



Passive Below-Knee Prosthetic Legs Suitable for Sub-Saharan Africa: A Survey

Ameer Mohammed^{1*}, Ubaidullahi M Abdullahi², Bello Yekini¹, Bashir G. Ibrahim³, Bashir B. Muhammad¹, and Ahmed A. Shinkafi².

¹Mechatronic Engineering Department, Air Force Institute of Technology, Kaduna, Nigeria.
(e-mail:belloyekinnitopa@gmail.com, +2348134766903).

²Aerospace Engineering Department, Air Force Institute of Technology, Kaduna, Nigeria.

³Centre for Defence Procurement Logistics and Supply Chain Management, Air Force Institute of Technology, Kaduna, Nigeria.

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ABSTRACT

With the increase in security challenges in Nigeria, there is a rise in the number of below-knee amputees, especially for members of the armed forces. Other causes of amputation include trauma such as ageing, accidents and surgery related to peripheral vascular disease, cancer, and infection. Loss of limbs affect patients' quality of life by leading to job losses, limited freedom of mobility, and increased difficulties in daily activities; thus, requiring prosthetic limbs to enhance their quality of life. Hence, this paper surveys passive below knee prosthetic legs designed to provide easy rotation of the leg to enable it to adjust to rugged terrain as is common in sub-Saharan Africa – with marshy and desert terrains. The goal of the survey is to establish a suitable prosthetic foot for the poor that will mimic normal human gait such that amputees are better-off with the prosthesis compared to other supporting devices. The reviewed foot types were selected from the most commonly used in parts of the world with nearly similar terrain and climate as sub-Saharan Africa. A suitable foot was identified from the reviewed foot types under factors including cultural, cost and terrain. The recommended foot under the factors considered was the Jaipur foot. It was established that the most suitable prosthetic foot design for sub-Saharan Africa does not translate to the cheapest design, but a robust design suitable for unstructured terrains.

Keywords: Ankle Foot, Below-Knee Prosthesis, Passive Prosthesis and Prosthetic Legs

Introduction

The human body needs stability and balance when standing, squatting or moving, and thus requires feet. Amputation of a foot significantly reduces the amputee's ability to perform normal daily activities such as walking (Arifin, *et al.*, 2014). The basic goal of any type of prosthesis is to restore function to a physically challenged individual. Prosthesis is an artificial device used to replace a missing body part which is intended to restore a degree of normal function to amputees. There is a natural human desire to feel whole and complete, and providing artificial devices has helped to satisfy this need (Al-Khazraji *et al.*, 2012). Lower limb amputees rely on prosthetic feet for stability and weight acceptance during stance, as well as and to facilitate symmetrical and energy-efficient gait (Rajula *et al.*, 2021). Lower-limb amputees lack the function of the gastrocnemius and other distal leg muscles that are important to normal

walking. The gastrocnemius and soleus are the primary ankle plantar flexors. Both muscles have distinct and important roles in healthy gait, while, the gastrocnemius accelerating the leg into swing during late stance phase, and the soleus accelerates the trunk forward (Willson *et al.*, 2020). The work by (Van der Krogt *et al.*, 2012) showed that human gait is sensitive to gastrocnemius and soleus weakness. So, a prosthetic foot should mimic the functions of the human-ankle foot complex. The large number of amputations during World War II influenced research in prostheses. Since then, a lot of new and efficient prostheses were invented. These prostheses were designed to assist return the full functionality of the lost body part to amputees. A natural foot can store energy during stance and returning it to the amputee to assist in propelling the body forward at push-off (Zeng, 2013). The most important characteristics of a human foot used in its functional operations are dorsiflexion,

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eversion, impact absorption, energy return and the torque generated at the ankle. These are the characteristics used in determining an appropriate prosthesis, according to prosthetic feet patients (Rihs, 2001, Stevens, *et al.*, 2018).

Presently in Nigeria with the lingering insurgency and security challenges, a major issue is the rise in the number of below-knee amputees, especially for members of our armed forces. Other causes of below-knee amputation include ageing, accidents (environmental accidents, farming accidents, motor vehicle accidents) and surgery related to peripheral vascular disease, cancer, and infection. Amputation could also be because of inadequate access to basic healthcare, which could exacerbate diseases like diabetes, gangrene, ischemic disease and infection; thus, leading to limb amputation. All of these causes are adding to the number of amputations (Strait, 2006). The increase in the amputee population, especially in Nigeria and parts of sub-Saharan Africa, makes research in prosthesis management necessary. Hence, this paper presents a survey on passive below-knee prosthetic leg that are suitable for use in sub-Saharan Africa. Some of the notable prosthetic foot types analysed include the Solid Ankle Cushion Heel (SACH) Foot developed by J. Foort and C. W. Radcliffe of the Prosthetic Devices Research Project, Institute of Engineering Research, University of California (Berkeley), the Jaipur foot which was developed by Professor P. K. Sethi for barefoot amputees and the prosthesis could be tailor-made in front of the user in less than an hour, the Niagara Foot, the Shape and Roll Foot which produces a good approximation of the step of a physiological foot, ESAR among others.

