University of Strathclyde

Department of Biomedical Engineering

Comparison of a Natural Fibre Based Composite Prosthetic Foot with a Carbon Fibre Prosthetic Foot for Use in Low-Income Countries

Melanie Simons

201490663

Supervised by Dr Arjan Buis

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Declaration of Authenticity and Author's Rights

This thesis is the result of the author's original research. It has been composed by the author and has not been previously submitted for examination which has led to the award of a degree.

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Abstract

There is a large and growing number of lower limb amputees across the globe. In the Western world, the main cause is diabetes, whereas in low-income countries, wars and residual landmines are a major contributor. Many lower limb amputees in these low-income countries do not have access to a prosthetic device mainly due to their unaffordable cost.

State-of-the-art prosthetic devices are constructed from carbon fibre, which exhibits the ideal characteristics of being lightweight, stiff and durable, but unfortunately this comes at great financial cost. Natural fibre based composites may offer an affordable alternative material for prosthetic manufacture. However, in order to obtain the high-performance such as that seen with the carbon fibre prosthetic devices, the design is key to its success.

In this study, a prosthetic foot prototype was produced from birch veneer with a similar design to that of a commercially available carbon fibre prosthetic foot. Static proof tests were carried out according to BS EN ISO 10328:2006. In addition, the deformation under these loads was tracked using Photoshop software. The results of the natural fibre and carbon fibre prosthetic feet were then compared.

The overall observation was found to be that the natural fibre prosthetic foot was considerably stiffer compared to the carbon fibre prosthetic foot. In an attempt to increase its flexibility, two thinner natural fibre prosthetic feet were constructed and tested, but without successful outcomes.

The initial natural fibre prosthetic foot produced did comply with BS EN ISO 10328:2006 for the static proof test up to test loading level P5. Even though more extensive research is to be carried out regarding its design, it shows considerable potential for low-cost, high-performance prosthetic manufacture for low-income countries.

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Chapter 1: Introduction

1.1 Introduction

Amputation of a lower limb is extremely challenging on both a physical and psychological level, affecting not only the patient but also their family and friends. Amputees face the possibility of losing their jobs and can become highly dependent on those around them for daily support, often resulting in them being socially isolated (Joint Master Project, 2015).

In England alone there are over 25,000 major lower limb amputees (where the amputation level lies anywhere above the ankle) with 44% of these cases due to distal ulcers as a result of diabetic complications (Ahman *et al.*, 2014; Holman *et al.*, 2012). With the correct monitoring and treatment of diabetes, the World Health Organisation (WHO) claims that 80% of diabetic foot amputations can be prevented. This would reduce the use of diabetic healthcare resources by 15-25% (World Health Organisation, 2005a). The International Diabetes Federation states that 80% of people with diabetes live in low- and middle- income countries where poorer health systems and inferior diabetic management lead to countless numbers of lower limb amputations (International Diabetes Federation, 2013).

The ProPortion Foundation (2014) has estimated that the low-income communities of Colombia are home to approximately 40,000 lower limb amputees. With a population of just over 46 million, this leaves 1 in every 1,156 people missing a lower limb. This extraordinarily high number is not solely due to diabetes; landmines and traffic accidents are predominant contributors. Being either a past or present conflict zone is a connecting factor amongst countries with a high prevalence of amputees. Iraq and Afghanistan count 1 in every 987 and 1 in every 631 people respectively as being an amputee (NBC News, 2010). Residual landmines from past conflicts in Angola has earned it the unwelcome title of the country with the highest rate of amputees in the world, where up to 1 in every 334 people are missing a body part (International Committee of the Red Cross, 2014). Incomplete health records in these developing countries and conflict zones means that these prevalence rates may be inaccurate, leaving the number of lower limb amputees worldwide very hard to estimate (NBC news, 2010). However, in 2004 the WHO estimated that approximately 25.5 million people required a prosthetic or orthotic device (which replaces or stabilises a limb respectively) in Africa, Asia and Latin America (World Health Organisation, 2005b).

Prosthetic devices can be manufactured from a countless number of different materials. Carbon fibre is considered to be at the top end of the range of materials, providing state-of-the-art, high-performance characteristics for lower limb prostheses; it is lightweight, stiff and durable. Being the ideal material means that it does not come cheap, at approximately £1,500 per prosthetic foot. Even in high-income countries (e.g. those in Western Europe such as the United Kingdom), this cost limits its use. With 80% of the world's disabled people living in low-income countries, finding a low-cost alternative material suitable for prosthetic manufacture is vital. Numerous attempts have already been made with only limited successes (World Health Organisation, 2005b).

One category of alternative material that has shown great potential is natural fibre based composites. These cost a fraction of the price of carbon fibre and offer lightweight, biodegradable, and sustainable characteristics, all of which are desirable properties for prosthetic components. Their performance and durability as a prosthetic device are determined by their design, hence this is of utmost importance and holds the key to its success of delivering high performance prostheses. With a suitable design, low cost prosthetic devices which maintain the high performance characteristics associated with the carbon fibre foot could be delivered to low-income countries, revolutionising the lives of its amputees and their families.

1.2 Aims of the Project

The aim of this study was to determine the potential of birch veneer as a low-cost alternative to carbon fibre for prosthetic manufacture in low-income countries. Birch veneer was used as the natural fibre composite for the production of a prosthetic foot prototype. The prototype was manufactured to be similar in respect of its design to a commercially available carbon fibre type.

The main objectives of this study are as follows:

- To mechanically test the natural fibre prosthetic foot prototype alongside a commercially available carbon fibre alternative according to BS EN ISO 10328:2006, by means of an Instron ElectroPuls E10000 and compare the results obtained
- To determine the differences in deformation upon loading between the natural fibre prosthetic foot prototype and the carbon fibre prosthetic foot
- To optimise the design of the natural fibre based composite foot prototype in order to obtain more comparable results to the carbon fibre prosthetic foot.

Chapter 2: Literature Review

2.1 Lower Limb Amputations

In the UK the majority of lower limb amputations occur as a result of vascular disease with 60% caused by atherosclerosis and 30% due to diabetes mellitus. This can be associated with a sedentary lifestyle, poor diets and smoking. Trans-tibial (a.k.a. below-knee) amputations are the most common at 50.9% of all lower limb amputations in the UK (Stewart, 2008).

The main cause of lower limb amputations in low-income countries (those with a gross national income of $1,045 or less per person) varies (TheWorldBank.org). In current or previous conflict area such as Cambodia, Iran and Afghanistan, war, landmines and other explosive remnants of war account for up to 80-85% of amputees (Strait, 2006). The population in other low-income countries experience insufficient health care due to a corrupt government and ineffective health care systems. This leads to many un- or ill-treated diseases such as diabetes. With amputation being cheaper than treating diabetic wounds, 8 out of 10 diabetics are mistreated, being subject to an amputation, resulting in major limb loss and accounting for the majority of amputees in Colombia (Joint Master Project, 2015).

Unlike in the UK, where the average age of lower limb amputees is approximately 70 years and 22% are over the age of 80 (Stewart, 2008), amputees in low-income countries tend to be much younger with an average age of 40-50 years (Joint Master Project, 2015). This makes it even more vital that they have access to a prosthetic device and can integrate back into society.

2.2 Prosthetic Industry

Following a lower limb amputation, it is vital that the amputee retains as much function of the residual limb as possible. Ensuring their joints do not cease up and maintaining muscle strength are key to a smoother transition to rehabilitation with a prosthetic device (Walsh & Walsh, 2003).

2.2.1 Western World

In the Western world (i.e. Western Europe and North America), the process for obtaining a prosthetic device is initiated by a physiotherapist. Firstly, the intentions and expectations of the amputee need to be discussed to determine what the patient would like to be able to achieve again and from this a suitable prosthetic device can be selected. The amputee will then receive their first liner, which is a soft cover that fits over the residual limb to protect it and that also creates a more comfortable fit with the socket that will ultimately sit around it. The patient is given a week to become accustomed to the feel of it and learn about caring for both the liner and the residual limb. When the residual limb appears at its thickest a plaster impression will be created with the use of strengthened plaster or plaster of Paris, which is moulded into shape by a trained prosthetist. This cast is then used to create a first socket within 2 weeks of taking the measurements. The amputee gets to test this socket and adjustments are made according to their feedback. Once the prosthetist and amputee are satisfied, a final socket is produced (Joint Master Project, 2015). A follow-up appointment is set 3 weeks after the patient receives their final socket, to allow them time to establish any problems that they may encounter with their prosthetic device which can then be discussed with the prosthetist and altered (CentersForMobility.com).

Prosthetists in the Western world manage to get a success rating of over 80% for creating a comfortable prosthesis that is accepted as a limb by trans-tibial amputees (Stewart, 2008). However, this is very difficult to achieve and requires the input of trained, experienced prosthetists that can identify how to alter and modify the prosthetic device to the patients' satisfaction.

2.2.2 Low-income Countries

The process of obtaining a lower limb prosthesis in developing countries is considerably more difficult. The amputees themselves have to initiate the process by visiting their local hospital. Depending on the exact location of their rural village, this can require travelling for long distances for the 24% of the population that live in the rural areas to the more central regions of such countries. For many of them this journey is unaffordable and the hope of obtaining a prosthetic device ends there. Even for those living in the urban regions, only few trained prosthetists can be found (Joint Master Project, 2015). The WHO reports that, in developing countries, less than 5% of those with disabilities have access to rehabilitation services (Walsh & Walsh, 2003).

The inability of the amputee to afford the travel expenses is only partly the reason why they are inaccessible for most. The health care system and health insurance companies also play a major role in limiting progress with excessive waiting times and insufficient funds that need to be distributed to the many people requiring attention and health care. In Colombia, the process for obtaining a prosthesis goes as follows: amputees are expected to travel to their local hospital after amputation which can then refer them to a physiatrist. This specialist can start rehabilitation with the amputee and has the authority to write a prescription with the need for a prosthetic device. It can take up to a month before the patient can attend an appointment with the physiatrist. The insurance company then has to approve and claim a prosthetic device for the patient, which can take up to 3 months to get accepted. This actually represents the best possible outcome, because more often than not the insurance company will decline the claim, even though they are legally obliged to cover prosthetic devices, due to insufficient funds. This leaves the option of paying for a prosthesis completely by the amputees themselves which is impossible for those with a low income. Many amputees will return to their homes with no prosthesis or likelihood of obtaining one, not realising that they can enforce their rights to obtain a prosthetic device through the Colombian Department of Justice. Those fortunate enough to understand this system will eventually receive their prosthesis, but they have more waiting time and travel expenses ahead of them. The health insurance company need to select a prosthetic lab for the production of their device, usually the one offering the best price. The amputee needs to wait for an appointment and is then expected to travel to the prosthetic laboratory in order for the prosthetist to make a mould of their residual limb. They then have to return to fit a test socket and finally to fit the final socket. Any further adjustments needed require further visits. If everything runs smoothly, the overall process would take approximately 4-5 months. This is a rare outcome, with amputees battling for up to 2 years before receiving their final prosthesis (Joint Master Project, 2015). These delays have obvious consequences for the amputee, needing to accept living without a prosthesis for longer. In addition, the amputee’s residual limb may undergo structural changes within this time to the extent that the prosthetic socket has to be re-made to accommodate these changes. This is especially true for younger amputees who are still growing.

Even once a prosthetic device has been obtained by the amputee, regular replacement or repair of parts is necessary, especially for those living and working in a rural environment. Such an environment tends to be harder on prosthetic devices, resulting in the need for more regular servicing and the associated travelling costs. It is not a surprise that many amputees scrape by or manage with a home-made prosthetic device.

2.2.3 Modern Lower Limb Prosthetic Devices

Lower limb prostheses are comprised of various components including the socket, the knee joint with adapters (for a trans-femoral a.k.a. above-knee amputee), the shin, the ankle joint with adapters and the foot. A cosmetic cover is usually used to encase all these components and produce a life-like look (figure 1) (OrthocareIndonesia.com).

Figure 1. *Image illustrating the components comprising an above-knee and below-knee prosthetic device. Both include the socket, the shin, the ankle joint with adapters and the foot with the addition of the knee joint and adapters for the above-knee prosthesis only. All these components can then be enclosed in a cosmetic cover (OrthocareIndonesia.com)*

There are many different types of lower limb prosthetic devices currently available to choose from in the Western world, each one composed of various combinations of different kinds of components to meet the individual needs of the amputee. The socket provides an interface with the end of the residual limb and allows for attachment of the prosthetic leg. For trans-tibial amputees, the patellar tendon bearing (PTB) socket used to be very popular when it was introduced around the 1960s (Radcliffe & Foort, 1961). It soon became clear that the high pressure experienced at the patellar tendon for long periods of time was causing problems and the attention turned to pressure casting which allowed an even pressure distribution across the residual limb. For trans-femoral amputees, the primary quadrilateral socket was altered in design to produce the ischial containment socket, also to achieve a more even pressure distribution and increase stability (figure 2) (Schuch, 1992).

