

Analysis of Energy Storage And Return Foot Stiffness By Coupling Musculoskeletal And Finite Element Simulations

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Abstract: *Transtibial amputees currently have numerous prostheses in the market which are aimed at improving the control, cosmetics and comfort. Each of the three categories of prosthetic feet namely; conventional, energy storage and return, and bionic feet have different characteristics. Current studies reveal that energy storage and return feet offer better performance as compared with conventional prostheses. In this study, evolution of the prosthesis and the significance of mimicking human ankle-foot biomechanics is highlighted.*

Lower limb amputations occur due to Peripheral Vascular Disease, Diabetes, War and accidents .It is associated with mortality, morbidity, and disability. Transtibial amputees exhibit loss of plantar flexor musculature^[1, 7] resulting in greater intact leg stance times^[12] and asymmetrical gait comorbidity in their residual and intact legs^[2, 7]. Below-knee amputees lose the functional use of the ankle muscles, which are critical during walking to provide body support, forward propulsion, leg-swing initiation and mediolateral balance^[3, 7]. During early and pre-swing, amputees exhibit increased hamstring and rectus femoris activity on residual leg^[12]. Prosthetic foot do not allow sufficient dorsiflexion even on level terrain and possess inertia asymmetry. Further improvements ought to be incorporated to adjust the degree of dorsiflexion^[4], absorb shock on impact^[10], and improve inertia gait. Unilateral, transtibial amputees' clinical efficacy is dependent on appropriate prosthetic foot stiffness^[6]. Proper prosthetic foot selection with appropriate design characteristics is critical for successful amputee rehabilitation. Use of laminated composites in the manufacture of prostheses is vital due to their high stiffness and low density.

Many researchers have reported that unilateral below-knee amputees (BKA) walk asymmetrically and differently from able-bodied people^[1-10, 17]. Researchers have given varied reasons. It is generally believed that socket fit, prosthetic alignment, and prosthetic components (including prosthetic parts' weight and design) can all influence the gait of amputees^[2-7,12,18-19]. Others argue that due to loss of plantar flexor muscles, there would be greater intact leg stance times and asymmetrical gait comorbidity in their residual and intact legs. Moreover, degenerative changes in the lumbar spine and knees would occur due to the asymmetrical walking that overloads the musculoskeletal system^[7,12].

This research will come up with an analysis of the energy storage and return foot coupling musculoskeletal and finite element analysis with aim of improving amputee gait. The analysis of the foot is performed using the boundary conditions of ISO-10328 and ISO-22675. The prosthetic foot serves to substitute the loss of tendons and muscles of the intact foot due to amputation. Further series of computer simulation of ESAR foot is performed using Altair Hyper works 14.0 to investigate the effect of stiffness on the tibia section of foot, muscle activity, residual and intact ground reaction forces with aim of coming up with an optimal design.

The results of this study would add to the core knowledge regarding prosthetic feet features and their effects on gait, making them directly relevant to prosthesis design and prescription.

Keywords: *Transtibial amputee, Energy Storage and Return, Gait, Prosthetic foot, Dorsiflexion, plantarflexion, and Stiffness*

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I. Introduction

One of the most important goals of rehabilitation following a transtibial amputation is to return an individual to the highest functional level of ambulation possible .It involves proper socket design, alignment and proper choice of prosthetic componentry^[14]. Prosthetic feet of the 1980s were merely for ambulation. Amputees however have additional goals of being able to jump, run and participate in sports thus the need for an energy storing prosthetic feet^[14].

There are millions of individuals with gait disabilities, requiring either rehabilitation or permanent assistance. There are over two hundred and thirty thousand amputees in Kenya based on extrapolated statistics of the USA^[20]. Hans Mauch in 1950 and 1960s developed hydraulic ankle foot mechanism that adjusted its

alignment based on the surface. His design however had limitation due to malfunctioning of the hydraulic system. In 1981 the first energy storing foot was introduced; the Seattle Foot™ which incorporated Delrin Keel inside a polyurethane shell. Dynamic, STEN, SAFE and CARBON Copy II feet followed.^[14]

Prosthetic feet serve to provide unilateral, transtibial amputees with a high functional level of ambulation^[1-9]. Amputee rehabilitation require proper selection of a prosthetic foot to achieve clinical efficacy. Each amputee require different stiffness level^[5, 6]. Lower limb amputation has been defined as a complete loss in the transverse anatomical plane of any part of the lower [16] Transtibial amputees exhibit loss of plantar flexor musculature [1, 7] resulting in greater intact leg stance times^[12] and asymmetrical gait comorbidity in their residual and intact legs^[2, 7]. Below-knee amputees lose the functional use of the ankle muscles, which are critical during walking to provide body support, forward propulsion, leg-swing initiation and mediolateral balance^[3, 7]. During early and pre-swing, amputees exhibit increased hamstring and rectus femoris activity on residual leg^[12]. Prosthetic foot do not allow sufficient dorsiflexion even on level terrain and possess inertia asymmetry. Further improvements ought to be incorporated to adjust the degree of dorsiflexion^[4], absorb shock on impact^[10], and improve inertia gait. Unilateral, transtibial amputees' clinical efficacy is dependent on appropriate prosthetic foot stiffness^[6].