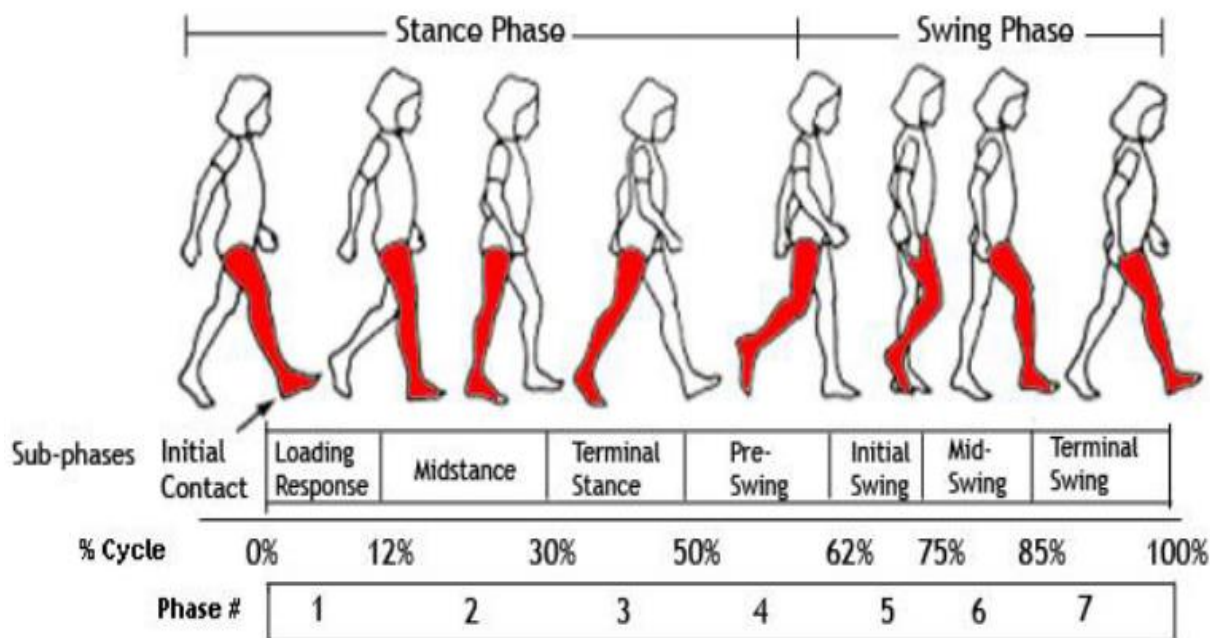
The rest of the paper is organized as follows. Section 2 presents some basic knowledge of normal human gait and existing prosthetic ankle designs. The general

categories of lower-limb prostheses consisting of passive, active and semi-active prosthesis are explained in Section 3. Section 4 presents an overview of the current state-of-the-art passive prostheses that could be suitable for use in sub-Saharan Africa. Section 5 evaluates the candidate prostheses based on the major characteristics needed for walking such as eversion, dorsiflexion, energy return, ankle torque and impact absorption at heel strike of the reviewed feet. Future trends in below-knee prosthetic devices and concluding remarks are presented in Sections 6 and 7.

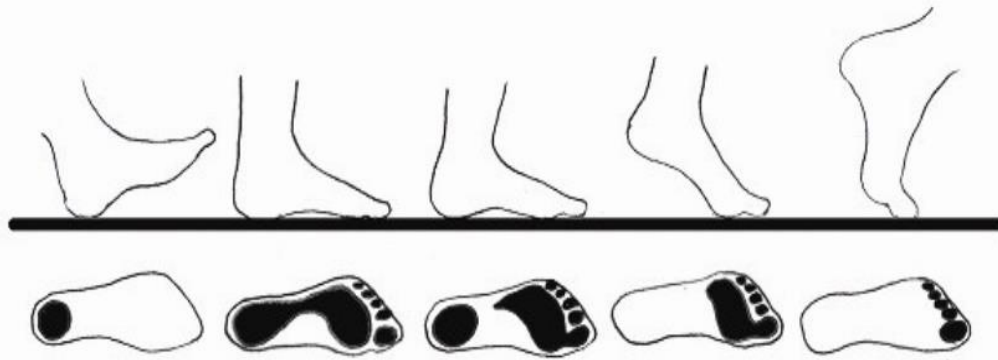
Gait Cycle Analysis

Gait Cycle

Gait is a periodic phenomenon, and gait cycle can be defined as the time interval between two consecutive occurrences of the same recurring event during walking, where one period is known as a stride. A stride is defined as the duration from when the heel of one leg strikes the ground (0% of stride) to when that same heel strikes the ground again (100% of stride). Figure 1 describes the various phases involved in the gait cycle (Herr, 2008; Rajčúková, *et al.*, 2014). In Figure 1, the key points of the stride include heel strike (0%), foot flat (FF) is 8%, heel off (43%) and toe off (TO) is 62%. As shown in Figure 1, a stride can be broken into stance phase (or weight bearing) (0-62%) and swing phase (62-100%). Different phases are defined according to the position and kinematic relationships of the limb while walking. Energy return is the ability of the prosthesis to store energy. Energy is stored in the operation of a sound foot during the stance phase of walking and is released on the transferal of weight. This is important in order to provide enough momentum for the rest of the prosthesis to roll over the foot. A sound foot releases an average of 15.74 Joules, stores an average of 14.18 Joules and therefore has an efficiency of 119.6%.



(a) The stride consisting of stance and swing phases.



(b) Pressure distribution of foot during gait.

Figure 1: Gait phases and Pressure distribution of foot during gait (Borjian, 2008).