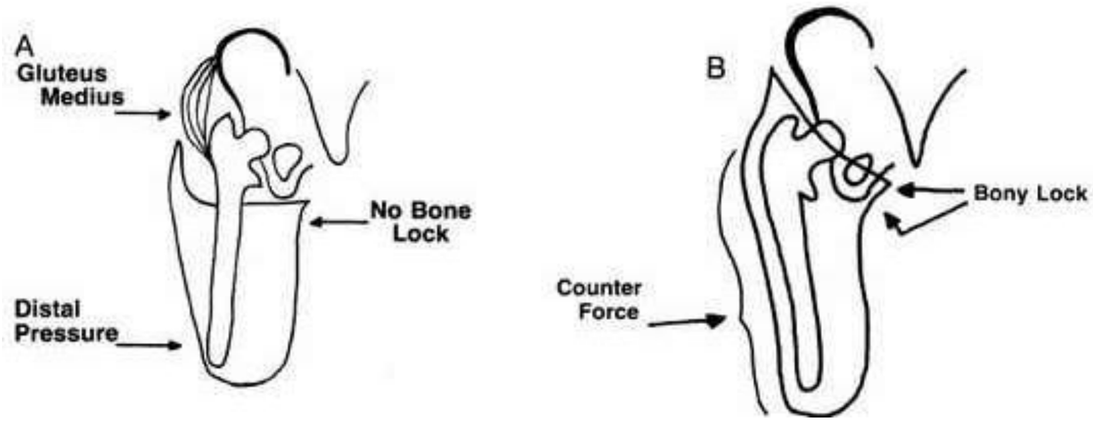


Figure 2. *Image illustrating the two trans-femoral sockets: the quadrilateral socket (A) and the ischial containment socket (B). The ischial containment socket has a more evenly distributed pressure along the whole length of the residual limb (as represented by the counter force), whereas the quadrilateral socket creates a distal pressure point. The ischial containment socket also provides more stability with containment of the ischial tuberosity (bony lock) compared to the quadrilateral socket which does not have a bony lock (Schuch, 1992).*

The knee mechanisms for trans-femoral amputees show the greatest range. They are available with a single axis or with a polycentric knee and with various stance phase and swing phase control mechanisms. The polycentric knee more accurately simulates the physiological knee than the single axis knee. However, it comes at the cost of being bulkier and thus heavier. Stance phase control can be achieved with simple alignment of the knee joint centre, with the incorporation of a polycentric knee or with a mechanical lock or weight activated brake. Swing phase control can either be free or achieved with a mechanical lock or with mechanical friction. More advanced mechanisms such as the use of fluid resistance (hydraulic or pneumatic) or a microprocessor can control both the stance and swing phase (Radcliffe, 1977).

The whole prosthetic device also needs to be suspended from the amputee, which can be achieved with the use of a simple belt strap, suction, specialised liners or, in the more advanced designs, there is osseo-integration, where the prosthetic device is implanted into the end of the residual limb. The decision on the method of suspension is a personal choice made by the amputee.

Examples of modern lower limb prosthetic devices available in the Western world include the C-leg, which was introduced in 1977 by Ottobock. It was the first to encompass a micro-processor controlled hydraulic system to automatically adapt to the individual’s walking speed and slope (Ottobock.co.uk). Endolite’s ESK is a single axis knee joint with weight activated stance control and pneumatic swing control which can be adjusted in order to suit the activities of the individual (Endolite.co.uk). Technology is constantly advancing, as seen with Össur’s power knee which is the first prosthetic knee to incorporate a motor-powered knee in order to reduce the effort of walking (ossur.co.uk).

It is of great importance to keep energy expenditure to a minimum as amputees require more energy during gait than non-amputees; the higher the level of amputation, the more energy that is required. It is seen that trans-tibial amputees require 30% more energy than a non-amputee, whereas trans-femoral amputees require up to 70% more energy (Stewart, 2008). The use of lightweight material or incorporation of a motor can aid in keeping this excess energy expenditure to a minimum.

Unfortunately, all of these state-of-the-art prosthetic devices come with a big price tag. As such, even though reducing the cost of prosthetic devices is especially beneficial for low-income countries, middle- and high-income countries will also benefit.

2.3 Cost Reduction

It is clear that obtaining and maintaining a prosthetic device does not come cheap. The problem is especially acute in low-income countries where the average annual salary is €300 and a prosthesis costs ~€1,400. With the inclusion of the travel and maintenance costs, it can take victims a decade before they can afford their initial prosthetic device (Strait, 2006). There are many different aspects of the process for obtaining a prosthesis or the production of the device itself which can be addressed in order to reduce the cost to a certain extent and make them more affordable and accessible, especially for those in low-income countries.

2.3.1 Prosthetic Components

The manufacturing of the components making up a lower limb prosthesis can be modified in order to produce them in the most cost-effective manner, either in terms of the material or method of production.

Research into a cheaper method for producing prosthetic sockets has resulted in the Majicast (figure 3), which is a 'hands-off' hydro-casting technique. The residual limb is inserted into a water filled casting device and the soft tissues are loaded with a uniform pressure by the water and the Chinese cuff device. The subject's own weight induces this pressure and is similar to loading a prosthesis during gait. This method ensures the stiffest coupling is achieved between the residual limb and the socket, maximising stability whilst minimising shear stresses. This can produce a suitable socket first time around, eliminating the need for fine tuning, adjustments and a test socket.

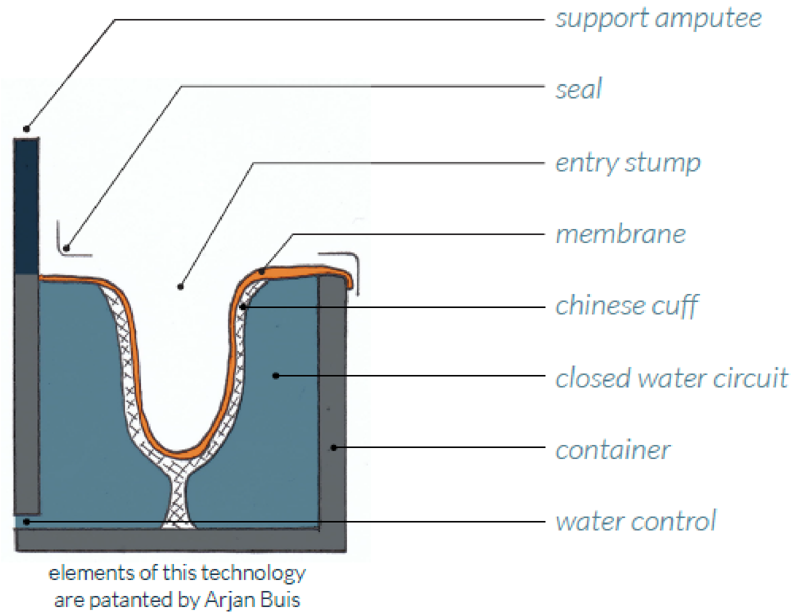


Figure 3. *Schematic representation of the Majicast, which allows a cheaper method of production for lower limb prosthetic sockets. The use of water and the Chinese cuff device ensures a more uniform pressure distribution upon casting, creating a more suitable socket first time around and discarding the need for adjustments and a test socket (Joint Master Project, 2015).*

The cost of material and time are saved as follow-up appointments are not required, allowing more time for the prosthetist to assist other patients. In addition, it significantly reduces travel costs for those patients from remote areas having to travel to the larger cities where qualified prosthetists are primarily based as fewer appointments are required. Customised sockets can also be produced more quickly compared to the 'hands-on' casting technique, once more allowing the prosthetist's time to be used more effectively, as well as decreasing the waiting time for patients. Another great potential of the Majicast is the production of direct sockets, bypassing the mould phase completely and creating the final socket immediately: speeding up the socket production even more. Consistent, replicable casts can be produced with the Majicast potentially discarding the need for a highly experienced prosthetist altogether. General hospital staff in rural hospitals could be easily trained to use the Majicast, making it possible for the sockets to be produced *in situ* and avoiding the need for a lengthy journey to the city. The Majicast has been shown to produce affordable yet high quality sockets that have a long life span with great potential in not only low-income but also high-income countries (Joint Master Project, 2015).

The foot is another component on which the cost could be reduced. Numerous prosthetic feet that can be produced cheaply have been designed and used. First designed in 1968 in India, the Jaipur foot has seen many refinements over the years which have led to its continuing popularity. It is manufactured from pieces of wood and rubber, covered in vulcanised rubber (figure 4, left). It is lightweight, durable and waterproof. In addition, its flexibility and life-like look makes it suitable for the cultural requirements of low-income countries. It is robust enough to not require any shoes, which suits the Indian culture as most people go around bare-foot, in addition to the prohibition of wearing shoes in places of worship such as mosques and temples. The International Committee of the Red Cross also found that this particular foot could endure use in the mountainous regions (Sadikot *et al.*, 2007).

All of this at a cost of approximately €45 per foot makes it ideal for use in low-income countries (Joint Master Project, 2015).

Figure 4. *(Left) Image of the Jaipur foot illustrating the life-like look and a cross-section revealing the materials used for its manufacture (Cosgrove, 2013). (Right) Image of the Niagara foot manufactured entirely from Delrin and designed to provide a degree of energy return during gait (Suherman, 2012).*

Jensen & Treichl (2007) mechanically tested 21 currently available prosthetic feet in low-income countries, including the Jaipur foot. Although they are extensively used, none of them fully comply with the European and British testing standards of BS EN ISO 10328:2006, which specifically assess the ultimate strength and durability of the prosthetic device.

The Niagara foot is another low-cost foot designed for those walking on rugged grounds with an active life-style. The claims are that it is compliant with ISO 10328 standards, in terms of strength and durability. It is made entirely out of Delrin, a synthetic polymer, which is impact-resistant, and designed in such a way as to provide some energy return during gait (figure 4, right). It is lightweight and available at an affordable price of €35 (niagarafoot.com).

2.3.2 Organisations for Delivering Prosthetic Devices

The production of affordable prosthetic devices is a redundant activity if they cannot be made accessible to those in need. A number of organisations have been set up to deliver affordable prosthetic devices for those in low-income countries. The International Committee of the Red Cross (ICRC), for example, sponsor limb-fitting centres in landmine affected countries. They promote the production of unique prosthetics manufactured from polypropylene which is lightweight, easy to replace or repair and recyclable. Local technicians are trained in the use of this material to ensure the continuation of this program once it is transferred on to local authorities (International Committee of the Red Cross, 2007).

The LegBank is a newly set-up organisation aiming to not only locally produce affordable, high-quality and individually customised prostheses, but also aims to provide a local financial system which would give out loans to those in need and thus making a prosthetic device available to everyone who requires one in Colombia (Joint Master Project, 2015).

Exceed, formerly known as The Cambodia Trust, work to tackle the problem of a lack of trained prosthetists and orthotists in developing countries. They have established schools in five different countries in Southeast Asia which are accredited to International Society for Prosthetics and Orthotics (ISPO) Category 2 standards and that can train local prosthetists and orthotists (Exceed-worldwide.org). The aim is to reduce waiting times for patients as more specialists will be available. Furthermore, the specialist will be more evenly distributed across a country, reducing the extensive travel to the bigger cities where specialists currently tend to be situated.

2.4 Materials for Lower Limb Prosthetic Manufacture

Carbon fibre is a state-of-the-art material providing the ideal characteristics; it is lightweight, stiff and durable. Various modern lower limb prosthetics are manufactured from carbon fibre such as Össur's Vari-Flex prosthetic foot and Ohio's Willow Wood's Carbon Copy II prosthetic foot. As this is one of the most expensive materials that can be used, many other materials are used or considered in order to keep the cost of the prosthetic device to a minimum. These include glass fibre and synthetic fibres (e.g. Kevlar rubber). Glass fibre is stronger than carbon fibre with a failure stress of 3,400 MPa compared to 2,600 MPa for carbon fibre, however it is not as stiff as carbon fibre, with a Young's modulus of only 70 GPa (glass fibre) compared to 210 GPa for the carbon fibre (Klasson, 1995). Glass fibre is cheaper than carbon fibre and ideal as reinforcement as used in the new generation of the SACH foot (Ottobock.co.uk).

More recently, attention has been turned to the potential utilisation of natural fibre composites for prosthetic manufacture as these can offer an even lower-cost alternative to carbon fibre. Natural fibre based composites are not only much cheaper, but are also renewable and sustainable and tend to be found locally. This reduces the cost of producing prosthetic devices from natural fibre based composites as the need for importing materials will be greatly reduced or negated. Their biodegradability also brings with it environmental benefits (Saheb & Jog, 1999).

Natural fibres are already extensively used for reinforcing polymers in load bearing applications. This is due to the high tensile properties seen in natural fibres such as bamboo, jute and pineapple with tensile strengths of approximately 200 MPa, 393 MPa and 170 MPa respectively (Deshpande *et al.*, 2000; Jain *et al.*, 1992). Bamboo grows very fast and easily on ground that other plants find non-viable and is therefore a possible ideal potential resource for prosthetic manufacture in low-income countries.