A promising strategy to improve amputee gait is to optimize ESAR foot design. The prescription practice is to combine design optimization with a rapid prototyping technology of selective laser sintering (SLS) to develop novel designs that improve the biomechanical quantities. Prosthetic foot (ESAR) is intended for provision of proper gait, support (due to absence of plantar flexors), and forward propulsion by storing and releasing elastic energy during stance^[7-10, 12]

Studies show that amputees using ESAR feet exhibit delayed residual leg activity and decreased residual hamstring activity as compared to when using solid ankle feet^[8]

Carbon Fiber has facilitated the manufacture of ESAR feet with high strength and lightweight^[5, 6]. These feet store elastic energy during the stance phase, and release a portion of it near toe-off to aid in propulsion and leg-swing initiation^[7]

This study is focused on developing a design that optimizes prosthetic foot stiffness to achieve a better gait performance. To test the effectiveness of the prostheses model, OpenSim software would be used for simulation and visualizing of the results. The Design should offer better ambulation, improved stiffness, better shock absorption and energy return.

The need for new designs/innovation is driven by the increasing number of amputations worldwide^[10] thus the need for a versatile and cost-effective design^[6].

Amputation in North America(Extrapolated Statistics)	
USA	2,055,587
MEXICO	734,717
CANADA	227,555
Amputation in ASIA(Extrapolated statistics)	
CHINA	9,091,933
JAPAN	891,331
SOUTH KOREA	337,636
Amputation in AFRICA(Extrapolated statistics)	
KENYA	230,874
ETHIOPIA	499,355
SOUTH AFRICA	311,139

Table 1: Amputation statistics

II. Literature Review

One of the most important goals of rehabilitation following a transtibial amputation is to return an individual to the highest functional level of ambulation possible .It involves proper socket design, alignment and proper choice of prosthetic componentry^[14]. Lower limb prostheses can improve the quality of life for amputees. Development of such devices, currently dominated by long prototyping periods, could be sped up by predictive simulations^[22].

Studies show that amputees using ESAR feet exhibit delayed residual leg activity and decreased residual hamstring activity as compared to when using solid ankle feet^[8]

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In this research the human-prosthesis interaction is modeled to produce a prediction of the amputee's walking kinematics. We obtain simulations of an amputee using an ankle-foot prosthesis by simultaneously optimizing human movements and prosthesis actuation, optimizing on stiffness.

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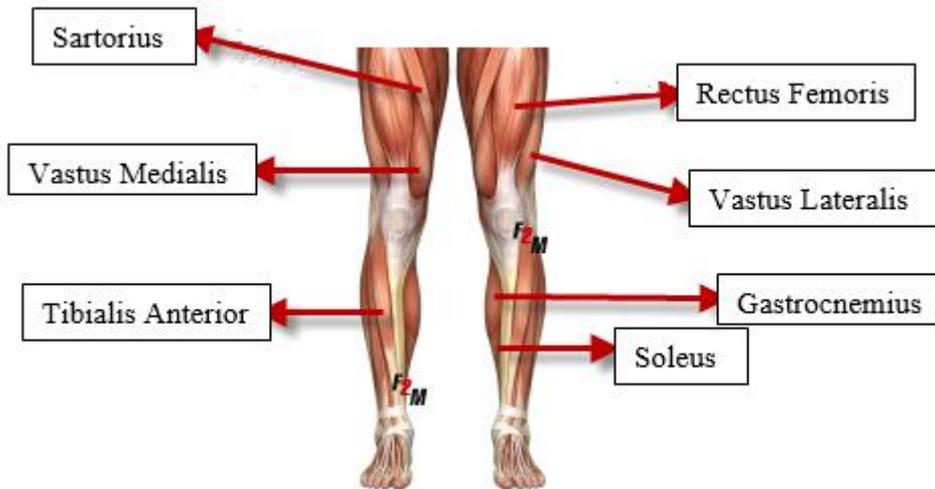


