMEN

El conjunto tobillo-pie desempeña un papel muy importante para el movimiento humano, como caminar o correr, ya que proporciona apoyo vertical y propulsión de la progresión de la marcha humana mediante la extensión y contracción muscular. Ante lesiones o enfermedades que afectan la extremidad debajo de la rodilla[1], se han desarrollado muchos diseños para replicar la función de la marcha normal. En Colombia la rehabilitación motora se ha convertido en un campo de amplio interés, debido a la alta incidencia de personas con lesiones, mutilaciones en sus miembros, discapacidades por accidentes cerebrovasculares, daño medular y otros traumas. [2], [3]. Este artículo describe el proceso para obtener un modelo del mecanismo SEA en Matlab, vinculando el VR-World de Simulink con un modelo 3D de la prótesis en Solidworks. El objetivo es validar el diseño y verificar las curvas características de la marcha normal a 1,5 m/s con la prótesis SEA. Este enfoque integrado permite un análisis detallado y una evaluación precisa del rendimiento de la prótesis en condiciones de marcha simulada.

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Citar este artículo como: F. N. Giraldo Ramos, "Serial Elastic Actuator Model for Ankle-Foot Prosthesis Matlab Simulation", Visión electrónica, vol. 17, no. 1, pp. 152-160, january-june 2023, https://doi.org/10.14483/22484728.20389

1. Introduction

There are 10 million amputees in the world, that is 1.5 per thousand people according to the University of Stanford [4] of which according to the DANE census of 2005 in Colombia there are 46,200 amputees of lower limb, amputation being predominant transtibial by antipersonnel mines. According to PAICMA, in Colombia between January 1990 and October 2016, 11,460 victims of antipersonnel mines were reported, this fact classifies it as the third country in the world, with the most victims of antipersonnel mines. These cases have incidence in the research's number in the design of prostheses [5].

Human body requires a flexible foot to facilitate movement over different types of environments and in situations such as walking or running. Traditional prostheses do not solve the problem of walking, since they can generate problems of hip and back [3]. Many of the commercial prostheses for ankle - foot, are completely passive during the support phase in the march, therefore their mechanical properties remain fixed during the walk at a higher speed or in variable terrain. For this reason, this conventional kind of prosthesis have many problems in the locomotion phase, such as non-symmetrical gait patterns, slower speeds and higher metabolic rates, compared with normal people performance. Transtibial amputees spend between 20 % to 30 % more metabolic energy to walk at the same speed as healthy people, for this reason prefer a slower walking speed to travel for covering the same distance [4].

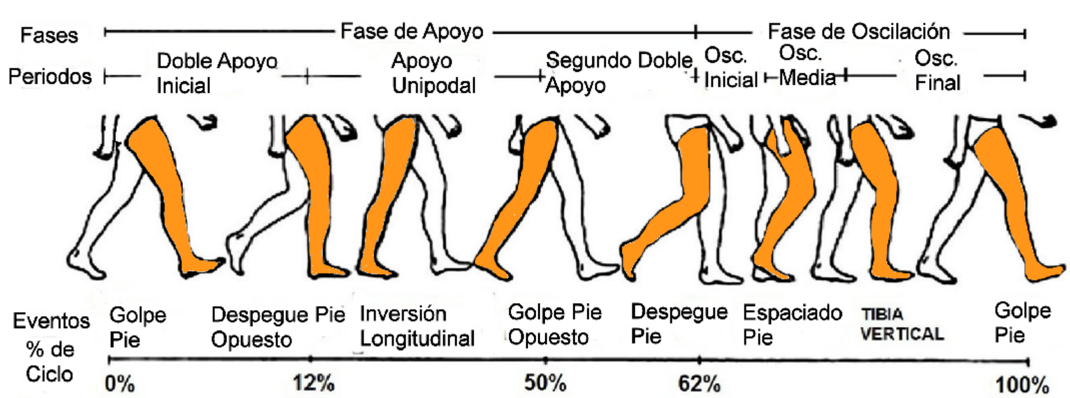
To develop an efficient ankle foot prosthesis, ankle-foot biomechanics must be studied and modeled [5]. The ankle works like a hinge that supports until two times the body weight when walking and eight times more when running.