It is important to note that the orientation and processing of the materials can drastically change their mechanical properties. Jain *et al.* measured a tensile strength of only 8.6 MPa for bamboo fibres cross-sectionally (i.e. across the fibres), compared to the 200.5 MPa when measured longitudinally (i.e. along the fibres). Arranging the fibres in various directions would provide the best overall strength. In addition, Chand & Tamrakar (2015) found that a tensile strength of 730.2 MPa could be reached for bamboo fibres if they were extracted after dipping the fibres in alkali and then steam blasting them for 30 minutes.

Many different materials are being used for the manufacture of prosthetic devices, including carbon, glass and synthetic fibres. Natural fibres have the potential to offer a cheap alternative for prosthetic manufacture. However, it is important to understand that for any material to accommodate the desired characteristics for prosthetic manufacture, the ultimate key to success is its design.

Chapter 3: Prosthetic Design

The ultimate aim of a prosthetic foot is to reproduce all the functions of a normal human foot in order to allow a normal gait. As the human foot has an extensive list of capabilities and many degrees of freedom, this is virtually impossible. The solid ankle cushion heel (SACH) foot has been the conventional prosthetic foot since its introduction in 1958. This simple and reliable prosthetic foot comes with many limitations, mainly the inability to actively contribute throughout stance, making them very energy inefficient. To overcome this limitation, the attention turned to improving the design of the SACH foot in order to lower the energy cost of ambulation for the amputee, thus providing better performance in enabling a more natural gait pattern. This gave rise to energy storing dynamic feet (van Drongelen, 2000).

3.1 Energy Storing Dynamic Feet

Energy storing dynamic feet behave, as the name suggests, by storing energy. More importantly however is the fact that this stored energy is returned at a key point in the gait cycle, namely at toe push-off. As an individual loads the energy storing dynamic foot (also known as dynamic elastic response (DER) foot) in early and mid-stance, the foot will store a certain amount of energy by deflection. During late stance, as the load is being reduced, this energy is returned in a propulsive manner by the foot returning to its original form (Geil, 2000). This is unlike earlier designs of prosthetic feet, such as the SACH feet, where the energy is simply dissipated through the material, usually with the aid of a shock absorber such as the cushioned heel (van Drongelen, 2000).

A normal human foot relies on the plantar-flexors (i.e. the calf muscles) for generating 80% of the total energy utilised during gait (Winter & Sienko, 1988). Energy is stored during mid-stance by eccentric contraction of the plantar-flexors, controlling the forward rotation of the tibia over the supporting foot. This energy is then released in late stance by concentric contraction of the plantar-flexors, generating the force required for push-off and propelling forward (Postema *et al.*, 1997).

DER prosthetic feet have different mechanisms for imitating the plantar-flexors' ability to store and release energy. The first effective energy storing dynamic foot was the Seattle foot. Introduced in 1985 by Burgess *et al.* (1985), it consists of a short canti-leaver spring (the keel), that is fabricated from Delrin, and a flexible toe pad which is contained in polyurethane foam (figure 5 left) (Michael, 1987). Another well-known prosthetic foot is the Carbon Copy II, which has a similar design to the Seattle foot with the addition of offering two-staged resistance at heel-off. At normal walking speeds, the primary deflection plate provides a small degree of energy return, whereas at higher cadence the auxiliary deflection plate will act to provide a bigger degree of energy return (figure 5 right) (van Drongelen, 2000).

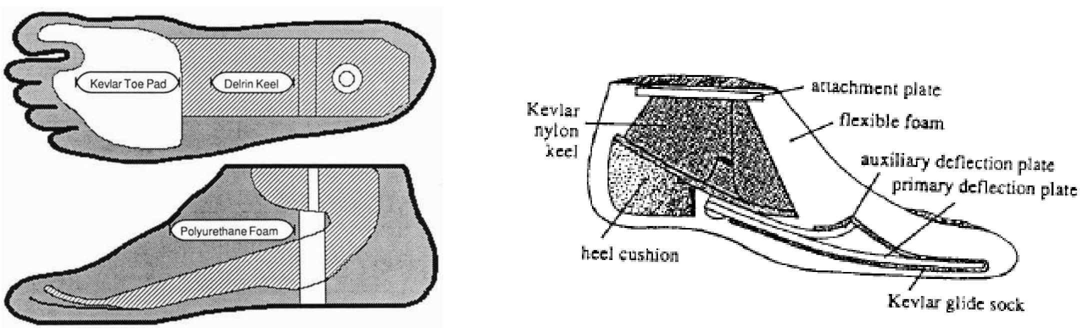


Figure 5. *(Left) The Seattle foot consisting of a Delrin keel that acts as a cantilever spring for storing and releasing energy and a flexible Kevlar toe pad contained in polyurethane foam (Michael, 1987). (Right) The Carbon Copy II prosthetic foot with two deflection plates that allow for a two-staged resistance at heel-off (van Drongelen, 2000).*

The Flex foot is of a different design, where the entire prosthetic component distal to the socket contributes to energy storage and return. This provides a much greater energy return, 66% for the Flex foot compared to 21% energy returned by the SACH foot (Schneider *et al.*, 1993). In addition, instead of a shock absorbing heel seen with the Seattle foot and Carbon Copy II, the heel is also designed to act as a leaf spring, providing a smoother shock absorber on heel strike and aiding energy storage (van Drongelen, 2000). Durability of the prosthetic foot is increased as the heel lever is bolted to the forefoot, relieving some of the stress experienced at the ankle, where the heel cushions are usually fixed (Michael, 1987). Figure 6 left shows Össur's Vari-Flex Modular prosthetic foot with the long prosthetic tibia contributing to its energy efficiency. This particular model claims to provide an energy return of 95%. In addition, the toe and heel are split in the sagittal plane to allow a certain degree of inversion and eversion of the foot to accommodate walking on uneven ground whilst retaining stability.

Figure 6. *(Left) Össur's Vari-Flex Modular with a long prosthetic tibia which acts to aid energy storage and return. (Right) Össur's Vari-Flex prosthetic foot which has the same design as the Vari-Flex Modular, however does not possess the prosthetic tibia and is therefore slightly less energy efficient (Össur.co.uk).*

Flex feet also come with a shorter prosthetic shank such as Össur's Vari-Flex (figure 6, right) (Össur.co.uk). This prosthetic foot consists of the same design as the Vari-Flex Modular, but without the prosthetic shank. This makes it suitable for those amputees with a longer residual limb or children, but compromises slightly on the energy efficiency (van Drongelen, 2000).

The Vari-Flex Modular and Vari-Flex prosthetic devices are constructed entirely from carbon fibre, as are the Carbon Copy II deflection plates described above. Carbon fibre is an ideal material for prosthetic device manufacture as it is lightweight, durable and stiff. Unfortunately, this increases the cost of the prosthetic devices.

Carbon fibre can be orientated in two different ways, unidirectional or bidirectional, establishing different properties for the same material. Unidirectional orientation is when the carbon fibres are all aligned parallel to one another. This gives maximum strength along the length of the fibres and also allows bending perpendicular to the fibre orientation. Bidirectional orientation is when the carbon fibres do not run parallel to one another. They are woven at 90° to each other. This gives maximum strength in varying directions.

The outer layer of Össur's Vari-Flex prosthetic foot has the carbon fibre at a bidirectional orientation of 45° to the line of the foot. This provides maximum strength against torsion. It is expected that the carbon fibre is unidirectionally orientated underneath this layer to also provide the strength it requires along the length of the foot.

3.1.1 Energy Efficiency

The energy efficiency of prosthetic feet can be determined in two main different ways: clinically or mechanically.

Biomechanical gait analysis with the use of force plate data and a motion capture system of an amputee wearing a specific prosthetic device will allow the generation of a power-time graph at the ankle (figure 7). The area under this graph represents the energy stored and released. For accurate energy analysis, the heel and the keel are to be considered separate as they both store and release energy in different manners. In figure 7, area A represents the energy stored by the heel, area B represents the energy returned by the heel, area C is the energy stored by the keel and area D is the energy returned by the keel (Hafner *et al.*, 2002).

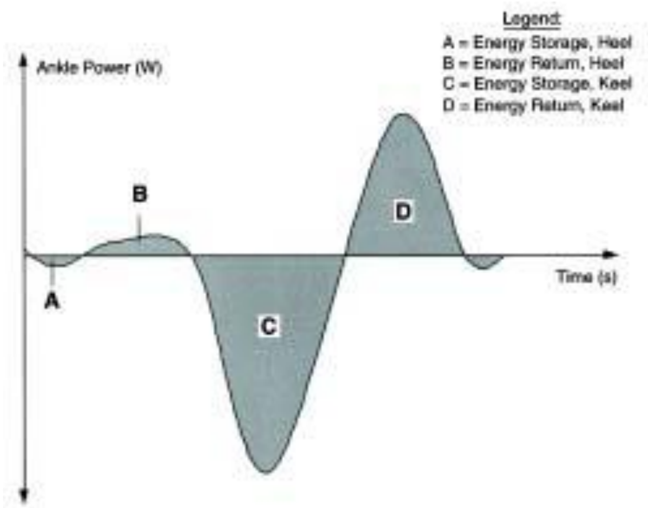


Figure 7. *Power (W)–time (s) graph at the ankle that can be generated with the use of force plate and motion capture data. Area A and C represent the energy stored by the heel and keel respectively; area B and D represent the energy released by the heel and keel respectively (Hafner et al., 2002).*

With the values for the energy stored and released, the dissipated energy and energy efficiency can be calculated as follows:

$$Energy\ dissipated = Energy\ stored - Energy\ returned \quad (1)$$

$$Energy\ efficiency = \frac{Energy\ returned}{Energy\ stored} \quad (2)$$

Mechanically determining the energy efficiency of a prosthetic device requires the use of an Instron machine for applying loads on it. A graph of load against displacement can be generated and will be expected to look similar to the graph seen in figure 8.

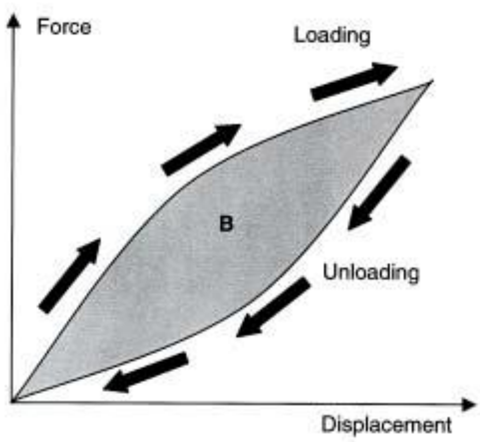


Figure 8. *Graph of load against displacement that can be expected when loading and unloading a prosthetic device with an Instron machine. 'B' represents the energy dissipated by the prosthetic foot (Hafner et al., 2002).*

In this case, the energy stored by the prosthetic device is the area under the curve upon loading. The area under the curve on unloading is the energy released. Using equation 1, this leaves area B in figure 8 the energy dissipated by the prosthetic foot. This method is less accurate than the clinical method described above as the heel and the keel are not considered separately, but rather the entire prosthetic device is considered as a whole (Hafner *et al.*, 2002).

3.2 Natural Fibre Prosthetic Foot

The natural fibre based prosthetic foot produced for the experimental testing of this study is shown in figure 9 (left). It was constructed from birch veneer. This material was acquired from an IKEA POÄNG armchair where the birch veneer is layered and glued together. It is thought that the fibre orientation of the birch is at 90° for each consecutive layer to provide more strength in varying directions compared to if they were aligned only in parallel, however this has not been verified.

Figure 9. *Image of the natural fibre based composite prosthetic foot prototype (left) and Össur's carbon fibre Vari-Flex prosthetic foot (right). The natural fibre prosthetic foot was constructed using a similar design to that of the carbon fibre type.*

The complete design of the natural fibre prosthetic foot was based on that of Össur's Vari-Flex (figure 9, right). The curve in the frame of the armchair was used to act as the transition from the shank to the foot. This curve is also seen in the Vari-Flex prosthetic foot design and is inspired by nature, known as the Baud curve. Baud curves reduce the stresses experienced at that point, such as those seen at the point where branches of a tree meet the main tree trunk. This is a very beneficial characteristic for the ankle joint in a prosthetic device where the stresses experienced can be very high.

The thickness of the natural fibre material was much greater than that of the Vari-Flex, with a thickness of approximately 17.0 mm at the shank, 15.3 mm in the Baud curve, 15.0 mm at the forefoot region and gradually decreasing to 6.6 mm at the toe region. The Vari-Flex had an approximate thickness of 9.7 mm at the shank, 8.6 mm in the Baud curve and 6.2 mm at the forefoot region, gradually decreasing to 3.0 mm at the toe region. The heel of the natural fibre prosthetic foot was thickest at the most posterior end at 9.6 mm whereas it was the thinnest part of the heel of the Vari-Flex at only 3.2 mm. For the Vari-Flex, the heel gradually increased in thickness to 8.5 mm at the point where the heel was connected to the forefoot. At this joint in the natural fibre prosthetic foot, the heel was slightly filed to a thickness of 8.0 mm. Some layers were filed off at the Baud curve to ensure enough flexibility for walking. The forefoot was split in the sagittal plane, similar to that seen with the Vari-Flex and the heel was produced to consist of two components that were bolted to the forefoot. The entire natural fibre foot was then lacquered in an attempt to decrease surface tension and prevent the veneer layers detaching from each other and effectively tearing through the material. Rubber pads were attached to the toe and heel region.