Fig.1: Leg muscles involved during gait

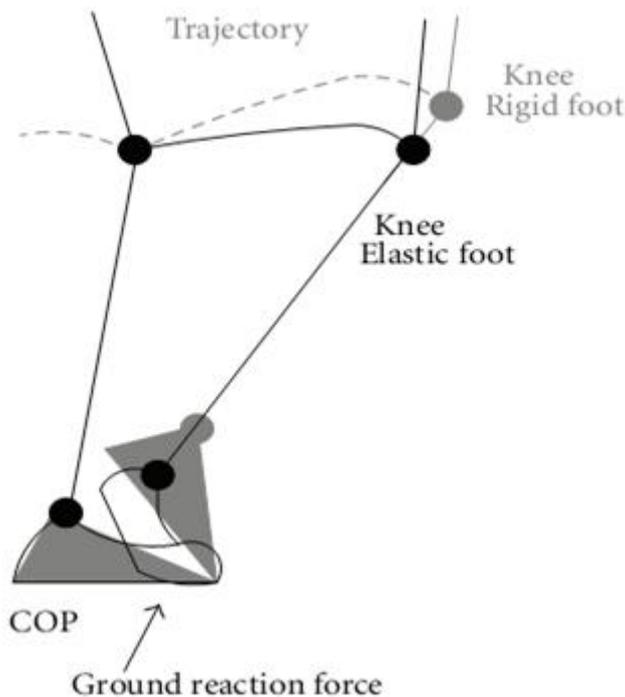


Fig. 2. Trajectory of knee joint for rigid and elastic foot

III. Methods

1.1. Musculoskeletal model

A 2D musculoskeletal OpenSim model was used. Gait simulations was performed using OpenSim simulation platform ^[32]. This facilitated the analysis and simulations of stance and swing phase of the gait. The modified model is shown in the figure below. It is a sagittal plane model of a unilateral amputee using an ankle-foot prosthesis (Fig. 1) with empirically-based properties. The human component has six rigid-body segments: one head-arms-torso segment, one segment for each thigh, one segment for each shank, and one segment for the biological foot. There are thirteen uni- or bi-articular muscles with constant moment-arms and Hill-type force-velocity relationships: eight muscles on the biological (non-prosthesis) side and five on the prosthesis side ^[22]. The model had a total of nine degrees-of-freedom.

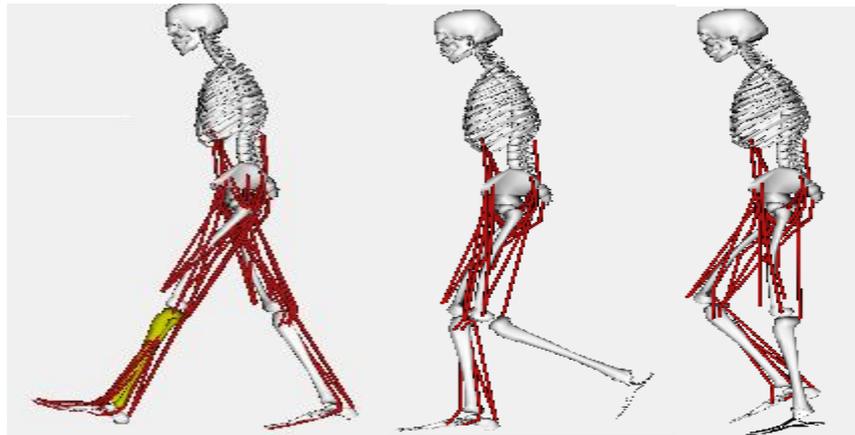


Fig. 3: A sagittal-plane rigid-body model, with six human body segments and one prosthesis segment, connected via revolute joints.

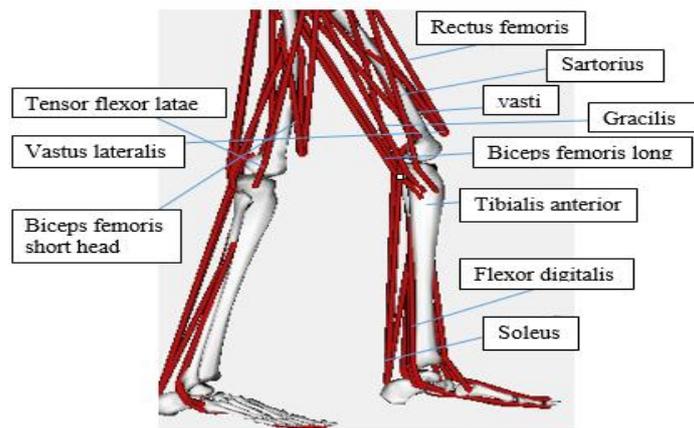


Fig. 4: Musculoskeletal Simulation model. The muscles are shown as straight lines. The gluteus maximus and quadriceps muscles include non-fixed via points for provision of appropriate lines-of-action through all joint excursions.

1.2. Dynamic optimization

Amputee and non-amputee walking simulations of a complete gait cycle were generated using OpenSim. Rigid multi-body dynamics was used to perform simulation. OpenSim uses computed muscle control and residual reduction algorithm to minimize error and estimate related muscle forces (individual muscle excitation onset, duration and magnitude).^[13] Ground reaction forces could be used as the boundary condition. The simulations mimicked non-amputee experimentally measured kinematic and ground reaction force data.