1.1. Gait Cycle

Gait cycle, can be understood as a series of alternating movements of the trunk and extremities, which determine a walking rhythmic movement forward of the center of mass, where the lower extremities alternate following a succession of double support and unipodal support in order that the body never stops touching the ground. Walking cycle, is usually defined as the one that starts with the heel strike of one foot and ends in the next heel strike of the same foot. The walking cycle has two phases: support phase, which is taken during the moment in which the foot that has reference is in contact with the ground (60% of the cycle) and the oscillation phase that is taken during the moment in which the foot that is taken as reference is suspended in the air. (40% of the cycle) [3].

The support phase (SP) starts when the heel touches the ground and ends at the tip of the foot,

Figure 1. Human Gait Cycle [5].



when the same foot starts to rise. The support phase can be divided into three subphases: Controlled Plantar flexion (CP), Controlled Dorsiflexion (CD) and Powered Plantar flexion (PP). CP starts at the heel and ends at the foot level; the behavior of the ankle is like the linear response of a spring. CD starts after CP, at the level of the foot and continues up to the ankle going to a maximum dorsiflexion state, in this case the ankle behavior is similar to a non-linear spring that stores energy [5]. PP starts after CD and ends at the moment of lifting the foot, the ankle-foot set, can be understood like as a generator pair in parallel with the spring CD. Swing phase (SW) starts with the lifting of the foot and ends in the heel strike. In the SW phase, ankle-foot set, can be modeled as a position generator to restore the foot to a desired equilibrium position before the next heel strike [5].

During walking, the ankle provides three functions mainly:

- It behaves like a spring of variable stiffness between CP to CD.
- It provides additional energy for the lifting during PP.
- Adjust the position to control the orientation of the foot during SW.

Table 1. Natural ranges of foot movement [2]

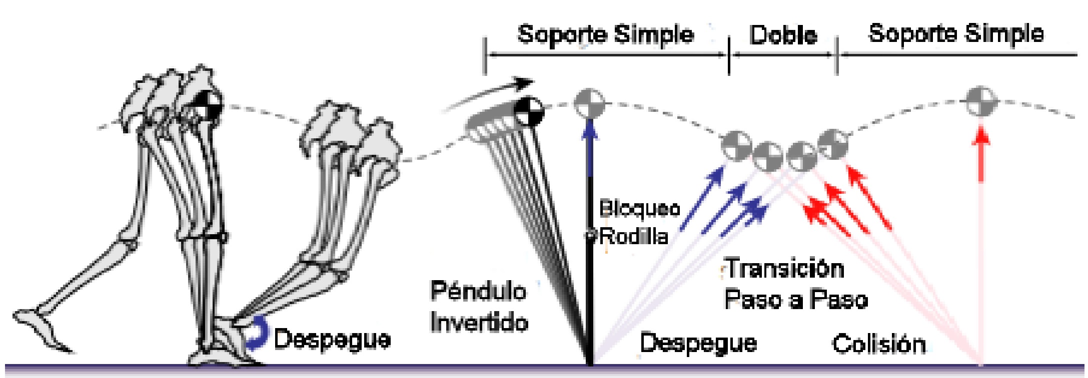
Movement	Range Max. (°)}	Reference frame
Dorsiflexion / Plantar flexion	20/50	Sagittal
Inversion/Eversion	35/25	Transverse
Abduction/Adduction	20/25	Frontal

1.2. Ankle - Foot Biomechanics

Ankle-foot set, can be modeled as a Series Elastic Actuator (SEA), which is a mechanism that has a linear spring in series between a motor and the system output. The spring tension is measured to obtain an accurate estimate of the force. The model used by the author for a SEA, corresponds to a system of second order mass, spring and shock absorber, with force conduction on the mass and a position input.