Chapter 4: Materials and Methodology

4.1 Prosthetic Feet

The natural fibre based prosthetic foot was manufactured as described above (Chapter 3) at the University of Strathclyde by Dr Arjan Buis. The carbon fibre prosthetic foot, used for comparison, was the Vari-Flex foot (category 7; size 26), purchased from Össur.

4.2 BS EN ISO 10328:2006

BS EN ISO 10328:2006 outlines the structural testing procedures and requirements of lower limb prosthetic components, according to British and European standards.

4.2.1 Coordinate system

The coordinate system followed by BS EN ISO 10328:2006 is shown in figure 10.

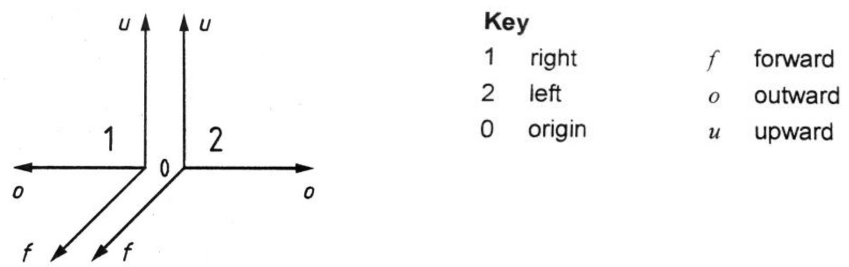


Figure 10. *Coordinate system for right- (1) and left-sided (2) application and key as taken from BS EN ISO 10328:2006. u, o and f are the axes that run from the origin of the coordinate system in the proximal, lateral and anterior direction respectively.*

The *u*-axis runs from the origin 0 of the coordinate system through the effective ankle and knee joint centres in the proximal direction.

The *o*-axis runs from the origin 0 of the coordinate system perpendicular to the *u*-axis and parallel to the effective ankle and knee joint centrelines in the lateral direction.

The *f*-axis runs from the origin 0 of the coordinate system perpendicular to the *u*-axis and the *o*-axis in the anterior direction.

This coordinate system can be utilised for right- (1) and left-sided (2) application.

4.2.2 Test Loading Levels

Due to differences in an individual's physical parameters and the biomechanical gait of amputees, prosthetic devices are categorised to accommodate the individual's need. The categories range from P3 to P6 based on the increasing loads which the prosthetic device must be able to withstand. The values of these loads for various tests are specified in BS EN ISO 10328:2006. In this study, tests were only performed up to test loading level P5, as a lower limb prosthesis able to sustain loads up to this level was deemed sufficient.

4.2.3 Prosthetic Foot Alignment

Testing of the prosthetic foot was carried out in two parts: forefoot testing and heel testing. The loads applied to either the forefoot or heel were set at specific angles in order to simulate an individual's behaviour at specific points during the gait; namely heel strike and toe-off. For separate tests on ankle-foot devices and foot units, BS EN ISO 10328:2006 specifies that forces applied to the heel (F_1) are to be at an angle of 15° (α) to the *u*-axis and forces applied to the forefoot (F_2) are to be at an angle of 20° (β) to the *u*-axis (figure 11).

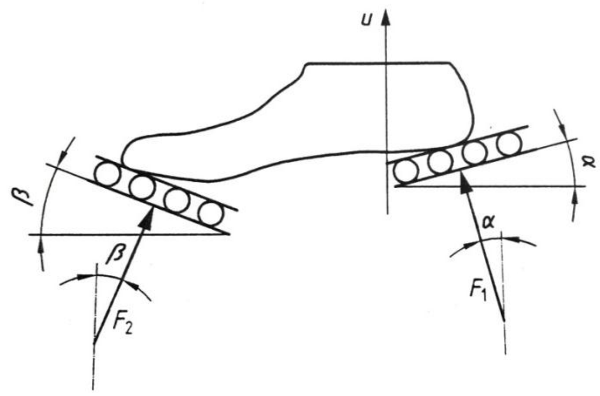


Figure 11. *Schematic illustrating the angles of the forces to be applied to the prosthetic foot according to BS EN ISO 10328:2006, where α = 15°, β = 20°, F_1 = the force applied to the heel and F_2 = the force applied to the forefoot (BS EN ISO 10328:2006).*

4.3 Mechanical Testing

Mechanical testing was performed according to the International Test Standard, BS EN ISO 10328:2006, by means of an Instron ElectroPuls E10000 (figure 12). The prosthetic foot undergoing testing was attached to a metal test rig with screws and mounted in the Instron. Tests were carried out at increasing test loading levels from P3 to P5. Due to time limitations, only the static proof test was completed, but not the ultimate strength test or the cyclic fatigue test.

Figure 12. *The Instron ElectroPuls E10000 utilised for the mechanical testing of the natural fibre based prosthetic foot prototype and Össur's carbon fibre Vari-Flex prosthetic foot (Instron.co.uk).*

4.3.1 Static Proof Test

The natural fibre based composite prosthetic foot was mounted in the Instron and aligned appropriately for forefoot testing (figure 13).

Figure 13. *Image of the set-up for forefoot testing of the natural fibre prosthetic foot according to BS EN ISO 10328:2006. The prosthetic foot was attached to a metal test rig and clamped in the Instron with the desired angle of 20°.*

A minimum test force of 50 N was applied at a rate of 10 N/s and increased at a rate of 100 N/s to the proof test force for test loading level P3. This compressive force was maintained for 30 seconds and released to return to 0 N at the rate of 100 N/s. Following visual inspection to ensure that the prosthetic foot had not undergone any major damage, the above procedure was repeated for test loading levels P4 and P5. The prosthetic foot was then re-aligned in the Instron to obtain the desired angle of 15° for heel testing, which was carried out exactly the same as described for the forefoot testing. This was repeated on 3 different occasions, with 6 or 7 days in between each occasion, to allow time for the prosthetic foot to return to its former state in case deformation occurred. The carbon fibre prosthetic foot was also tested as described above for comparison.

The values of the proof test force for forefoot and heel testing at test loading levels P3 to P5 as specified in BS EN ISO 10328:2006 are presented in table 1. Note that the values for forefoot loading and heel loading are the same within a specific test loading level.

		Test loading level		
		P3	**P4**	**P5**
Proof test force (N)	**Forefoot loading**	1,610	2,065	2,240
	Heel loading	1,610	2,065	2,240

Table 1. *Proof test force values for forefoot and heel loading at test loading levels P3, P4 and P5 as specified in BS EN ISO 10328:2006.*

Using Microsoft Excel 2010 software, graphs were produced, for both prosthetic feet, of displacement against time to determine the differences in stiffness between the natural fibre based composite prosthetic foot and the carbon fibre prosthetic foot. Graphs of load against displacement were also constructed to illustrate the energy efficiency between the two different prosthetic feet.

4.3.2 Deformation Tracking

Coloured stickers were attached to the side of both the natural fibre based composite prosthetic foot (red and blue) and the carbon fibre prosthetic foot (yellow and green) at 20 mm intervals as shown in figure 14. A camera (from a Samsung Galaxy Note 10.1) was set up to record the deformation of the prosthetic foot throughout one static proof test for test loading levels P3 to P5 at 60 frames per second.

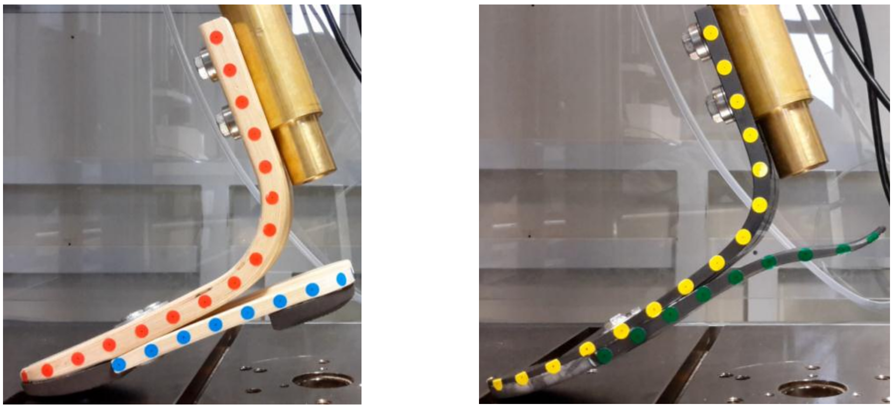


Figure 14. *Picture illustrating the markers attached to the side of the natural fibre based composite prosthetic foot (left) and the carbon fibre prosthetic foot (right) at 20 mm intervals.*

Adobe Photoshop CS4 software was used to mark the position of the markers on one in every 60 frames (i.e. one frame every second) during the loading period only. A line was drawn through the markers on each analysed frame and these marked frames were then superimposed to allow clear visualisation of the deformation of the prosthetic foot as it was undergoing loading.

The amount of deformation was also quantified by measuring the distance travelled by a specific marker from the first frame to the last frame of the loading period. The same marker was chosen for analysis for all test loading levels and for both the natural fibre prosthetic foot and carbon fibre prosthetic foot to allow for accurate comparison. A marker in the curve of the prosthetic foot was chosen as this showed the greatest differences.

Chapter 5: Results

5.1 Static Proof Test Results

The graphs presented in this section were chosen to illustrate key observations. For completeness, the rest of the graphs produced from the data obtained are presented in appendix A.

5.1.1 Comparison of Natural Fibre with Carbon Fibre

The static proof test results for test loading level P5, test 2, on the carbon fibre and natural fibre forefoot and heel are shown in the figures 15 and 16 respectively.

The graphs shown in figure 15 clearly illustrate that the natural fibre prosthetic forefoot was much stiffer than the carbon fibre alternative, reaching only a maximum of -19.07 mm compared to the carbon fibre foot reaching a maximum of -36.79 mm displacement. The energy dissipated by the natural fibre prosthetic forefoot, as determined by the area bounded by the curves, was however similar to that dissipated by the carbon fibre prosthetic forefoot. The carbon fibre forefoot showed a more linear relationship between the load and displacement compared to the natural fibre heel, which displaced less at higher loads.

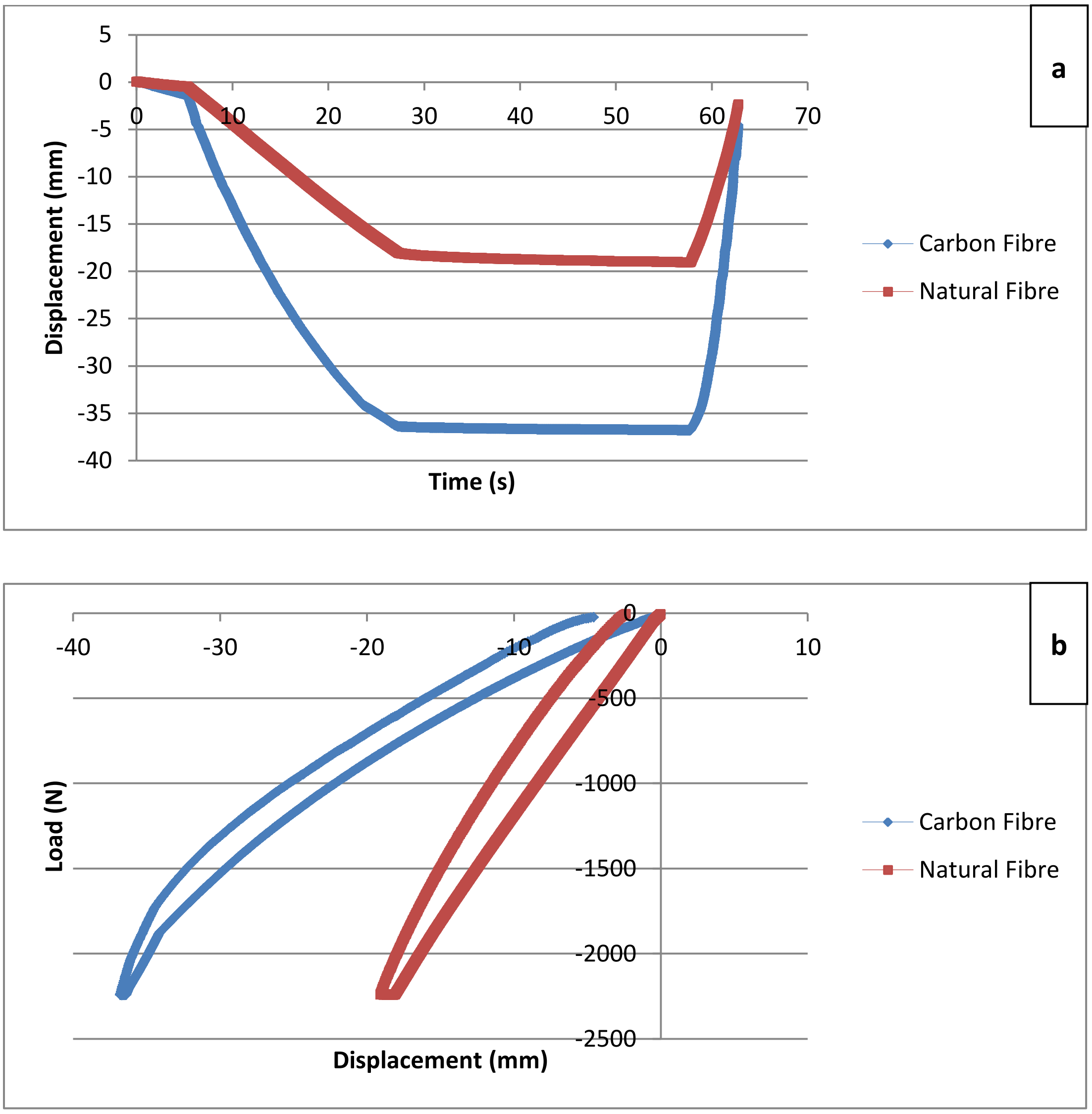


Figure 15. *Comparison of the displacement (a) and energy efficiency (b) of the forefoot of the carbon fibre prosthetic foot and natural fibre prosthetic foot for test loading level P5, test 2. The carbon fibre prosthetic foot showed a much greater displacement with time and therefore also a much greater flexibility. The energy efficiency was only slightly less for the natural fibre prosthetic foot. A more linear relationship was seen for the carbon fibre forefoot between the load and displacement compared to the natural fibre forefoot, which displaced relatively less at higher loads.*