The objective function was formulated to minimize the squared error normalized by the inter-trial variability for each quantity tracked. The objective function was defined as

$$J = \sum_{j=1}^m \sum_{i=1}^n \frac{(Y_{ij} - \hat{Y}_{ij})^2}{SD_{ij}^2} \dots \mathbf{I}$$

Where Y_{ij} is the experimental measurement of variable j at time step i , \hat{Y}_{ij} the simulation data corresponding to Y_{ij} , and SD_{ij}^2 the average inter-subject variability of variable j at time step i . Specific quantities evaluated in the objective function included hip, knee and ankle joint angles, pelvis rotation, trunk translation in both the x - and y -directions, and horizontal and vertical ground reaction forces. The amputee (non-amputee) simulation began with residual (ipsilateral) leg heel-strike and concluded with the second residual (ipsilateral) leg heel-strike

1.3. Foot design

A modified CT-Scan ESAR foot was used in this study to optimize the prosthetic foot stiffness of unilateral transtibial prosthesis users. The foot is a dynamic energy return and storage type having flexible keel. The foot is designed to meet the needs of persons (amputees) with level two of functionality based on USA Health care Finance Agency (HCFA).

Stiffness in the ESAR feet was altered by modifying the thickness of the heel and keel section to achieve optimum stiffness using MIMICS and Geomagic softwares. The ESAR limb was carefully modified in order to match high strength and minimum weight criteria

The next step in the design process was to perform finite element analysis. Modifications of the prosthetic foot is done by adding material to the weak areas based on the stress distribution. The generated model was simulated again and again till an optimal design was achieved keeping in mind the minimum stress value, weight and flexibility. Dynamic analysis was then performed with the load and corresponding boundary conditions corresponding to heel strike, heel-toe contact and toe off.

Static and dynamic analysis were performed for standing and walking conditions respectively. The obtained results were compared to that of a normal person. The muscles behavior of the normal and unilateral transtibial Amputees were compared. The prosthetic foot mass and inertia properties are identical to the human foot unless otherwise stated.

Prosthetic foot conditions was analyzed in the Ls Dyna to determine the behavior when subjected to various loading conditions. Ambulation angle was between -10° to 20° . Effects of altering the prosthetic foot stiffness was also recorded. Normal gait data was obtained from OpenSim Gait2345 model. This data was compared with FEA data obtained from FEA of the ESAR foot in the Ls Dyna software.

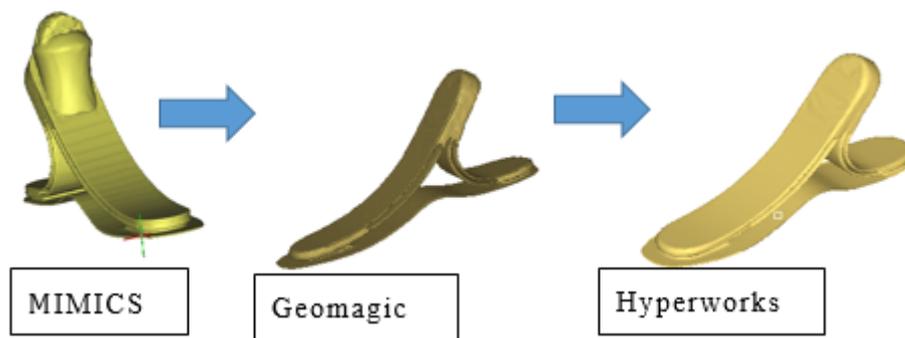


Fig. 5: ESAR Foot model

1.4. Tibia Model

The design used was the reversed engineered prosthetic limb obtained from CT-Scan. The tibia was obtained from Opensim then cut to imitate a unilateral transtibial amputee as in fig. 2 below.

Model	Density	Poissons ratio	Young's modulus (GPa)	Model
Tibia	Cortical bone	390	0.36	2,3
	Shaft of tibia	700-1600	0.36	4.2-9.6
	Cancellous bone	100	0.30	0.49

Table 2: Material properties of Tibia section

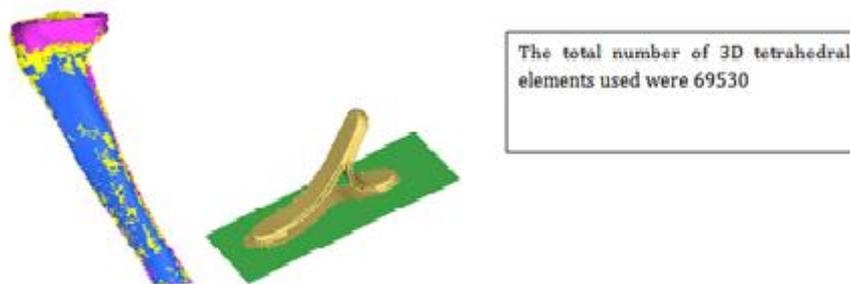


Fig.6: Unilateral Transtibial Amputee Tibia and Foot Model (Meshed)

1.5. Material Selection

The behavior of materials under dynamic loading offers a challenge especially modeling it in order to perform engineering analysis. For this research, Mat54 (Composite) of LS Dyna was used. MAT54 model uses Chang matrix failure criterion as shown below.