Dynamic Analysis of Gait

The dynamic analysis of walking in humans is an important aspect of gait research. It includes the displacements, forces, moments and energies of the system. During gait cycle, the ankle behavior is described using different characteristics. The presented parameters were all measured in the plane of progression, which closely corresponds to the sagittal plane of the body in normal walking. The relative (in-plane) rotation between the foot segment and the lower leg segment is described by the term dorsiflexion and plantarflexion (as shown in Figure 2). The movements of the ankle joint with 3 degrees-of-freedom (DOF)

are referred to as abduction/adduction, dorsi/plantar flexion and ankle inversion/eversion. In Figure 2, dorsiflexion implies the bending of the foot towards the knee, and on the other hand, plantarflexion is the rotation of the foot away from the knee. The angle between the foot and the leg will simply be called the 'ankle angle'. When the leg is perpendicular to the foot's sole, the ankle angle equals zero and the angle is defined positive in dorsiflexion and negative in plantarflexion in Figure 2. The acceptable dorsiflexion of normal foot is 3-5° in walking range. While eversion is the ability of a human foot to roll from side to side, and mostly occurs when walking on uneven surfaces as shown in Figure 2.

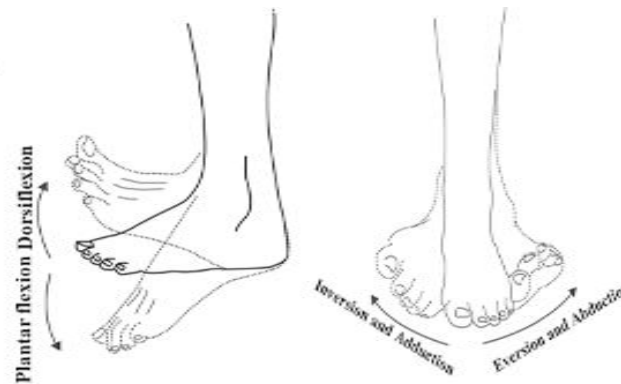


Figure 2: Dorsiflexion and plantarflexion angle of ankle (Kalita et al., 2020).

The angle of the leg and the torque around the ankle joint were measured in Figure 3. Average ankle behaviors are presented in Figure 3. They were obtained from a large number of natural human gait analyses (Winter, 2005). According to Winter's data, the deflection range of the ankle joint is approximately 27°. The ankle starts from a neutral position at heel strike and then goes to negative (plantar flexion) angle so that the forefoot is lowered to touch the ground. During the mid-stance phase and terminal stance

phase, the ankle joint angle becomes positive (dorsiflexion). Then a large negative angle is developed during pre-swing phase. The ankle joint is moved back to neutral position during the swing period to prepare for the next gait cycle. The torque profile of the ankle has a small negative torque followed by a substantial positive torque. While the body continues to move forward, a large positive torque is created to propel the body forward (Northeast, 2018).

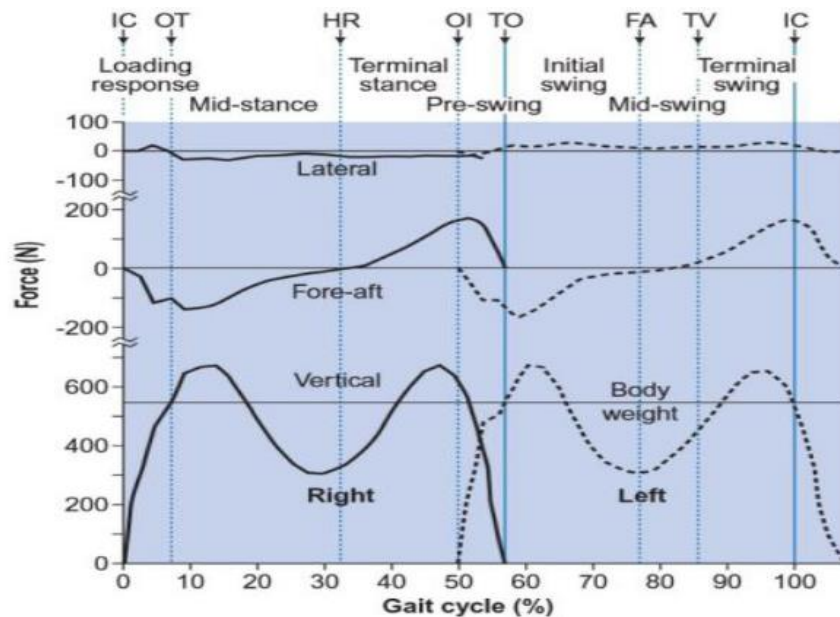


Figure 3: The ground reaction forces during different stages of the gait (Kubasad et al., 2020)

In dynamic analysis, three distinct phases are often used to describe the torque-angle relationship in the stance period (in Figure 3): controlled plantar flexion (CP), controlled dorsiflexion (CD) and powered plantar flexion (PP). During CP and CD, the muscles of the leg and foot are used to absorb the energy during

these phases. In PP, additional energy from the muscles is used to provide push-off. The instantaneous slope of the torque-angle curve indicates the instantaneous ankle stiffness. These three phases are each characterized by the instantaneous stiffness observed during the phase.

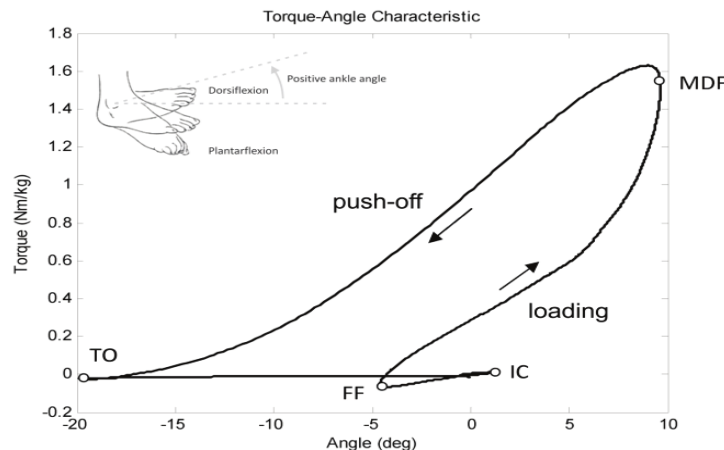


Figure 4: Torque angle plot of the human ankle in a natural gait (Brackx et al., 2013).