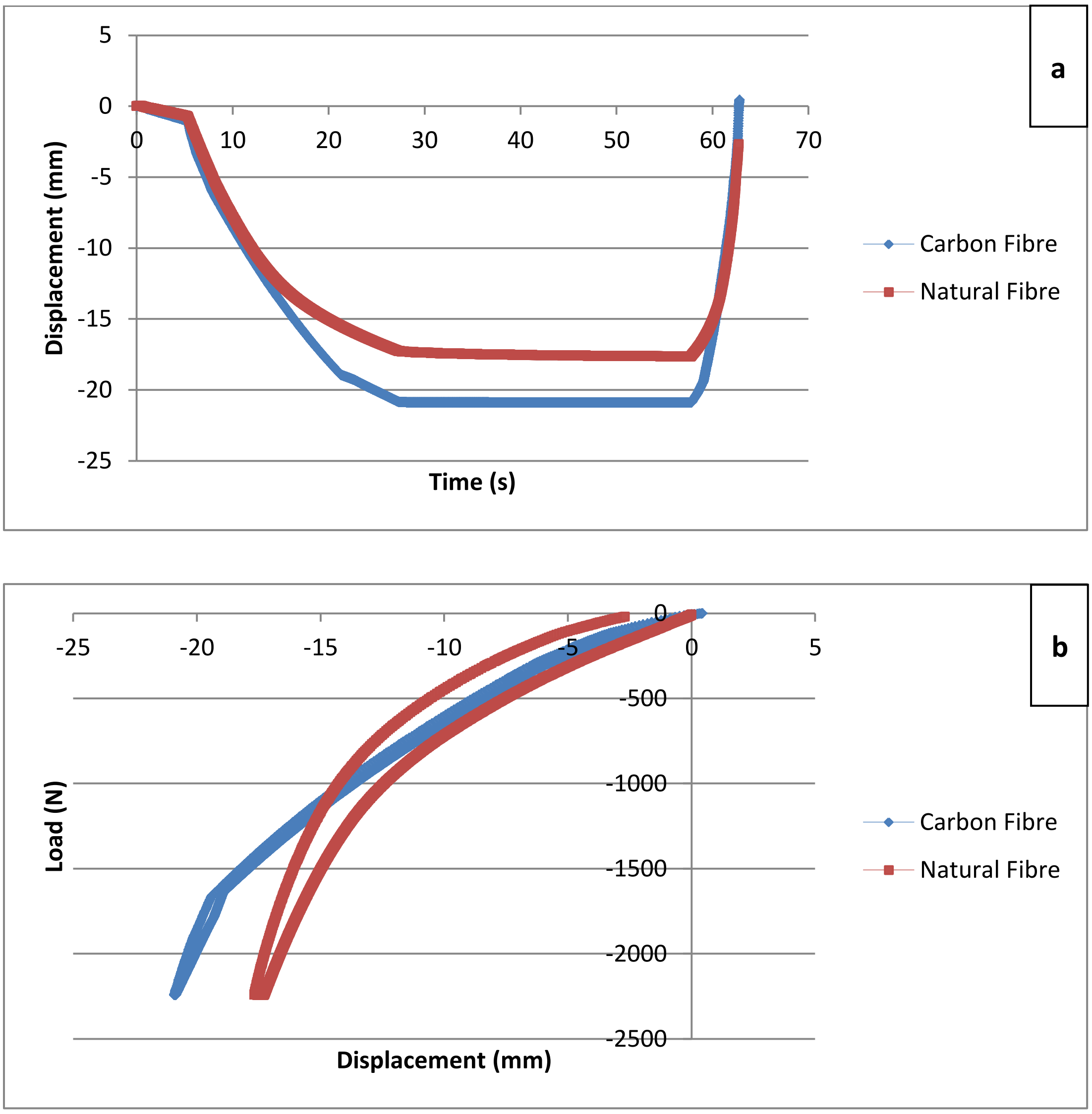


Figure 16. *Comparison of the displacement (a) and energy efficiency (b) of the heel of the carbon fibre prosthetic foot and natural fibre prosthetic foot for test loading level P5, test 2. The carbon fibre prosthetic foot showed a slightly greater displacement with time. The flexibility between the two prosthetic feet was comparable, where the natural fibre prosthetic heel became stiffer at higher loads. The energy efficiency was much less for the natural fibre prosthetic foot. The natural fibre heel displaced less at higher loads. The carbon fibre heel had a more linear relationship between the load and displacement until approximately 1,600 N.*

When comparing the two different heels as shown in figure 16, the natural fibre prosthetic heel displaced by a maximum of -17.67 mm and the carbon fibre foot displaced by a maximum of -20.9 mm. The natural fibre is stiffer than the carbon fibre, however to a lesser extent than for the forefoot. Energy dissipated by the heels did show a greater difference between the natural fibre and carbon fibre material, with much more energy being lost by the natural fibre prosthetic heel. Such as that seen with the forefeet in figure 15, the natural fibre heel displaced relatively less at higher loads. The carbon fibre heel showed a slightly more linear relationship, however after reaching approximately 1,600 N the relationship changed to a similar relationship as the natural fibre heel.

The trends between the natural fibre and carbon fibre prosthetic foot described above were similar for test loading levels P3 and P4 and for tests 1 and 2. One exception was the displacement of the heel for test 1 at test loading levels P3, P4 and P5 where the natural fibre prosthetic heel displaced more than the carbon fibre heel. For test loading level P5, the natural fibre prosthetic heel displaced to -20.16 mm compared to the carbon fibre heel only displacing to a maximum of -18.06 mm (figure 17). The carbon fibre heel showed a linear relationship between load and displacement, where no change in the relationship was seen once a certain load had been reached, such as that observed in figure 16. The natural fibre heel displaced less at higher loads as seen before (figure 16).

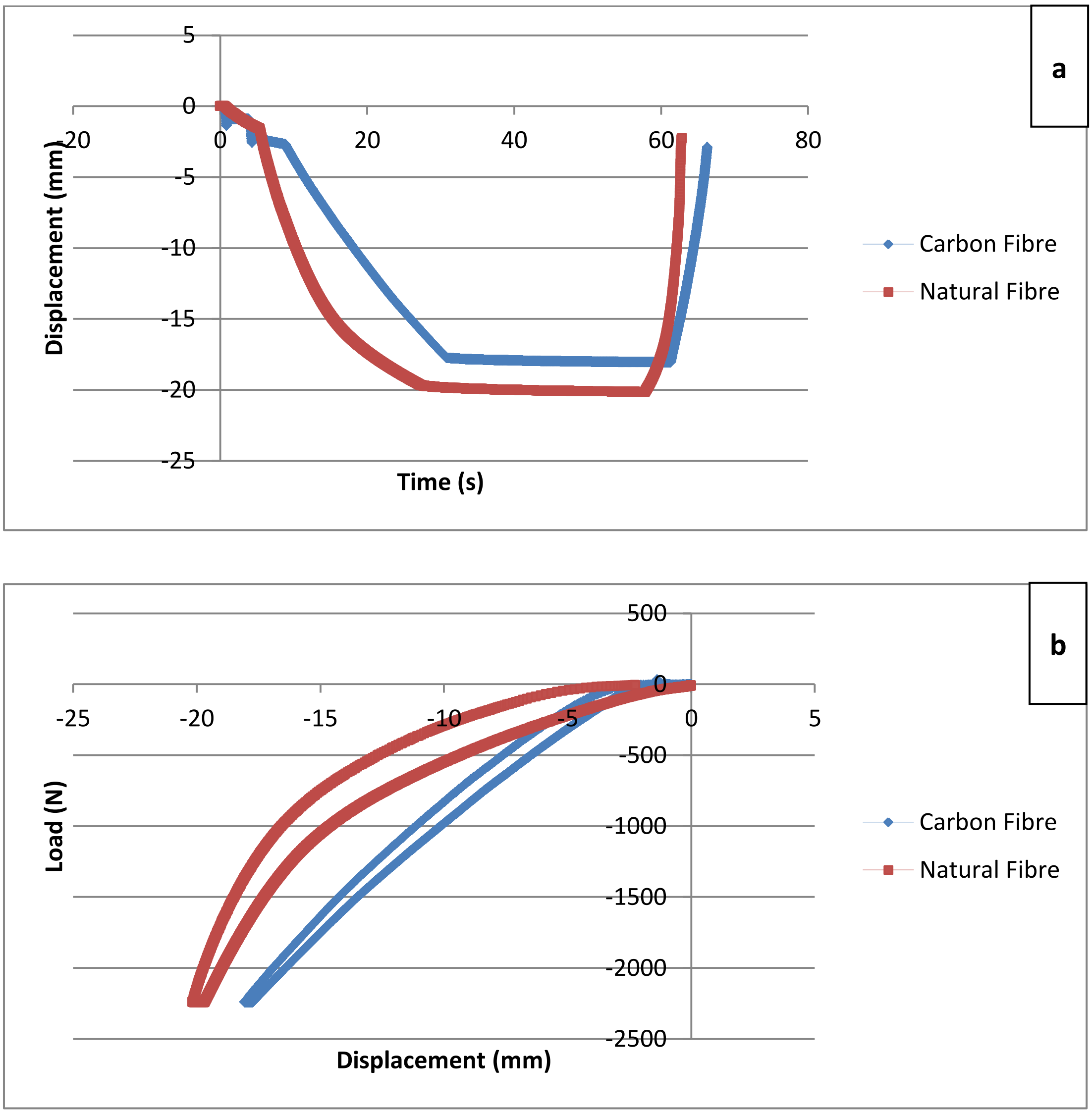


Figure 17. *Comparison of the displacement (a) and energy efficiency (b) of the heel of the carbon fibre prosthetic foot and natural fibre prosthetic foot for test loading level P5, test 1. The natural fibre prosthetic foot showed a greater displacement with time. The carbon fibre heel displaced linearly with increasing load, whereas the natural fibre heel displaced less at higher loads.*

5.1.2 Carbon Fibre Results

The maximum displacement of the carbon fibre prosthetic foot gradually increased for both the forefoot and heel as the test loading levels progressed from P3 to P5. Figure 18 illustrates the displacement with time for forefoot loading of the carbon fibre prosthetic foot during test 2 for the three different test loading levels. The carbon fibre prosthetic forefoot reached a maximum of -32.78 mm displacement for test loading level P3, -35.43 mm for test loading level P4 and -36.79 mm for test loading level P5.