For the tensile fiber mode;

$$\sigma > 0$$

Then

$$e_f^2 = \left(\frac{\sigma_{aa}}{x_t}\right)^2 + \beta \left(\frac{\sigma_{ab}}{s_c}\right) - 1 \begin{cases} \geq 0 & \text{failed} \\ < 0 & \text{elastic} \end{cases} \dots\dots\dots \text{II}$$

Upon failure: E1 = E2 = G12 = v12 = v21 = 0.

For the compressive fiber mode,

$$\sigma > 0$$

Then

$$e_c^2 = \left(\frac{\sigma_{aa}}{x_c}\right)^2 - 1 \begin{cases} \geq 0 & \text{failed} \\ < 0 & \text{elastic} \end{cases} \dots\dots\dots \text{III}$$

Upon failure: E1 = v12 = v21 = 0.

For the tensile matrix mode,

$$\sigma > 0$$

Then

$$e_m^2 = \left(\frac{\sigma_{bb}}{y_t}\right)^2 + \left(\frac{\sigma_{ab}}{s_c}\right)^2 - 1 \begin{cases} \geq 0 & \text{failed} \\ < 0 & \text{elastic} \end{cases} \dots\dots\dots \text{IV}$$

Upon failure: E2 = v21 = G12 = 0.

And for the compressive matrix mode,

$$\sigma > 0$$

Then

$$e_d^2 = \left(\frac{\sigma_{bb}}{2s_c}\right)^2 + \left[\left(\frac{y_c}{2s_c}\right)^2 - 1\right] \frac{\sigma_{ab}}{y_c} + \left(\frac{\sigma_{ab}}{s_c}\right)^2 - 1 \begin{cases} \geq 0 & \text{failed} \\ < 0 & \text{elastic} \end{cases} \dots\dots\dots \text{V}$$

Upon failure: E2 = v21 = v12 = 0 = G12 = 0.

X_c = 2Y_c for 50% fiber volume.

1.6. Simulation

The parameters used in simulation was as shown in the table below.

Property	Symbol	Value
Density	P	1.52g/cm ³
Modulus in direction 1	E ₁₁	127GPa
Modulus in direction 2	E ₂₂	8.4GPa
Shear modulus	G ₁₂	4.21GPa
Major Poisson's ratio	v ₁₂	0.309
Minor Poisson's ratio	v ₂₁	0.20409
Strength in tension direction 1	σ _y ^{1t}	2200MPa
Strength in tension direction 2	σ _y ^{2t}	48.9MPa
Strength compression direction 1	σ _y ^{1c}	1470MPa
Strength in compression direction 2	σ _y ^{2c}	199MPa
Shear strength	σ _y ¹²	154MPa

Table 3. Simulation parameters as adopted from ^[49]

For analysis purposes, the connection between the prosthetic feet to the tibia section is made rigid by use of rbe-2 elements. The total number of 3-D tetrahedral elements used is 168128.

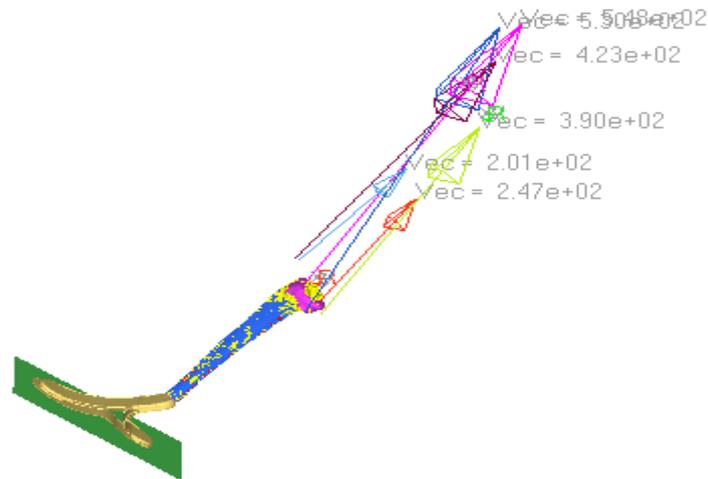


Fig. 7: Loaded meshed model of lower limb amputee (Tibia and prosthetic foot)

Six functionally independent muscle groups were used to drive the model. Each muscle was modeled as vector origination from the tibia section. The muscles include gluteus maximus, sartorius, Vas_int_r, bifemsh_r, rect_fem_r, and grac_r muscles.

The muscles used in our analyses were those above the knee i.e. Bifemsh_r, Grac_r, Sar_r, Tfl_r, Vas_int_r, and Rect_fem_r. The corresponding graphs of the muscle forces with respect to time i.e. one complete stance phase are as below.