The CP phase, between time initial contact (IC) and FF in Figure 4, corresponds to initial contact phase and load response phase. During this phase, the foot initially contacts the ground, and a nearly linear stiffness relationship is observed. The CD phase is the interval from FF through the maximum dorsiflexion angle (MDF). It corresponds to mid-stance phase and terminal stance phase. A nonlinear stiffness relationship occurs during this phase, and it shows that the ankle stiffness significantly increases with an increasing ankle angle. The interval between MDF through TO is described as the PP phase. The ankle achieved the maximum ankle angle and then the maximum ankle torque at MDF. Between MDF and

TO, the ankle torque decreases linearly with decreasing ankle angle. The ankle stiffness changes from positive to negative at MDF. The sign change of ankle stiffness relates to the active ankle behavior (Hitt, et al., 2009, Azocar & Rouse, 2017). A large amount of torque used to propel the body forward needs to be generated while the ankle angle is decreasing. The reason why this graph is considered useful is because it shows another important parameter: the ankle stiffness. This is especially important in prosthetics, where attempts are made to mimic the ankle behavior by using elastic springs. These are used to create the same ankle stiffness as for a healthy ankle.

3 General Categories of Lower Limb Prostheses

There are three different techniques used to design ankle prostheses, passive, active and semi- active. Passive uses the body power to trigger and contain only mechanical elements to control the relative motion between the shank and foot. While, active are simulated by applying power to the device, thus contain onboard control system and actuator to control the relative motions of the foot (Kubasad *et al.*, 2020). In semi-active prostheses, it uses a mixture of both power of the user as well as external power. The traditional lower limb prosthetic devices have been around for decades and are passive in nature to mimic the biomechanical behavior of the human leg. They are cheap easy to use and light. However, they are unable to adapt biomimetically due to static mechanical characteristics. The active prostheses have had a swift development over the past two decades, since they provide energy to propel the body forward. Active prosthesis has the potential to mimic more features to achieve near natural movement but also have some disadvantages in weight, size, cost, electric power as well as complex control systems. Semi-active prostheses provide a trade-off between active and passive prostheses. It has optimum weight, size, complexity, and cost. During operation, the semi-active prosthesis uses the power of the patient during the stance phase while provides the power to the prosthetic device in the swing phase. The approach allows the user to practice body power as an input to balance a portion of the human gait cycle (Kuo and Donelan. 2010). One of the most significant challenges in the development of a semi-active lower limb prosthesis is to provide self-powered actuation competencies comparable to the biological counterparts (Asif & Tiwana, 2021). Considering the challenges of high technology solutions consisting of power, actuation and complex control mechanisms, passive prostheses are recommended for use in sub-Saharan Africa. The following section summarizes the various types of passive prostheses that have been in

use. It provides an overview of the performance of several existing state-of-the-art passive ankle prosthesis designs. The overview emphasizes their mechanical properties and their ability to enable the amputees to regain normal gait functions.

4 Passive Prostheses

Most commercially available prostheses are passive devices. These devices use passive components such as springs and dampers in various forms. In general, the different types of passive design are the conventional Solid Ankle-Cushioned Heel (SACH) Foot was known to be the first prosthetic foot that is attached directly to the prosthetic shank to achieve the foot desired movement through the elastic behavior of heel and toe and the Energy Storage and Return (ESAR) Foot. Also, there are other prosthetic feet that have curves with shapes mimicking human physiology which deform to simulate a true foot, these include the Niagara Foot, the Shape and Roll Foot and to some extent, the Jaipur Foot. Although not as spring-like as the others, the Jaipur Foot has achieved great success in India because of its high-durability and aesthetics which allow the user to wear sandals and kneel to pray – an important cultural activity in many regions. It is manufactured with a compression mold that encases the mechanical foot in vulcanized rubber, shaped like a human foot (Arya *et al.*, 2001).

In the past, different researchers investigated different techniques of reinforcing the foots to increase their mechanical properties and structure (Hashim *et al.*, 2016). The production and improvement of the prosthetic foot are important for rural amputees in developing countries. In addition, the prosthetic foot durability dictates the strength of the whole prosthesis. Several prosthetic feet have been developed to achieve rural amputees' requirements in developing countries. Table 1 gives the characteristics of some notable passive prosthetic feet. The table presents the capability, cost, mode of production of each foot type.

Table 1: Characteristics of different types of passive prosthetic foot.

S/N	Passive Feet	Characteristics
1.	Jaipur foot	<ul style="list-style-type: none"> ● The Jaipur foot looks like a real foot ● Has the ability to bend in all directions enough to allow a person to squat and walk on uneven terrain. ● very low-cost alternative costing less than \$5 USD to make and can be fabricated in less than 3 hours. ● The foot is made from wood and sponge rubber and then heat-molded using iron molds.
2.	SACH foot	<ul style="list-style-type: none"> ● Wedge shaped heel cushion ● Varying heel densities ● Varying material keel ● Rubber body and bolt attachment
3.	Seattle light foot	<ul style="list-style-type: none"> ● Delrin II dynamic response keel ● Lightweight design ● Sculpted life-like appearance ● Available in various skin color