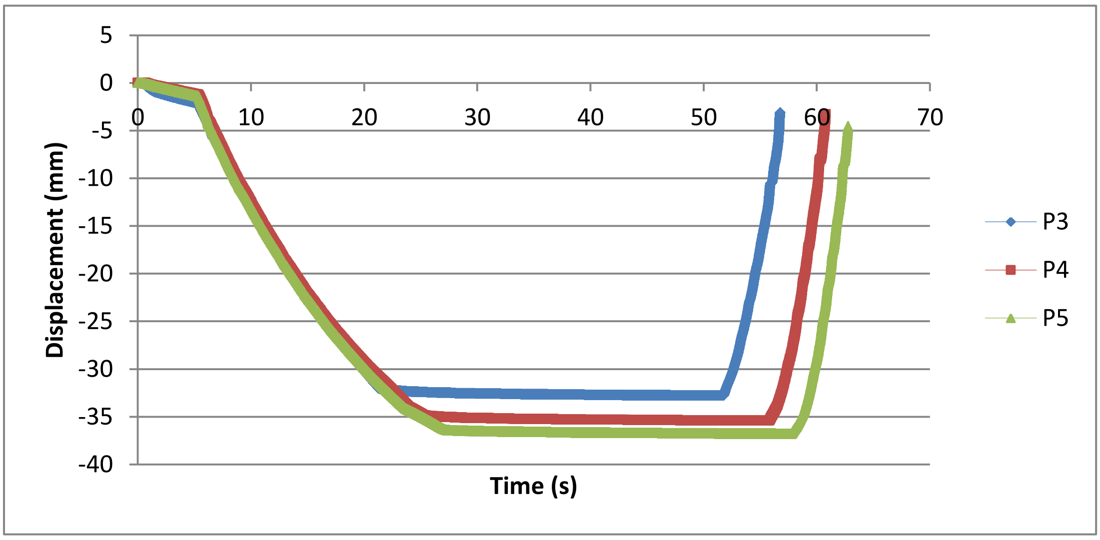
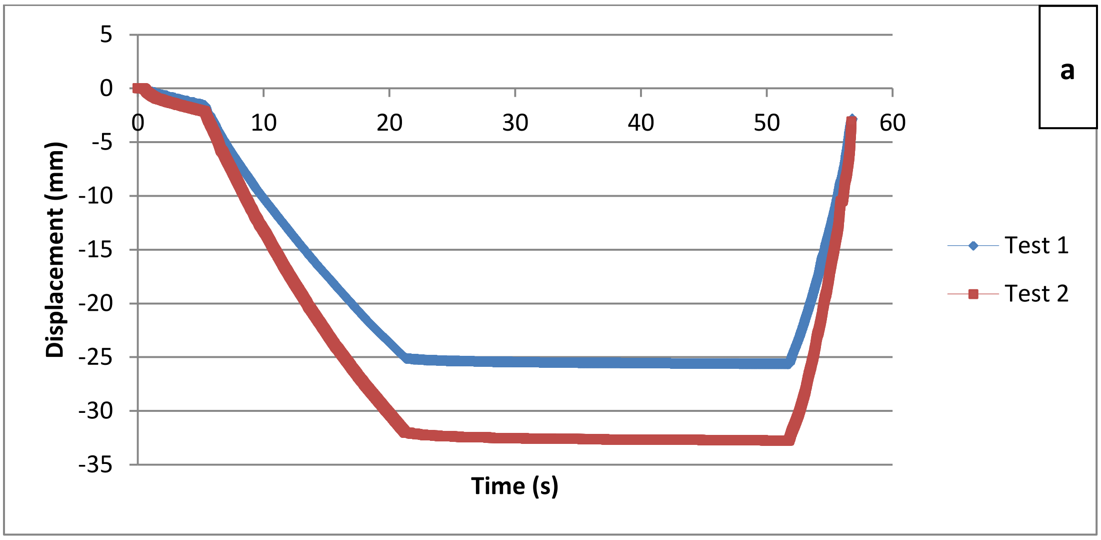


Figure 18. *Comparison of the displacement of the carbon fibre prosthetic forefoot for test loading levels P3, P4 and P5, test 2. A gradual increase in displacement was seen from test loading level P3 to P5.*

Interestingly, variation was seen in the maximum displacement of the carbon fibre material between test 1 and test 2 for the same test loading level. The difference in displacement for forefoot loading of the carbon fibre prosthetic foot at test loading level P3 is shown in figure 19. The carbon fibre prosthetic forefoot displaced to -25.63 mm in test 1, but to -32.78 mm in test 2.



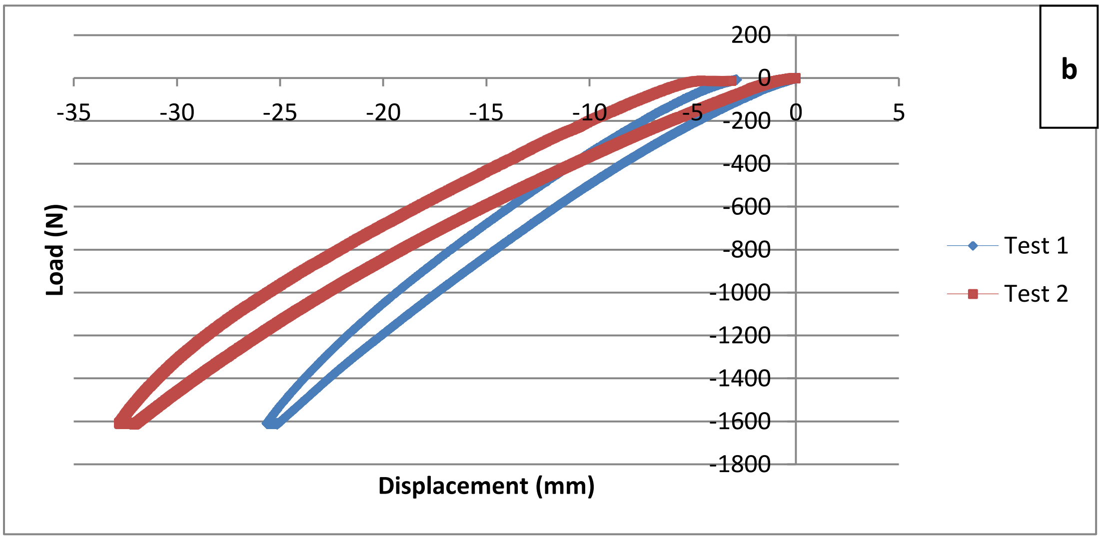


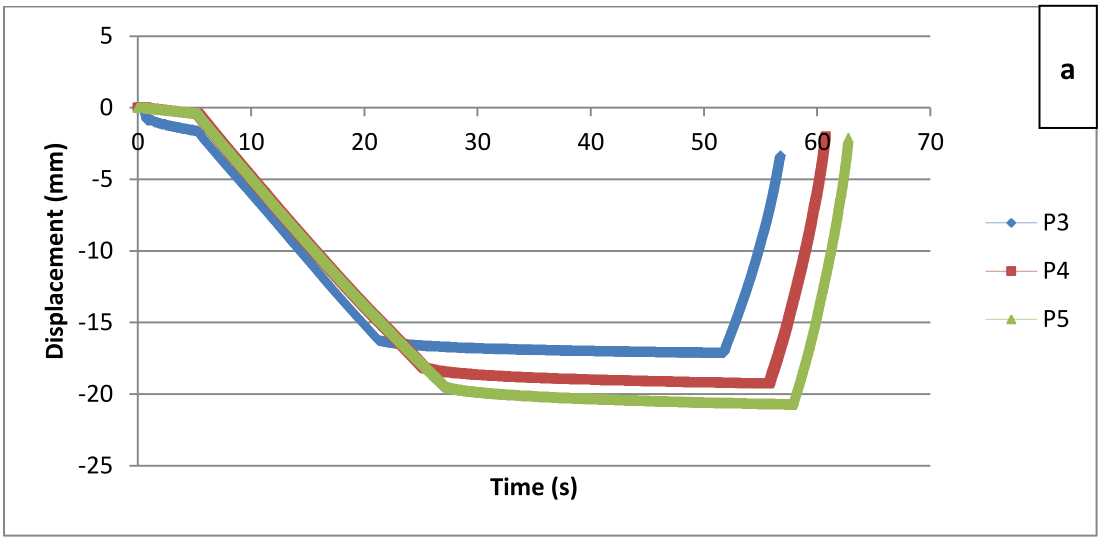
Figure 19. *Comparison of the displacement (a) and energy efficiency (b) of the carbon fibre prosthetic forefoot at test loading level P3 for test 1 and test 2. Despite these two tests having been carried out on the same carbon fibre prosthetic foot, differences in displacement and thus stiffness were seen. The energy efficiency between the two tests did not show such difference.*

5.1.3 Natural Fibre Results

Comparable to the carbon fibre, the natural fibre prosthetic forefoot's maximum displacement gradually increased with increasing test loading levels from P3 to P5 (figure 20a). The natural fibre prosthetic heel did not behave in a similar fashion, but rather seemed to be unaffected by the increasing test loading levels (figure 20b).

The natural fibre prosthetic forefoot's maximum displacement increased from -17.13 mm to -19.27 mm to -20.73 mm with increasing test loading levels from P3 to P4 to P5 respectively (figure 20a). The heel of the prosthetic foot remained relatively constant with increasing test loading levels at approximately -18.5 mm (figure 20b).

Some variation was seen between the results from different tests at the same test loading level, however not to the extent of the variations seen with the carbon fibre prosthetic foot. Figure 21 shows the variation in displacement of the natural fibre prosthetic forefoot at test loading level P5 between test 1, 2 and 3. The natural fibre prosthetic forefoot displacement for test loading level P5 was almost identical for test 1 and test 2 at a maximum displacement of -19.08 mm and -19.07 mm respectively. Test 3 showed a slight increase in maximum displacement to -20.73 mm, however this variation was much less than that seen between the different tests carried out on the carbon fibre prosthetic foot (figure 19). The energy efficiency seemed to remain relatively constant for the three different tests.



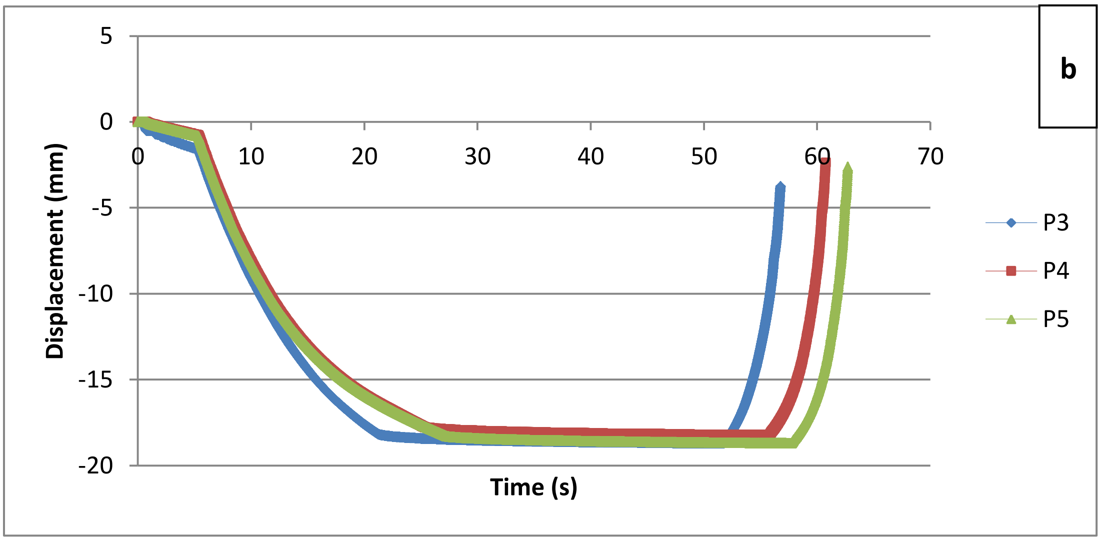


Figure 20. *Comparison of the displacement of the natural fibre prosthetic forefoot (a) and heel (b) for test loading levels P3, P4 and P5, test 3. A gradual increase in displacement was seen from test loading level P3 to P5 with forefoot loading. However loading of the heel at increasing test loading levels did not result in any significant changes in displacement.*

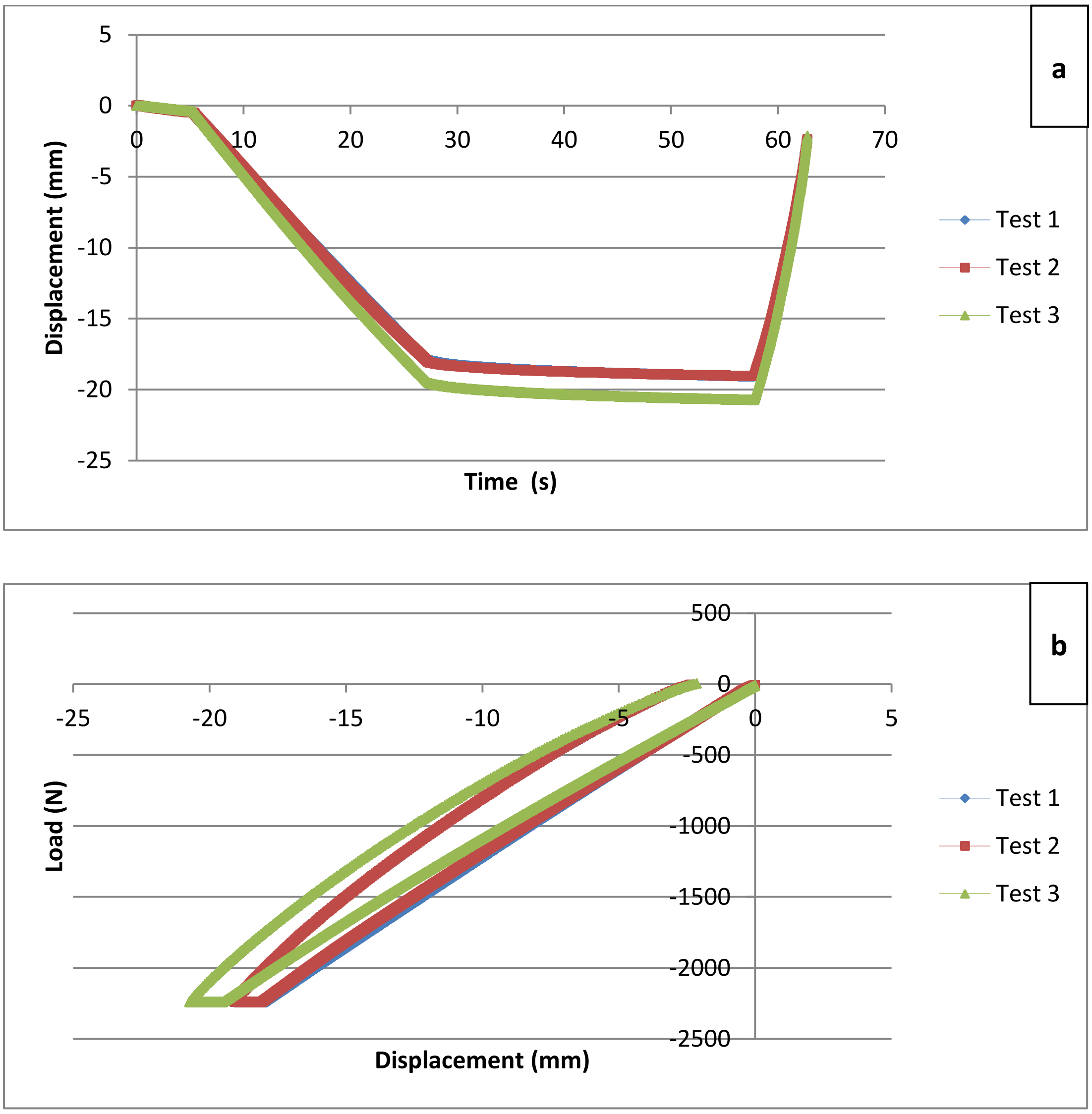


Figure 21. *Comparison of the displacement (a) and energy efficiency (b) of the natural fibre prosthetic forefoot at test loading level P5 for test 1, 2 and 3. Only a slight variation was seen between the three different tests, with the energy efficiency remaining relatively constant.*

5.2 Deformation Tracking Results

The images presented in this section were chosen as they show the most deformation, therefore making it easier to analyse the data and draw conclusions. For completeness, the rest of the images produced from the data obtained are presented in appendix B.