The best way to recover mobility in a natural way, is to restore the energy of the foot, as in the real ankle, essentially by means of a spring in the foot that produces a rebound after applying the force [8]. Oscillation phase: The desired behavior is to relocate the foot to a predefined equilibrium position. Support phase: the quasi-static rigidity is imitated, corresponds to the slope of the measured curve of the ankle torque angle during the posture [9]. The system must be able to control the position of the joint, while it happens the oscillation phase. The prosthesis must provide sufficient shock

Figure 2. Pendular Model [7].

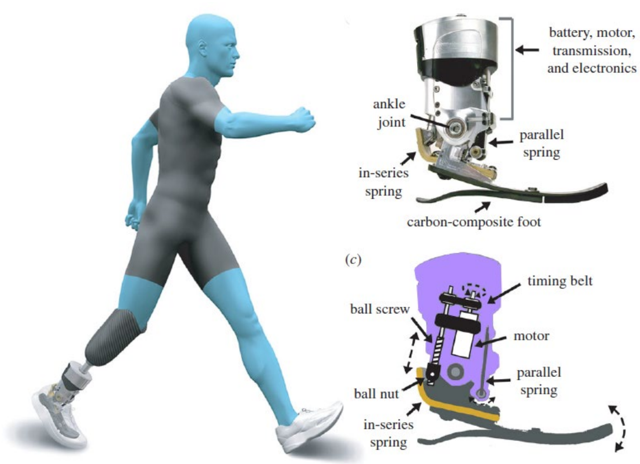


tolerance to avoid any damage to the mechanics during the heel strike [10].

Main Mechanical Elements Elastic Actuator Series:

- Motor D.C. (High Power).
- Transmission (Ball Screw)
- Series Spring
- Carbon composite blade.

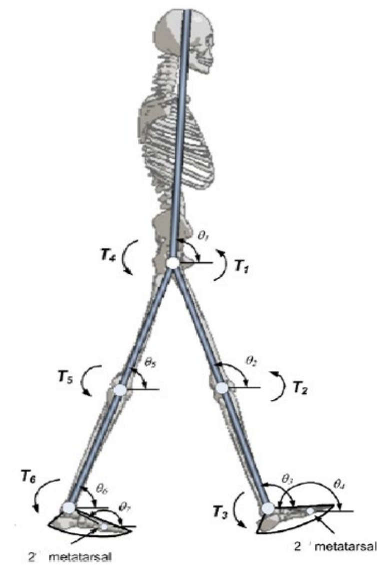
Figure 3. SEA Components [2].



1.3. Mathematical Model

The human body can be modeled like a seven-segment planar system (sagittal plane). Interaction between ankle-foot and the ground is modeled like a rigid reaction, where the point of contact is determined by the shape of the plantar surface and the orientation of the foot. The angles in the segments θ_1, θ_2, θ_3, θ_4, θ_5, θ_6 they are used to describe the orientation of each segment of the body, respect to the overall frame of reference. And the torques τ_1, τ_2, τ_3 they are the moments of muscular strength of each articulation in a multi - segment model [11].

Figure 3. Sagittal Plane Angles and Torques [8].



The actuator can apply a loading force on the spring Fs given a desired force. The output impedance is the amount of force at the output of the actuator given a change in load position. Ideally, the actuator as a pure force generator would have a zero-output impedance. Therefore, the dynamics of the actuator would be completely decoupled from that of the movement of the load. However, since the actuator is a real physical system with losses, it has a certain output impedance.

$$M\ddot{x} + b\dot{x} + k_s x = F_{in}$$

Where, $M = I_m R^2$, $F_{in} = \tau_m R$, $b = b_m R$

Figure 4. SEA Model [1].