4.	Seattle natural	<ul style="list-style-type: none"> ● Excellent roll over during walking ● Slim, sculpted, natural-looking cosmesis ● 3/8" heel rise ● Low to medium-low activity level ● Available in various skin colors
5.	Sure flex	<ul style="list-style-type: none"> ● 100% carbon fiber composite keel and heel ● Polyurethane foam sole available in three densities
6.	Niagara	<ul style="list-style-type: none"> ● Made from a single piece of Delrin plastic ● The shape of the foot is to provide energy return ● Cosmetic appeal of the foot and the ability to wear shoes created a few hurdles for this foot.
	STEN	<ul style="list-style-type: none"> ● three-piece maple keel with dense foam-rubber cylindrical plugs ● two high-density rubber blocks, which act as compressible spacers between the wooden blocks ● a layer of high-strength fabric attached to the plantar surface of the wood blocks ● plantar reinforcement fabric bonded to a rubber sole ● an exterior polyurethane foam matrix

SACH Foot

The SACH foot is usually made of wood and rubber. As shown in Figure 5, the prosthesis mainly consists of a wood keel, a cushion heel, belting and plastic covering. It usually uses a bolt to attach to the pyramid or leg pylon. The wood keel is designed to provide base stability and rigidity. The SACH foot mimics the appearance of the human ankle well, but does less well in other functions. The cushioned rubber heel absorbs shock at impact and the belting allows for bending of the foot to mimic human ankle deflection. The SACH foot is the most common prosthetic design and an excellent choice for amputees with an expected low-activity level. It is simple, inexpensive, durable and comfortable compared to dynamic-response or energy-storing feet that are more complicated and costlier (Zeng, 2013).

Daher performed an experiment to examine the durability of SACH foot subject to cyclic testing. The results showed that permanent deformation and changes occur at heel within only 5,000 cycles performed for an amputee of 100 kg (Daher, 1975. Kumar & Mukherjee, 2020). Also, the influence of the flexibility forefoot on the gait of 14 unilateral transtibial prosthesis users. The results of this work recommend solid-ankle prosthetic foot designs with more flexible forefoot sections can cause a “drop-off” influence in the late stance phase and through the transmission of loading between prosthetic and

contralateral limbs (Klodd, 2010, Kishner, 2018). Arya found out that the shock absorption capacity of the SACH foot was better when compared with the Jaipur and Seattle foot types. The Jaipur foot allowed a more natural gait and was closer in performance to the normal foot. None of the prostheses significantly influenced the locomotor style of the amputees (Arya et al., 1995). The SACH foot design has been continuously improved by new materials that can "store and release energy," allowing for controlled mobility. They are classified as dynamic elastic response feet in terms of functionality (Olewi, 2021; Mohammed, *et al*, 2020; Postema *et al.*, 1997; Lenka & Kumar, 2010).

Absence of mechanical devices and joint in the prosthetic foot eliminates maintenance problems due to frictional wear, joint looseness, and instability. The heel cushion provides, at heel contact, a shock absorption more than equivalent to the plantar flexion of a conventional ankle. As the amputee walks over his prosthetic foot following compression of the heel cushion, the foot begins to simulate ankle dorsiflexion. The toe approaches the floor, the prosthetic shank rotates forward over the foot, and the heel cushion decompresses. Disadvantages of non-articulated feet are the limited range of plantar-flexion/dorsiflexion, difficulty with inclines due to heel compression, lack of adjustability for different heel heights, and little torque-absorption capability.

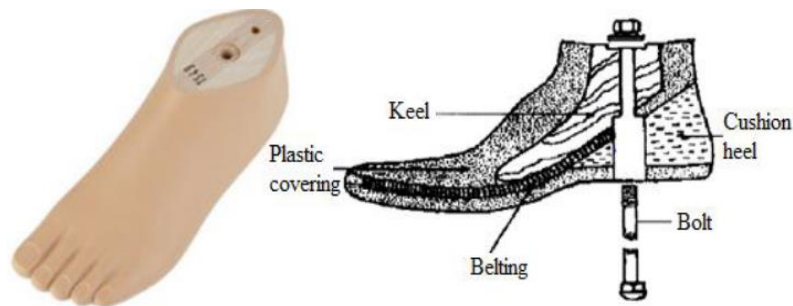


Figure 5: Models of SACH Foot (Zeng, 2013).

The Jaipur Foot

As shown in Figure 6, the Jaipur foot consists of three separate blocks. The micro-cellular rubber for the hind foot, a laminated wood for the ankle and a rubber block for the forefoot. These are appropriately shaped and wrapped by an inner layer of tire cord rubber and an outer layer of skin- soft rubber, with tough rubber for the sole. The Jaipur foot came into existence in response to socioeconomic and cultural needs of squatting, cross-legged sitting and barefoot walking of Indian amputees. It has not been recognized in the developed world presumably due to a lack of awareness and the absence of its biomechanical evaluation (Zmitrewicz, *et al.*, 2006). There is also a popular impression that it is meant for barefoot walking, although amputees do use it satisfactorily with shoes. It was observed that Indian lower-limb amputees preferred to use crutches rather than prostheses fitted with a SACH (Arya *et al.*, 2008). The SACH foot was not suitable for the floor-sitting lifestyle in a warm country such as India which were different from those of the chair-sitting habits of the cold countries of Europe. Arya compared between Jaipur, Seattle, and SACH prosthetic foot: the study was devised to compare both the shock absorption capacity and the influence on gait style. The conclusions indicated that the Jaipur foot performed

close to the human foot than the Seattle and SACH feet. The SACH foot had a superior impact resistance than Jaipur and Seattle feet (Arya, 2008).