5.2.1 Forefoot Loading Results

The deformation of the carbon fibre and natural fibre prosthetic foot for forefoot loading at test loading level P5 are shown in figure 22a and 22b respectively. Each consecutive line is the position of the forefoot and heel in the next video frame analysed, where one frame per second was analysed.

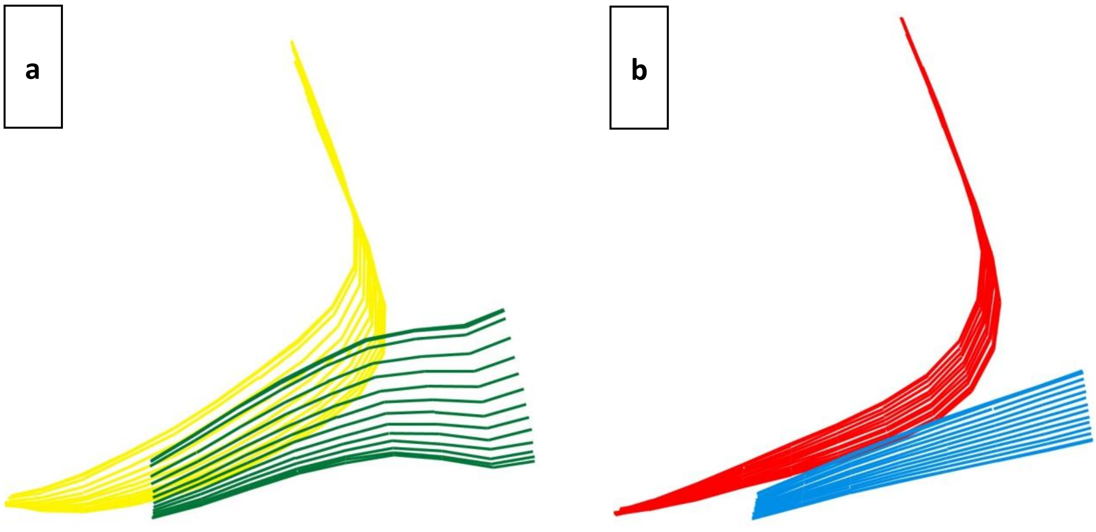


Figure 22. *Deformation of the carbon fibre (a) and natural fibre (b) prosthetic foot during the loading period of the static proof test for forefoot loading at test loading level P5.*

From figure 22 it can be seen that the carbon fibre forefoot deforms to a much greater extent than the natural fibre forefoot, especially towards the curve in the forefoot. This difference is quantified and can be seen in figure 23.

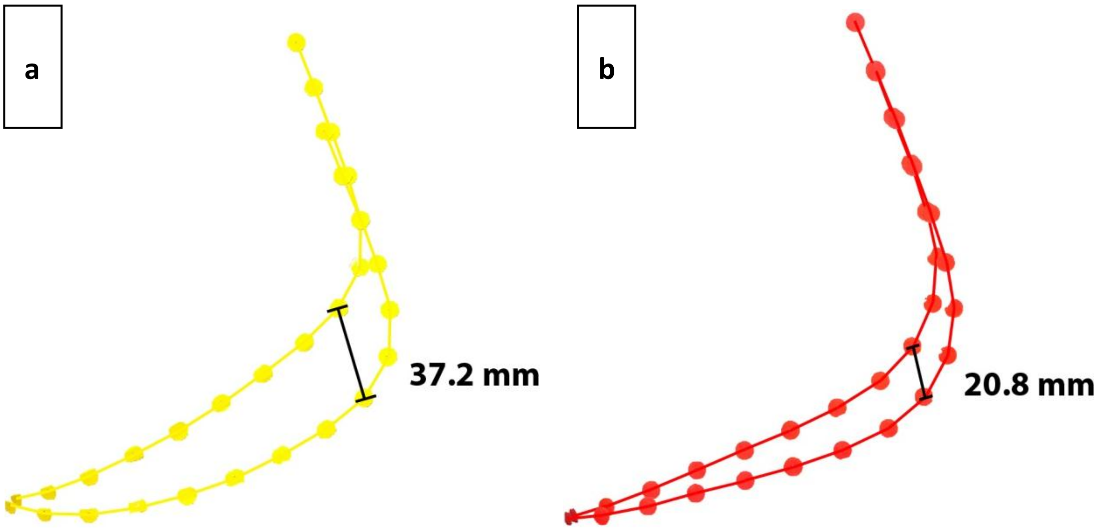


Figure 23. *Quantification of the maximum deformation experienced by the carbon fibre (a) and natural fibre (b) forefoot during forefoot loading at test loading level P5. The carbon fibre forefoot shows a much more flexible characteristic than the natural fibre forefoot.*

The carbon fibre forefoot is much more flexible than the natural fibre forefoot, as can be seen in figure 23, with the carbon fibre deforming a total of 37.2 mm at the chosen marker and the natural fibre forefoot only deforming 20.8 mm. A greater stiffness of the natural fibre forefoot is not only seen at this specific chosen point, but along the whole forefoot towards the toe region.

These results relate to the findings shown in figure 15 which also illustrate that the natural fibre forefoot is considerably stiffer compared to the carbon fibre forefoot.

5.2.2 Heel Loading Results

The deformation of the carbon fibre and natural fibre prosthetic foot for heel loading at test loading level P5 are shown in figure 24a and 24b respectively. Each consecutive line is the position of the forefoot and heel in the next video frame analysed, where one frame per second was analysed.

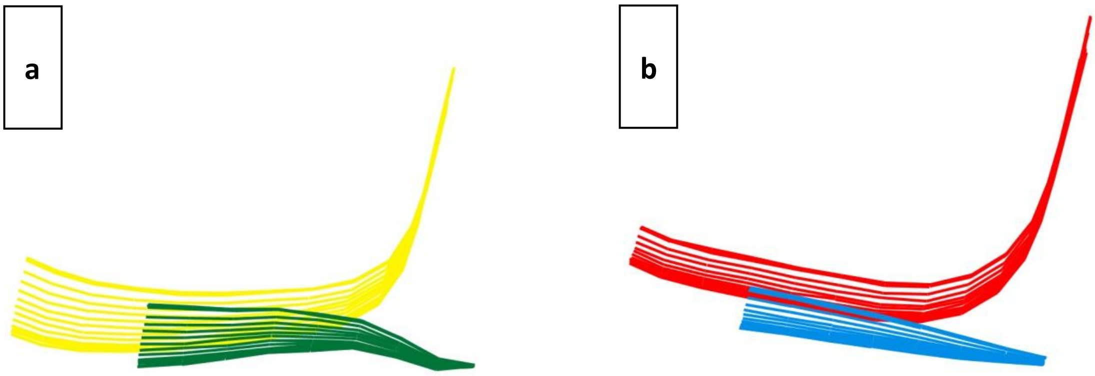


Figure 24. *Deformation of the carbon fibre (a) and natural fibre (b) prosthetic foot during the loading period of the static proof test for heel loading at test loading level P5.*

Similar to the observations for the forefoot, the carbon fibre heel is considerably more flexible compared to the natural fibre heel. Once more, this supports the previous findings illustrated in figure 16. It can also be seen that the carbon fibre forefoot also flexes on heel loading, whereas the natural fibre forefoot does not and only displaces down as a whole as the heel is being loaded.

Figure 25 illustrates the quantification of the deformation of the carbon fibre (a) and natural fibre (b) prosthetic heels at test loading level P5. The carbon fibre heel is compressed by a maximum of 24.9 mm, whereas the natural fibre heel is only compressed by 16.5 mm, once more demonstrating that the natural fibre in this design is stiffer than the carbon fibre material.

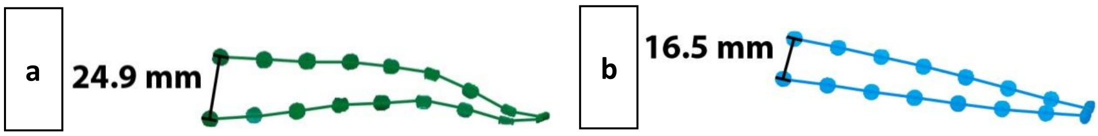


Figure 25. *Quantification of the maximum deformation experienced by the carbon fibre (a) and natural fibre (b) heel during heel loading at test loading level P5. The carbon fibre forefoot shows a considerably more flexible characteristic than the natural fibre forefoot.*

5.2.3 Overview

The above images were also created for test loading levels P3 and P4 and can be found in appendix B. Table 2 gives a summary of the amount of displacement that was experienced by the carbon fibre and natural fibre forefeet and heels at test loading levels P3 to P5, along with the percentage of deformation of the natural fibre compared to the carbon fibre. These results are from one test run only as no repeats were carried out.

The carbon fibre showed greater deformation than the natural fibre for both the forefoot and heel sections. The carbon fibre forefoot and heel also increased in maximum deformation with increasing test loading levels from P3 to P5, as expected. The natural fibre forefoot behaved in a similar fashion. Contrary to what might be hypothesised, the natural fibre heel decreased in maximum deformation with increasing test loading levels.

			Test loading level		
			P3	**P4**	**P5**
Deformation (mm)	**Forefoot**	**Natural fibre**	17.2 (52.1%)	19.1 (54.0%)	20.8 (55.9%)
		Carbon fibre	33.0	35.4	37.2
	Heel	**Natural fibre**	16.9 (72.8%)	16.8 (68.0%)	16.5 (66.3%)
		Carbon fibre	23.2	24.7	24.9

Table 2. *Overview of the amount of deformation experienced by the carbon fibre and natural fibre forefeet and heels at test loading levels P3, P4 and P5. The percentages represent the percentage of deformation of the natural fibre foot compared to the carbon fibre foot. The overall trend is that the natural fibre prosthetic foot is considerably stiffer compared to the carbon fibre prosthetic foot.*

Chapter 6: Optimisation

The natural fibre based composite prosthetic foot prototype achieved loading up to test loading level P5 without any damage. After analyses of the results obtained above, the general finding was that this prosthetic foot was considerably stiffer than the carbon fibre prosthetic foot. In an attempt to increase the flexibility of the natural fibre prosthetic foot such that it was similar in stiffness to the carbon fibre prosthetic foot, another two natural fibre prosthetic feet were produced. These were made with a slightly altered design with the hope that it would optimise its characteristics. One of the natural fibre prosthetic feet was produced with a thinner Baud curve of 14.0 mm thickness. For the other optimised design, a thinner toe section was produced, to a minimum thickness of 12.6 mm. This was achieved by filing off some of the natural fibre layers on the posterior side of the Baud curve and on the superior side of the toe region respectively. A trial-and-error approach was taken for this part of the study, so there was no defined reason for these chosen thicknesses, other than personal inkling.

Figure 26. *Image of the damage exhibited by the two ‘optimised’ natural fibre prosthetic feet with a thinner Baud curve (left) and a thinner toe section (right).*

Unfortunately, both of these prosthetic feet failed on forefoot loading at test loading level P3, approximately 2 seconds after the maximum load of 1,610 N was reached and held.

The image on the left of figure 26 shows the damage that was exhibited by the ‘optimised’ foot with a reduced thickness of the Baud curve; the image on the right of figure 26 was taken from the ‘optimised’ foot with a reduced thickness of the toe section. It can be seen from figure 26 that both feet failed at the scarf joints (i.e. where the composite layers connect), which were exposed due to filing in order to reduce the thickness of the natural fibre material and create a more flexible prototype.