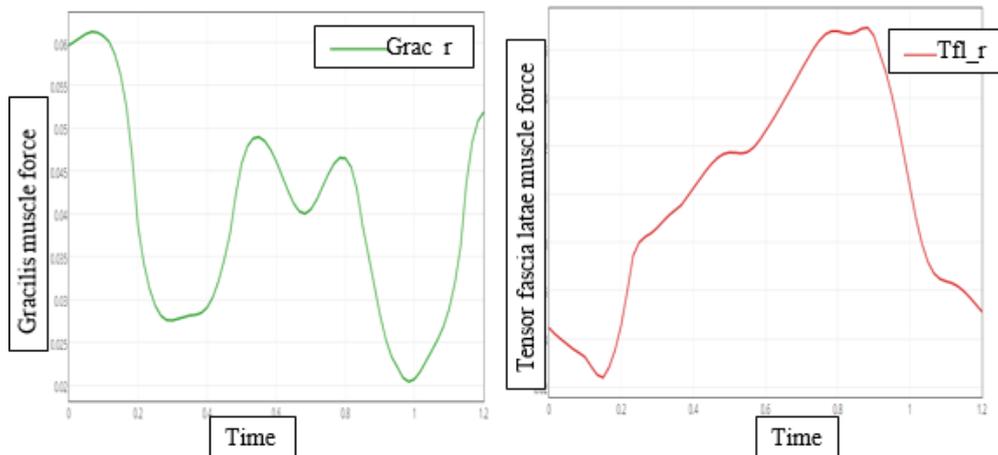


Fig.8: Sartorius and Biceps femoris short head Muscle forces

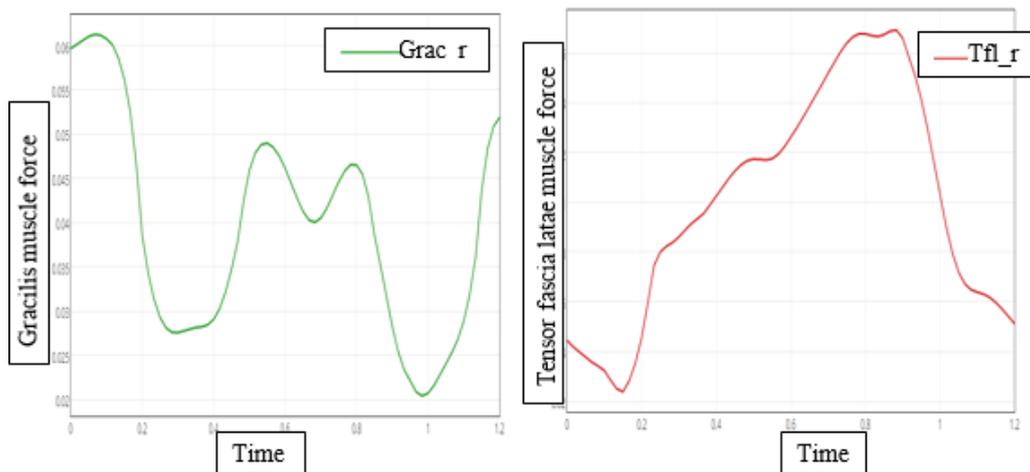


Fig. 9: Gracilis and Tensor fascia latae muscle forces

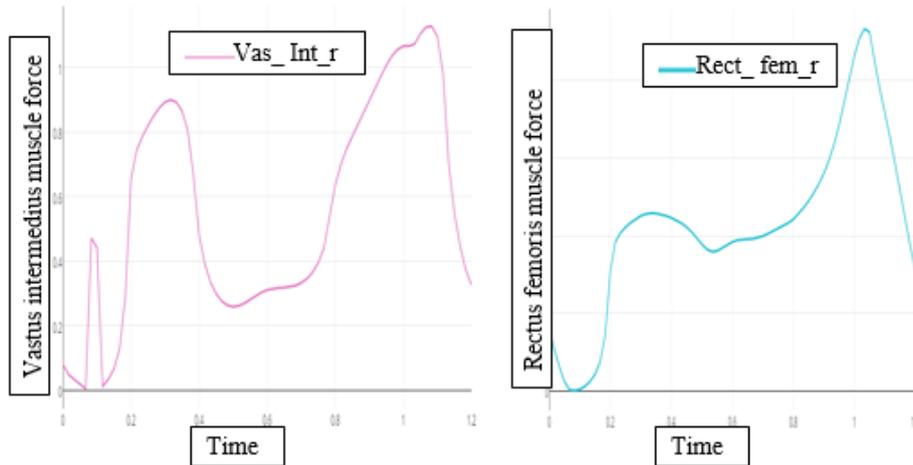


Fig.10: Vastus intermedius and Rectus femoris muscle forces

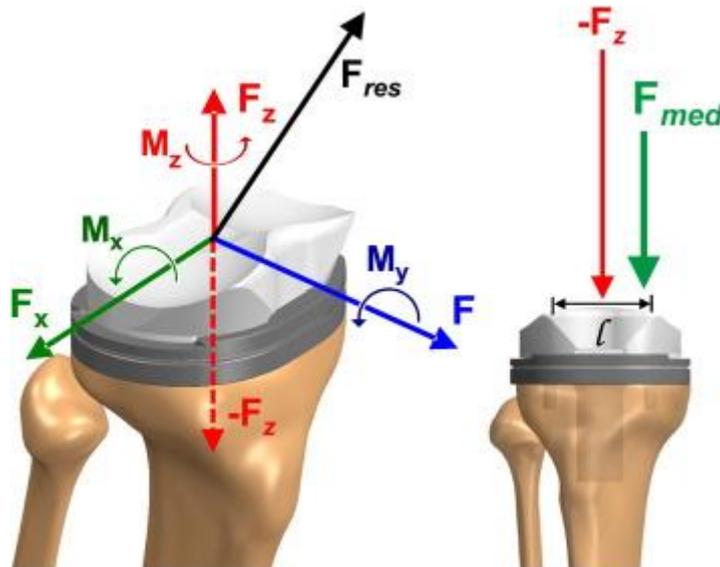


Fig. 11: Illustration of force/moment direction during analysis [4]

To be more anthropomorphic, the resistance to foot dorsiflexion has to increase nonlinearly from a very low value to a value able to lock the ankle and stop dorsiflexion.

Surface contact algorithm of the Altair Hyperworks software was utilized in order to simulate the interaction between the ground