The large sponge rubber block at the hind foot can provide multidirectional movements, thus acted as a universal joint, the foot could rotate on the leg. This allowed the amputee not only to squat and sit cross-legged, but also to walk easily on uneven terrain, since the foot could adapt to the underlying surface. It could be used in mud and water as it is waterproof. Its normal appearance was a great advantage since it did not need a shoe and amputees could walk barefoot, which followed the custom of not bringing shoes into kitchens or places of worship. The foot was sturdy and tough and could withstand the stresses of everyday use (Arya *et al.*, 2008). The only drawback to the Jaipur limb was that the height could not be adjusted. The manufacturing of the Jaipur foot needs to be standardized to improve life of foot (Huber *et al.*, 2017). The work in (Kabra & Narayanan 1991, Bhargava, 2019), developed a cyclic loading system that measured cyclic dorsiflexion to consider the functional parameters of the prosthetic Jaipur ankle-foot before and after long-term cyclic loading. The work discovered that the prosthetic foot has good efficiency and significant mobility in three planes based on the results on the Jaipur foot



Figure 6: A sagittal section of a Jaipur foot (Arya *et al.*, 2008).

ESAR Foot

Expectations of amputees have increased to a greater desire to participate in recreational and sporting activities, these advances have led to the evolution of several new designs of ankle foot assembly. The most

exciting amongst this is the energy storage and return (ESAR) foot, of which the Seattle foot is the most popular among the ESAR. A typical ESAR foot is shown in Figure 7. Even though the increasing popularity of these new designs have not affected the dominance of the SACH foot. Since the first ESAR

foot, the Seattle Foot, was introduced in 1981, many newer and more sophisticated designs have been developed to improve the performance of the ankle prosthesis. These prostheses are designed to store energy in early stance and return it to the amputee to propel the body in late stance. People who have had a lower limb amputation prefer energy storing and return (ESAR) feet over solid ankle cushioned heel (SACH) feet. While ESAR feet have only a minor impact on gait economy, other functional advantages should account for this preference. A simple biomechanical model suggests that improved gait stability and symmetry may explain some of the subjective preference differences between the two feet (Houdijk *et al.*, 2018).

An early ESAR foot looks very similar to a SACH foot. It usually incorporates a flexible keel and foam or rubber shell. It is the flexible keel that acts as an elastic spring, absorbing and releasing energy during

push off. A totally different type of ESAR foot, the Flex Foot, is now the most common prosthesis. It typically contains a flexible carbon fiber shank and a heel spring. Except for the ankle and foot portion, the Flex Foot extends the length of prostheses and allows the entire device to flex, to absorb and return energy. ESAR is preferred because of its elastic behavior for gait symmetry. ESAR has greater push-off power, a high center of mass velocity and extended forward propagation as compared to SACH. Other prosthetic foot manufacturers followed a similar strategy and incorporated a flexible keel surrounded by foam and/or a polyurethane cosmesis. Such feet include the Dynamic foot and C-Walk (Otto Bock HealthCare GmbH), SAFE (Campbell-Childs, Inc., White City, OR), Carbon Copy (Ohio Willow Wood Co., Mount Sterling, OH), STEN (Kingsley Manufacturing, Costa Mesa, CA) among others (Versluys *et al.*, 2009).



Figure 7: ESAR Feet (Upper: Seattle Feet from Trulife; Bottom: Flex Feet form Össur) (Zeng, 2013)

Some researchers have reported the results of biomechanical analysis of ESAR and conventional components. Nielsen *et al.* 1988, elicited subjective feedback from seven transtibial amputees after quantitative analysis, finding positive feedback regarding the ESAR foot used (Flex-Foot) (Nielsen *et al.*, 1988, Müller, *et al.*, 2019). The users reported that the ESAR device allowed for increased velocity in walking and enhanced stability on uneven ground. Conversely, users found very slow and downhill walking to be easier with the conventional SACH foot. Torburn *et al.*, similarly analyzed the subjective feedback of five transtibial amputees following a biomechanical study on ESAR feet (Torburn *et al.*, 1990, Arifin, *et al.*, 2014). All subjects preferred an ESAR device over a conventional SACH foot; however, all respondents chose the Seattle foot over the Flex-Foot, often citing the appearance of the foot greatly affected their choice of limb. All subjects chose the foot that provided them the greatest walking

velocity, even if the velocity change did not reach statistical significance (Hafner *et al.*, 2002, Canete, 2021).

4.4 Niagara foot

The foot is made from a single piece of delrin plastic formed to imitate a normal human foot. The shape of the foot is to provide energy return. The Niagara foot (in Figure 8) is simple and inexpensive, sturdy. Although weight and energy expenditure are not an issue. The poor cosmetic appeal of the foot is the major challenge of this foot. The foot appears to be less stable as it produces irregular motion throughout the gait cycle (Strait, 2006).

The work by (Schmite 2007) analyzed the biomechanical properties of the Niagara foot model. In the analysis, heel and toe stiffness responses were evaluated by ISO standard 10328 using a rate of 5 mm/min. For displacement via a loading plate angled at 15° and 20° on the heel and toe, respectively and a force of 1600 N.



Figure 8: Niagara foot (Strait, 2006).

The Shape & Roll Foot

The Shape & Roll prosthetic foot is made of copolymer plastic and was developed to conform to the roll-over shape of the able-bodied ankle-foot system during walking (Sam et al. 2004, Carpenter, et al 2008). The roll-over shape effective rocker shape that is formed during walking is established through the closing of cuts on the forefoot surface of the Shape and Roll prosthetic foot as shown in Figure 9 (Hansen and Childress, 2010). For this foot, the roll-over shapes in able-bodied subjects do not change appreciably for conditions of level ground walking, including walking at different speeds, while carrying different amounts of weight, while wearing shoes of different heel heights. In fact, results suggest that able-bodied humans will actively change their ankle movements to maintain the same roll-over shapes. This rocker more appropriately mimics the forward

movement of the center of pressure that is seen during walking and provides a better approximation of vertical excursions of the center of mass (Gard and Childress 2001). It is difficult to create a light weight yet extremely durable cosmetic coverings for the foot. One of its limitation was the original Shape and Roll prosthetic foot, which was designed to accommodate shoes of only small (or zero) heel heights.