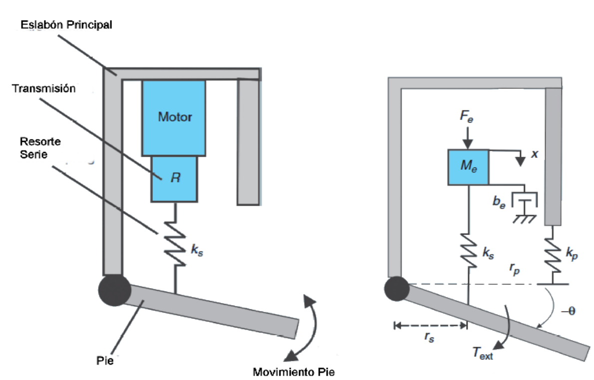
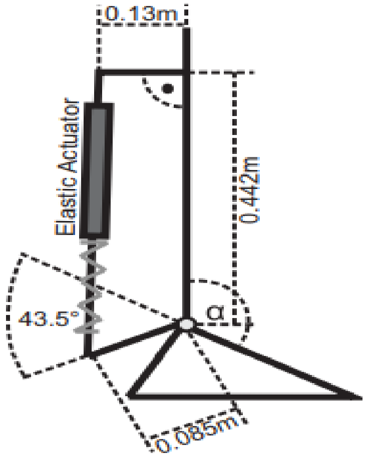


Figure 5. SEA Mechanism Prostheses [2].

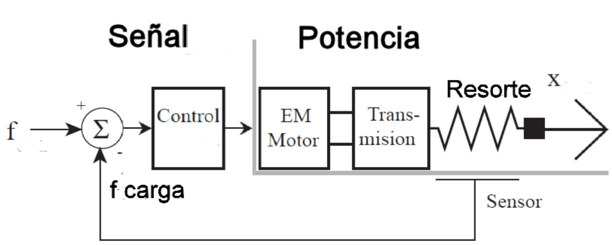


The brushless motor is the main actuator, which is used to provide additional power to drive the step [12]. The maximum power the motor that the supplied in the prosthesis is taken from to compensate in case of unforeseen accidents or a walking in irregular surfaces. The transfer function corresponds to a second-order system, like a mass, spring, damper mechanical model [13].

$$\frac{F_{out}}{F_{in}} = \frac{k_s}{Ms^2 + (b + k_{sat})s + k_s}$$

Where, k_s = Spring Constant, M = Motor Weight, b = Damping Force, $Fout$ = Output Maximum Force, Motor Saturation Constant ***ksat=Fsat/Vsat***, Vsat = Lineal Motor Speed, Fin = Input Maximum Motor Force k_s = Elastic Constant

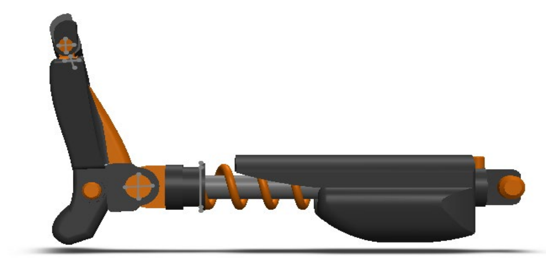
Figure 6. SEA Control System [2].



2. Methods

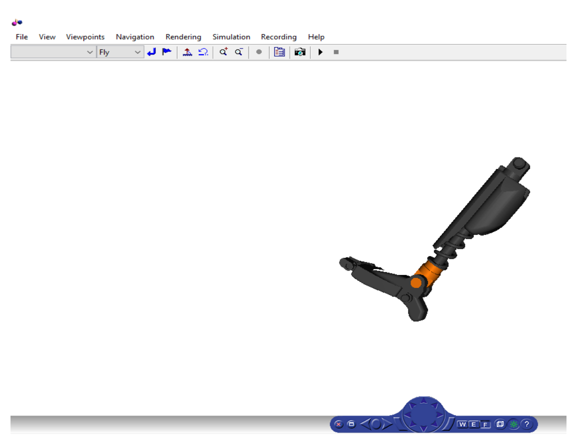
Let's start taking an assembly of a 3D ankle - foot Prosthesis model in SolidWorks [17], which is composed for joints and links. To create a joint between two prosthesis links in SolidWorks, we can constrain their axes and make them coincident. The next step of process requires the use of SimMechanics Link.