and the prosthetic foot. The time traces of these knee joint forces and moment were used as inputs to the FE model as the dynamic loading conditions by applying them to tibia bone as the only driving forces and moment. The initial conditions of the system were based on motion data at heel strike.

IV. Results

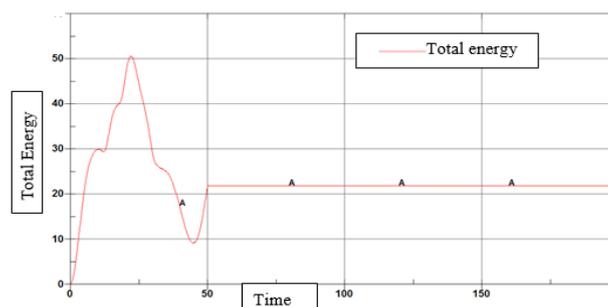


Fig.12: A graph of Resultant displacement against Time for the combination of ESAR foot and Tibia.

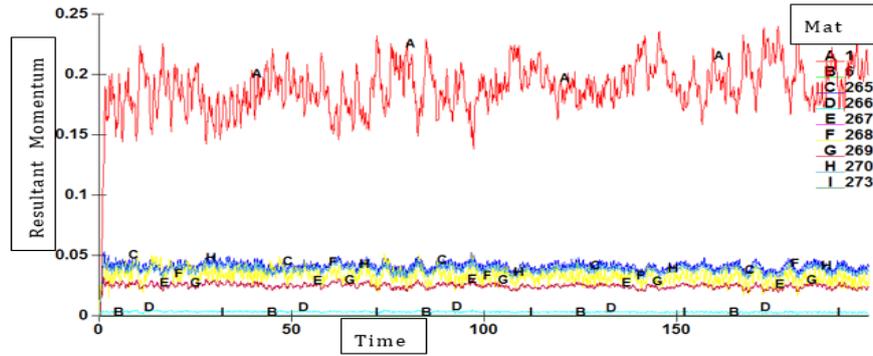


Fig.13: A graph of Resultant Momentum Versus Time of tibia and ESAR foot (Dynamic loading)

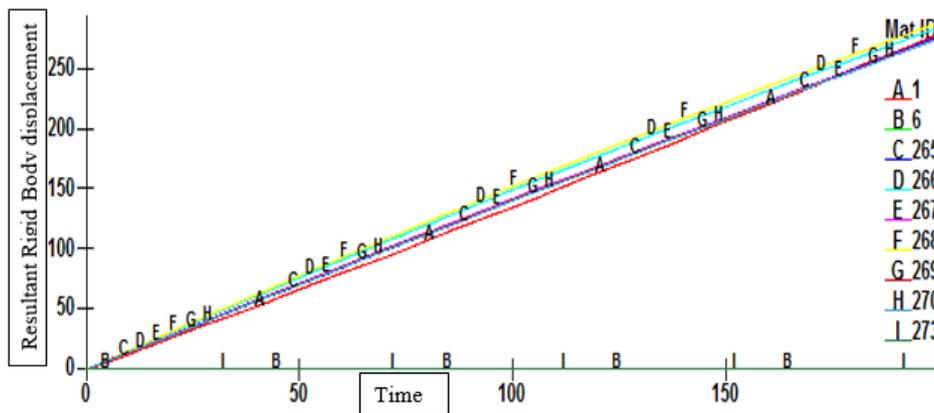


Fig.14: A graph of Resultant displacement of Tibia and ESAR foot (Dynamic loading)
2.