The Shape & Roll prosthetic foot (shown in Figure 9) was used because of its capacity to lower its effective foot length ratio with only slight changes while maintaining other qualities including heel stiffness, mid-foot stiffness, and weight of the foot. Cuts on the forefoot surface of the Form & Roll prosthetic foot are closed to create the roll-over shape. To blind the patient to the visual differences between the foot conditions, a simple removable shell composed of soft foam material (Pelite1) and a nylon stocking were employed to cover the foot during the experiment (Hasen and Childress, 2010).

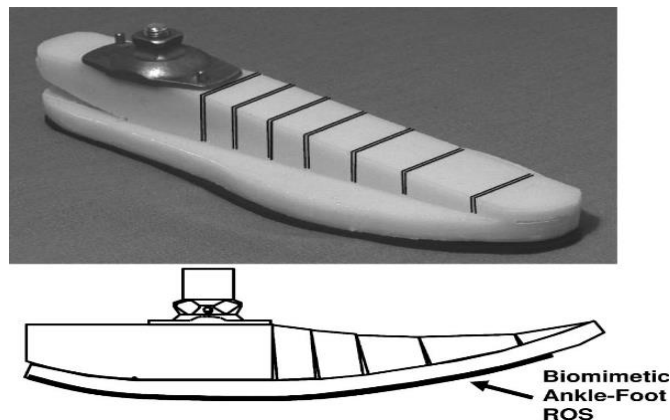


Figure 9: The Shape and Roll Foot. Top: The Shape and Roll Prosthetic Foot (saw cuts are accentuated in the picture using black lines). Bottom: The Shape and Roll Prosthetic Foot was designed to conform to the biomimetic ankle-foot roll-over shape (ROS) as the cuts close during walking (Hansen and Childress, 2010).

Seattle Lightfoot

Seattle Lightfoot (shown in Figure 10) is a prosthetic foot for people with lower extremity amputations of all ages and activity levels. The Seattle Foot was created with the goal of making prosthetics more cost-effective, with fewer component parts, less maintenance, and a more consistent response. Although the original anatomical molds resembled a real foot, some amputees preferred the blank-foot. The

Seattle Foot offers both an aesthetic model and a monolithic keel, which acts as a spring to help you run faster. The keel is made of Delrin, a strong and lightweight material/design. In comparison to previous prosthetics, it is designed to create a more natural and suspended step. The foot enhancements alter the prostheses overall dynamics (Gard *et al.*, 2011, Dante, 2006). The effective energy-absorbing system in the Seattle Foot's design revolutionized prosthetic development. Various designs before the

invention used metal springs or supple foam to absorb the amputee's weight as well as upward forces from the ground. The simple energy-absorption prosthetics were only effective for slow walking. The Seattle Foot

was the first device to attempt to mimic the natural movement of the foot during various human gaits (Deborah, 1985a, Peterson, 2012)



Figure 10: Seattle Lightfoot (Gard et al., 2011)

Stored Energy (STEN) Foot

A three-piece maple keel with dense foam-rubber cylindrical plugs at the metatarsophalangeal and tarsometatarsal joints makes up the Stored Energy (STEN) foot (Figure 11). A strong woven belting runs underneath the keel. In the back of the foot, a cushion heel sits beneath the belting. Synthetic foam rubber molded to look like toes and toenails surrounds the keel and cushion heel. With a mild spring action, motion at the keel segments accommodates ground

irregularities and smooths the stance transition. Because of its anterior support, which stabilizes the knee during foot-flat, the STEN Foot is preferred over the SACH Foot for active below-knee amputees. This allows for a longer stride and more loading during push-off for vigorous amputees. The increased knee stability can help amputees who are not very active. It is especially useful with hydraulic knee units, which typically require full knee extension to engage (Yang, et al., 2018).

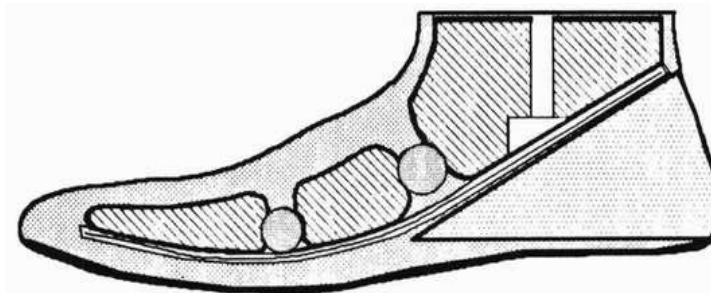


Figure 10: STEN foot (Hafner, 2002)

5 Comparative Analysis

This section discusses the comparative analysis of the existing foot types; SACH foot, Jaipur foot, ESAR foot and others as shown in Table 2. Bearing in mind that prosthetic feet are designed to perform different functions depending on the desired application.

However, most prosthetic feet only show some desired characteristics selected by the designer needed for normal walking. This can be used in helping an amputee to make the choice of a prosthetic leg (Pikhart, 2009). The scope of this review considers the major characteristics needed for walking such as eversion, dorsiflexion, energy return, ankle torque and impact absorption at heel strike of the reviewed feet.

Table 2: Comparative analysis of different type of prosthetic foot (Hansen and Childress, 2010).