Figure 7. SolidWorks 3D Model [17].



When SimMechanics Link is Installed and linked in SolidWorks. The next step was import and attach this model to VR-World Simulink, generating an XML file that corresponds to the assembly, well as STL files which corresponds to the geometry. The XML file contains assembly structure and all elements' parameters required to generate an equivalent SimMechanics model and 3D virtual reality Simulink VR-model, such as reference frames, mass and inertia, color, and location of child STL files [18].

Figure 8. Simulink 3D VR-World Model Linked.



Finally, is possible combine the second order model of the SEA in Simulink with the 3D Solidworks mechanism. Designing a controller and makes test emulating different in the SEA behaviors and is possible using a PID controller to optimize the performance, according to Figures (7 to 9).

Like validation methodology for the SEA and his profits in the reduce power requirements, the 3D prosthetic ankle-foot mechanical model in Solidworks, was linked to Matlab and animated in Simulink VR-World. According to Figure 9, the second order transfer function of SEA model was tested in gait cycle phases of walking. Finally, in the simulation the torque-angle relationship of knee flexion and extension shows a spring-like characteristic. Additionally, in this step has a linear behavior and carry on the mechanical power required in every phase during the gait cycle, further demonstrating the potential utility of the SEA in the design of an ankle - foot prosthesis.

The model of the SEA in ankle-foot prosthesis was kinetically activated with an input test signal simulating loads during stance phase ankle-foot behavior in the Simulink model. It should be noted that in this analysis, each configuration yields essentially identical electrical power profiles, but very different ankle–foot kinematics; hence the focus of comparison will be on kinematic similarity rather than electrical power.

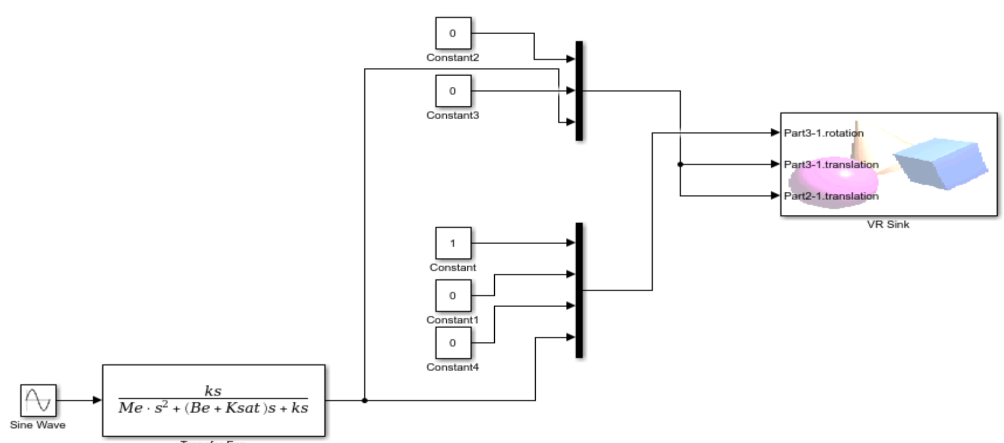
As a result of the varying torque-angle relationship across signals and walking speeds, the kinematic effect was assessed. The agreement between the SEA model and the experimental data were determined as a function of the SEA, where angular stiffness is referenced to the output.

3. Results

From the characteristic's curves of the simulation model, with the normal gait cycle to 1.5m/s (Figure 10), find the length of change and the required power for each gait cycle and show the animation. SEA was particularly well suited to take advantage of the linear region of the torque-angle relationship in human locomotion, often observed in prosthesis applications.