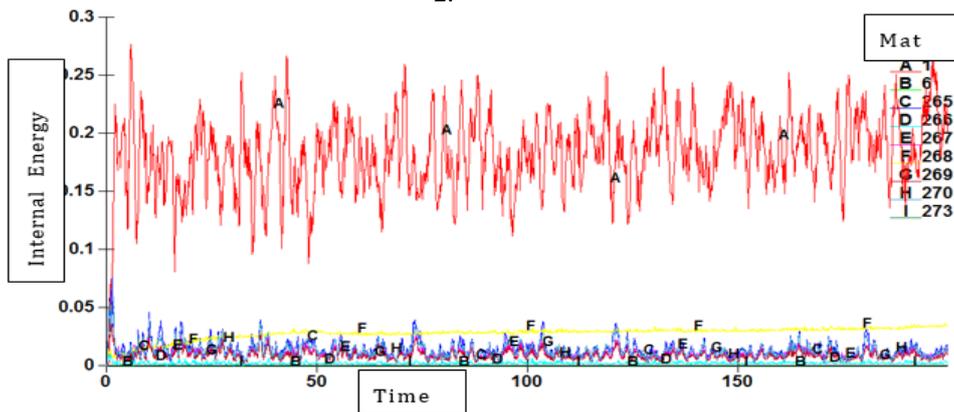


Fig.15: A graph of Internal Energy Versus Time of Tibia and ESAR foot (Dynamic loading)

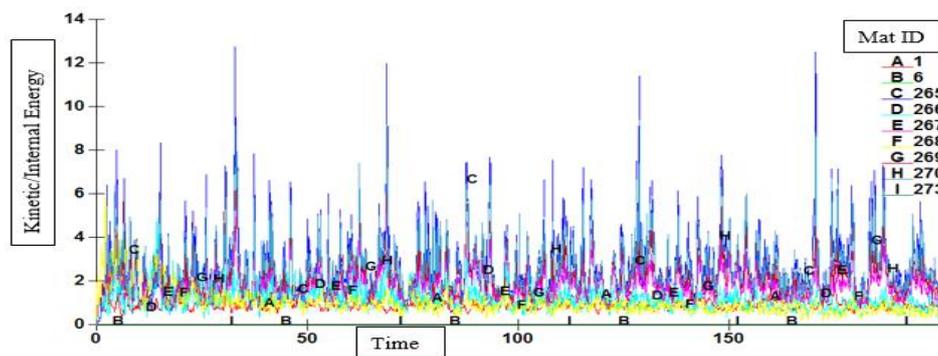


Fig.16: A graph of Kinetic/Internal Energy versus Time (Dynamic loading)

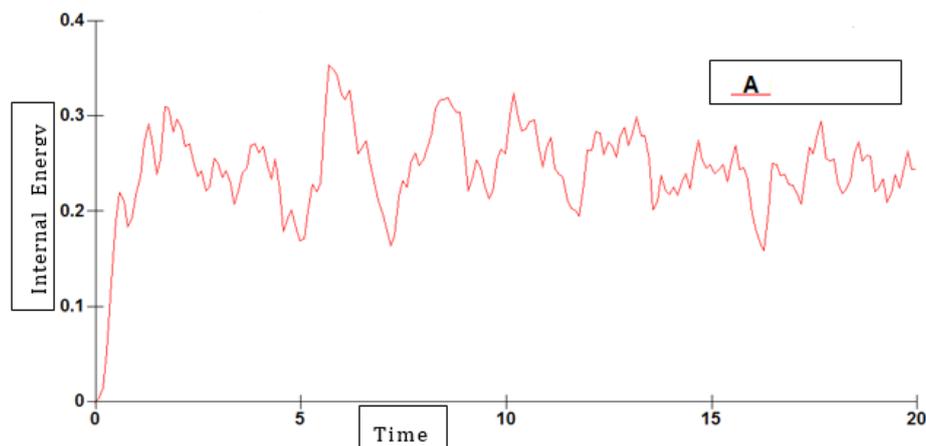


Fig.17: A graph of Internal Energy versus Time of Tibia and ESAR foot (Dynamic Loading)

V. Conclusion

In conclusion, in all walking conditions, the ESAR prosthetic foot provided an increased range of motion in both plantarflexion and dorsiflexion and better mimicked the normal leg. As compared to previous versions, it is the most suitable prosthetic foot for patients both in its potential of energy storing and energy return efficiency. Moreover, it better mimics the normal human foot. The dorsiflexion angle for ESAR foot was way better than previous designs like SACH among others.

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