Foot Types	SACH	Jaipur foot	The Shape and Roll Foot	ESAR	Niagara	Seattle Lightfoot	STEN foot

Mechanisms and Materials	It simulates plantarflexion at heel strike by the compression of the cushioned heel and provides dorsiflexion by the flexible belting.	Made from polyurethane.	Made of copolymer plastic and was developed to conform to the roll-over shape of the able-bodied ankle-foot system during walking	This prosthetic foot can move in the same way that SACH feet do. The foot store and release energy when the individual is walking or during push off	Made from a single piece of delrin plastic.	Made of Pylon - Titanium and Aluminum die-cast	Made of Wood/rubber /fabric/ composite
Advantages	Waterproof, cheap, durable and suitable for athletes.	<ul style="list-style-type: none"> • Superior range of movement over SACH. • Waterproof • Articulation at the ankle for rotation. 	Invariant to non-level walking surfaces.	Elastic with great push-off power.	Very simple, inexpensive, practical, energy-efficient and sturdy.	economically effective, with fewer component parts, less maintenance, and a more consistent response.	<ul style="list-style-type: none"> • provide smoother, more graded performance than a SACH Foot • can be used for running and active sports. • require less adjustment on the part of the amputee and prosthetist
Limitations	<ul style="list-style-type: none"> • Rigid keel that cannot bend. • Poor toe-off with quicker transition to the foot. 	<ul style="list-style-type: none"> • Damage severity dependent on usage and location. • Limb not standardized and adjustable. 	Only useful for a youthful group with lower limb problems.	Expensive with stiff keel.	<ul style="list-style-type: none"> • Poor cosmetic appeal • Inability to wear shoes • Less stable due to irregular motion throughout the gait cycle . 	The Seattle Lightfoot with Titanium Pyramid cannot be used with R.O.L rotators or other devices that require modification of the keel.	It is the heavier, more costly SACH foot.
References	(Zeng, 2013)	(Arya et al., 1995.) (Strait, 2006) (Huber et al., 2017)	(Hansen & Childress, 2010)	(Hafner et al., 2002), (Zeng, 2013), (Houdijk et al., 2018)	(Strait, 2006)	(Dante, 2006)	(Fereshtenejad et al., 2014)

The selection of the prosthetic feet was undertaken on the basis of the patient needs, material characteristics, cost, function, society, comfort and how effectively the foot reproduced the desired characteristics in reference to that of a human foot. The SACH foot imitates the appearance of the human ankle well but

does less well in other functions. From the viewpoint of its biomechanical performance, the foot is known to cause drop-off effect and asymmetry in amputee gait (Balaramkrishnan et al., 2020). The Jaipur foot has good biomechanical performance, but has only one drawback, the limb could not be adjusted. Also, lack of standardization is a major hurdle to further

improvement of the quality of the Jaipur Foot. Efforts to improve its design should run concurrently with efforts to standardize supply chain and manufacturing processes (Huber et al., 2017).

6 Future Trends for Below-Knee Prosthetic

Available passive prosthetic feet do not do well in restoring the function of the ankle and foot due to limited range of motion. They only nearly mimic the anatomical foot and ankle. The alignment of the prosthesis is also important, as continued usage of a misaligned prosthesis has been associated with chronic back pain, muscle atrophy, pressure ulcers, and osteoarthritis. Patients feedback after wearing the prosthetic limbs is a key source for design optimization. It is beneficial to improve the design parameters of prosthetic devices thereby facilitating quality of life to the wearer.

The work by (Asif & Tiwana, 2021) found the count and percentage of work for lower limb body parts. Overall trends analyzed are; 228 patents (48%) for the knee, 131 patents (28%) for the ankle, 105 patents (22%) for the foot, and 11 patents (2%) for the hip design. Design trends and advancements in patents show maximum work (48%) for knee design and minimum work (2%) for hip joints for human lower limb body parts. Quantitative analysis showed (22%) of designs exist for the foot reflecting the gap in the designs and demands for the transtibial foot (Asif & Tiwana, 2021). As there is huge opportunity for improvement in lower-limb prostheses foot design and especially hip joint. This should inspire many researchers to introduce state-of-the-art technology in different components of the lower limb prosthesis. The prosthetic component market contains manual alignment adapters, which can translate, tilt, or rotate the prosthetic components to achieve a fixed spatial arrangement between the socket and foot. It will be difficult to satisfy the demands of amputees in the society, and at the same time produce cost-effective limbs that fulfill the needs of amputees in low-income countries. The greatest prosthetic dissatisfaction areas were found to be color and weight, because people are very selective about the color and weight of prosthetic limbs. Nonetheless, transtibial amputees tend to be more satisfied than transfemoral amputees (Luza et al., 2021).

7 Conclusion

This paper examines passive below-knee prosthetic legs that allow for easy leg rotation to adapt to harsh terrain, such as marshy and desert terrains, which are typical in Sub-Saharan Africa. The survey's purpose is to find a viable prosthetic foot for the poor that will resemble natural human stride and provide amputees with a higher quality of life than alternative supporting devices. It concludes that there is no single foot that is perfect for every amputee. Knowing the available options will enable you to discuss this issue clearly with your prosthetist, evaluate the advantages and disadvantages of different feet so that the best choice for various individual can be selected. In comparing

the potential benefits of one foot over another, physicians and prosthetists focus on the level of suitability and functional aspects of the prosthetic foot, according to the user's personalized needs. From the information presented, the Jaipur foot is the most suitable for use in developing countries due to its adaptability to various terrains, simplicity and low cost. While the ESAR foot is recommended for the developed countries.

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