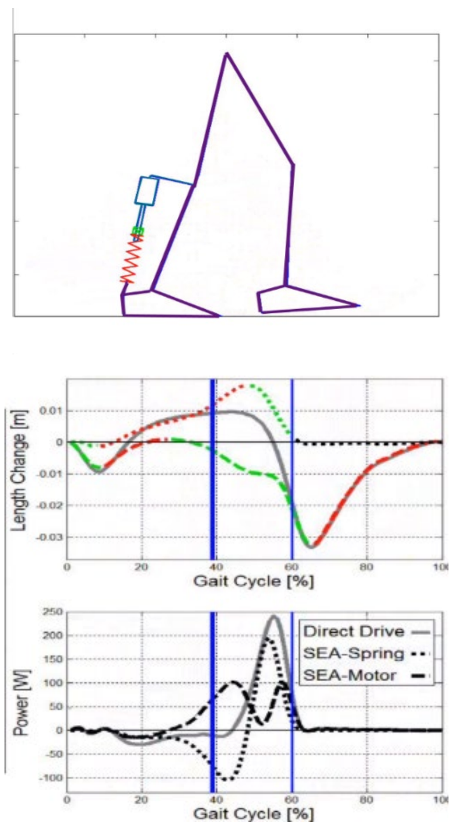
The series elasticity was chosen to emulate this linear region in stance phase ankle - foot flexion-extension, during this period to get electrical efficiency. It should be noted that there is a second semi linear region of the torque angle relationship. However, due to the limited torque needed during this region, it is not necessary to consider this region when tuning the series spring.

Figure 9. Matlab Simulink SEA Transfer Function Link to VR-World.



Since the electrical efficiency is a well result, the motor can be sized to the standard power requirements. During locomotion, there are spontaneous high-power modes that are required and the current design permits the use of a motor capable of these tasks. This is in contrast with a low power motor would be used to take advantage of the spring added in their design. Thus, the SEA is able to provide a wide spectrum of locomotory modes, despite being tuned for level ground walking. According to the next sequence Figure 10, that shows the characteristic curves and the behavior of SEA mechanism during normal gait cycle.

Figure 10. SEA Mechanism Behavior.



An increased in walking speed has been shown to increase the torque required by the ankle - foot prosthesis. This, combined with greater body weight, will increase the peak angle of ankle-foot flexion in the SEA ankle-foot due to series spring compression.

The SEA mechanism can easily capture and return this energy; thus, as the walking speed increases, the SEA ankle - foot becomes more efficient when compared to a traditional SEA. This is, however, a tradeoff as the ankle - foot flexion angle can be substantially greater than able-bodied kinematics.

4. Discussion

Adaptation to many kinds of terrain is an important aspect of walking. However, passive ankle-foot prostheses cannot provide this ability. For example, the Össur Proprio active ankle prosthesis is only able to reconfigure its ankle joint angle during the swing phase, requiring several strides to converge to a terrain-appropriate ankle position at first ground contact. Further, does not provide any of the stance phase power necessary for normal gait, and therefore cannot adapt net stance work with terrain slope. Also, the previous methodologies of the prototype powered ankle - foot prostheses used in this study exhibits an inherent adaptation to ground slope without explicit sensing of terrain variation [19].

The measured ankle torque profile of the prosthesis qualitatively matches those of a comparable intact individual for level-ground walking. The differences observed can be reasonably attributed to a number of factors resulting from amputation, differences in limb lengths, etc. In addition, the limited range of the prosthetic angle of the prosthesis from reaching the full range of motion of the intact ankle-foot.

5. Conclusions

Elastic Actuator elements in prosthetic devices can to reduce power and energy requirements for the system. Simulation showed that is not possible with current commercial motor technology to emulate human ankle behavior in detail for different situations like walking changing running speeds with a single motor and using a Serial Elastic Actuator (SEA).

The model verifies the restrictive design a terrain influence in the specifications in normal humans with ankle - foot prostheses. The transfer function provides the characteristic behavior which the motor is shown to be capable of satisfying optimally the restrictive specifications required by normal human of ankle-foot walking biomechanics.

Acknowledgments

Special acknowledgments to DIGITI - Smart Digital Systems research group and the engineering master of Distrital University.

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