Chapter 7: Discussion

7.1 General Results

Firstly, it must be noted that any characteristics of the natural fibre prosthetic foot found to be different from the carbon fibre prosthetic foot does not necessarily mean that the natural fibre foot is inferior. Indeed these properties may be an advantage depending on the environment in which the foot is to be used.

The key results from this study have shown that with the current design, the natural fibre based composite prosthetic foot is considerably stiffer compared to the carbon fibre alternative. The natural fibre forefoot was stiffer to a greater extent than the heel, only reaching a maximum of 55.9 % deformation of the carbon fibre forefoot whereas the natural fibre heel deformed by a maximum of 72.8 % of that of the carbon fibre heel (table 2). This difference was clearly evident when the P5 testing load was applied to either the forefoot or the heel (figures 15 and 16). The increased stiffness of the natural fibre prosthetic foot could be a result of its design. The material could have been too thick to allow for the desired flexibility. The relative thickness of the natural fibre forefoot when compared to the carbon fibre forefoot was greater than for the natural fibre heel when compared with the carbon fibre heel; this coincides with the stiffness. In light of the note above that different characteristics do not necessarily mean inferior characteristics, in this case, for example, a stiffer prosthetic foot may be more suitable for those patients above a certain body weight who would exert enough force on the prosthetic foot to allow an appropriate degree of flexibility.

The possibility that the greater stiffness of the natural fibre prosthetic foot was due to the greater thickness of the material was investigated with the production of two new natural fibre prosthetic feet. Their design was similar to the first prototype manufactured, with one of the ‘optimised’ forefoot having a slightly thinner material in the Baud curve and the other one have a slightly thinner toe section. Both ‘optimised’ natural fibre prosthetic forefeet tore after 2 seconds once the maximum load of 1,610 N was reached for test loading level P3 (figure 26). This could potentially be as a result of the thinner material being weaker and only able to withstand smaller loads. However, another likely possibility is the fact that the ‘optimised’ feet were not lacquered as was the case with the first natural fibre prosthetic foot prototype. The tear seen through the natural fibre material of the ‘optimised’ feet was clearly initiated at the scarf joint. This shows that this was the weakest point in the prosthetic foot. Lacquer acts to decrease the surface tension experienced when loads are applied and may have been the main reason for success in the first natural fibre prosthetic foot (Kruss.de). In order to establish this, more tests are required on prosthetic feet with ‘optimised’ designs that are also lacquered to see if this alters the results obtained in this study. Another option that could be investigated is the outcome of the results if different glue were used to fix the natural fibre layers together. The glue used in the case of the current prototype could have been a weak feature that may have resulted in the tear.

The carbon fibre material behaved relative to the test loading level, with an increase in maximum displacement experienced as the test loading level was increased from P3 to P5 (figure 18). This is also evident from the data presented in table 2 where the deformation of the carbon fibre forefoot and heel are quantified. In addition, the carbon fibre forefoot and heel showed a linear relationship between the load and displacement (figures 15-19). The displacement of the carbon fibre prosthetic foot can therefore roughly be estimated when different loads are applied by extrapolation of the graphs. The natural fibre prosthetic forefoot behaved in a similar fashion to the carbon fibre forefoot, with an increasing maximum displacement with an increase in test loading level (figure 20a and table 2). This suggests that, even though the natural fibre prosthetic forefoot is stiffer than the carbon fibre forefoot, it may behave similarly when loads are applied. The natural fibre heel, on the other hand, behaved rather unpredictably. The displacement remained relatively constant with increasing test loading levels (figure 20b). Even more surprising, the maximum displacement showed to decrease with increasing test loading levels when the deformation was tracked using the markers (table 2). The relationship between the load and the displacement of the natural fibre forefoot and heel are also not linear as seen with the carbon fibre prosthetic foot. Instead, the amount of displacement decreases as the load increases (figure 15-17). The characteristics of the natural fibre heel in this design are clearly different from that of the carbon fibre heel, which, as stated above, is not necessarily undesirable.

7.1.1 Energy Efficiency

From figure 15b it can be noted that the energy dissipated by the natural fibre prosthetic forefoot was similar to that of the carbon fibre prosthetic forefoot. For the natural fibre prosthetic heel, a greater amount of energy was dissipated compared to the carbon fibre heel (figure 16b). Once more, this does not signify that the characteristics of the natural fibre heel are undesirable, as this increased energy dissipation may aid in shock absorption on heel strike during gait. The energy efficiency of the natural fibre and carbon fibre prosthetic feet were, however, not studied extensively. These conclusions about the energy dissipated were simply drawn by observations of the load against displacement graphs generated from the static proof test data, and estimations of the area between the graphs of loading and unloading of a prosthetic foot specimen. No quantitative measures were calculated.

7.1.2 Unexpected Results

Throughout the data analysis, some unexpected results were discovered, such as the greater displacement exhibited by the natural fibre heel compared to the carbon fibre heel (figure 17). This did not follow the trends seen in the rest of the results that suggested that the natural fibre prosthetic foot was stiffer than the carbon fibre prosthetic foot. This contrasting outcome may be due to incorrect set-up of the natural fibre prosthetic foot for the heel testing. It must be ensured that the forefoot or heel only just touches the worktop of the Instron and that the displacement of the Instron is calibrated so that any displacement measured is solely due to displacement of the prosthetic foot. The natural fibre prosthetic heel may have been slightly above the Instron worktop. This would result in the measured displacement being more than the actual displacement experienced by the prosthetic foot.

Variation observed in displacement and energy efficiency between the two tests carried out on the carbon fibre forefoot and heel also presented unexpected results (figure 19). Such variation between the tests was not observed for the natural fibre prosthetic foot (figure 21) and this may also have been caused by incorrect set-up of the carbon fibre prosthetic foot in the Instron for one of the tests.

More puzzling, however, was the finding of the deformation tracking of the natural fibre prosthetic heel. This seemed to decrease as the loads were increased (table 2). Upon closer observation of the videos recorded during the static proof test, it was seen that during heel loading of the natural fibre prosthetic foot the forefoot remained so stiff that it effectively pushed on the heel section at the point where the heel joined the forefoot, curving the heel section upward. This action was exacerbated with an increase in load, curving the heel section more upward and thus reducing the amount of deformation calculated from figure 25 with the method used in this study. Other methods of quantifying the deformation of the prosthetic feet may eliminate this problem.

7.2 Future Considerations

Even though the proof of concept to use birch veneer for prosthetic manufacture has been verified with this study, many different aspects and experiments are yet to be reviewed before this could be considered for use in low-income countries, as desired.

First and foremost, the design of the natural fibre prosthetic foot must be considered and investigated. Some trials were conducted in this study with thinner sections of the natural fibre prosthetic foot in an attempt to increase its flexibility. However, the design adjustments resulted in failure of the prosthetic device. Adjustments to the design such as the application of lacquer may, however, result in an improved product that is able to withstand the loads; such experiments should be conducted. The material could also be thinned on different sections of the prosthetic feet, or on different aspects, such as, for example, thinning of the anterior aspect of the Baud curve instead of the posterior aspect. This would expose the scarf joints to compression rather than tension and may alter its properties. Another possibility for altering the characteristics of the natural fibre prosthetic foot is the type of glue used to connect the natural fibre layers. It is not known what glue was used in the birch veneer obtained, but this should be investigated.

The natural fibre prosthetic foot was designed to imitate the design of the carbon fibre Vari-Flex. This design is highly appropriate for the carbon fibre material. However, this design may not capitalise on the best characteristics of the natural fibre material. A completely different design should be considered with the objective of delivering a natural fibre prosthetic foot with the optimal characteristics, these being lightweight, stiff and durable, as seen with the carbon fibre prosthetic foot. Another thing to consider is the option to work in a 3 dimensional sense, rather than just 2 dimensional. With this it is meant that the layers could be constructed in a convex or concave manner. For example, the shank could be designed with a convex shape, which then changes to a concave shape at the forefoot. This would provide extra strength and thus may allow for a thinner material to be used to increase flexibility whilst maintaining its strength.

Another aspect to consider, that was not addressed in this study, is the roll-over of the prosthetic feet. Sam *et al.* (2004) determined the roll-over shapes of 11 different prosthetic feet commonly used in low-income countries, including the Jaipur foot and two of those manufactured by the ICRC. They found that all except the Jaipur foot experienced a roll-over shape comparable to the typical SACH foot. The Jaipur foot showed a smaller radius of the roll-over shape, allowing a greater amount of dorsiflexion of the foot which is beneficial for certain activities such as squatting and tree climbing, commonly carried out in countries such as India. The roll-over shape can therefore aid in characterisation of foot function and is important to study for the natural fibre prosthetic foot used in this study.

Before the full potential of this natural fibre prosthetic foot can be determined, the mechanical testing according to BS EN ISO 10328:2006 has to be completed with more repeats of the experiments conducted in this study in order to verify the results obtained in this study. Completion of the ultimate strength test to determine the load at fracture and cyclic fatigue testing to determine its durability are also required. Once this has been established, and if deemed durable, the natural fibre prosthetic foot will need to be tested in a clinical setting, where trans-tibial amputees would be required to wear the natural fibre prosthetic foot and biomechanical data could be obtained during gait. From this, quantified energy efficiency could be calculated. A questionnaire to be completed by the volunteers may also be beneficial to determine the individuals' experience on walking on the prosthetic device, in terms of comfort and effort. This is a fundamental factor as patient satisfaction is essential since only a foot that works for the patient will really benefit that patient to the fullest extent possible.

Lastly, the final design should be considered in terms of the cosmetics and the country where it will be used. The design of the prosthetic foot must be within certain parameters to allow the addition of a cosmetic cover whilst maintaining the size of a human foot. The prosthetic foot should also be weather proof. Those manufactured from rubber are naturally waterproof. The birch veneer used in this study is not, and it may be of use to explore the possibility of encasing the natural fibre prosthetic foot in rubber or an alternative waterproof cover. Most low-income countries also experience considerable amounts of sunlight and high temperatures. As such, determining the effects of UV-rays and temperature on the birch veneer could be beneficial. The Jaipur foot was designed for the culture of India, this foot should be designed bearing in mind the culture of the country to which it is to be delivered, otherwise patients might prefer not to have a prosthetic device as against one which does not suit their needs.

Chapter 8: Conclusion

The results from this study have illustrated that the natural fibre based composite prosthetic foot prototype as manufactured, complies with BS EN ISO 10328:2006 for the static proof test up to test loading level P5 (the maximum level tested). It can therefore be said that the use of birch veneer as a low-cost alternative to carbon fibre for prosthetic manufacture in low-income countries has some potential to be successful. Many more experiments are required, including fulfilment of the mechanical testing according to BS EN ISO 10328:2006 and clinical assessments with trans-tibial amputees. Certain aspects not addressed in this study are also to be considered in terms of the prosthetic foot design and the country in which the prosthetic foot will be made accessible. Once these factors have been investigated, an even better indication will be provided of the potential use of birch veneer as a low-cost alternative to carbon fibre for prosthetic manufacture in low-income countries.

References

Ahmad, N., Thomas, G. N., Gill, P., Chan, C., Torella, F. (2014). Lower limb amputation in England: Prevalence, regional variation and relationship with revascularisation, deprivation and risk factors. A retrospective review of hospital data. *Journal of the Royal Society of Medicine* **107 (12)**, 483-489.

BS EN ISO 10328:2006. (2006). Prosthetics – Structural testing of lower-limb prostheses – Requirements and test methods. *The British Standards Institution*.

Burgess, E. M., Poggi, D. L., Hittenberger, D. A., Zettl, J. H., Moeller, D. E., Carpenter, K. L., Forsgren, S. M. (1985). Development and preliminary evaluation of the VA Seattle foot. *Journal of Rehabilitation Research and Development* **22 (3)**, 75-84.

Centers For Mobility. Leg prosthesis and lower limb prosthetics. [online]. Available from: http://www.centersformobility.com/prosthetics_2.html#maintenance [Accessed 2 July 2015]

Chand, N., Tamrakar, G. (2015). Mechanical and surface studies of bamboo fibre extracted by different methods. *Journal of Scientific Research in Physical & Mathematical Science* **2 (5)**, 1-9.

Cosgrove, B. (2013). Sum of the parts: Prosthetics through the centuries. Jaipur foot, interior view. [online]. Available from: http://newsfeed.time.com/2013/02/23/sum-of-the-parts-prosthetics-through-the-centuries/slide/inside-foot/ [Accessed 30 July 2015]

Deshpande, A. P., Rao, M. B., Rao, C. L. (2000). Extraction of bamboo fibres and their use as reinforcement in polymeric composites. *Journal of Applied Polymer Science* **76**, 83-92.

Drongelen, S. van. (2000). Energy storing feet: Beneficial? [online]. Available from: http://www.monash.edu.au/rehabtech/research/enerfeet.pdf [Accessed 26 July 2015]

Endolite. Endolite ESK + MKL. [online]. Available from: http://www.endolite.co.uk/products/endolite-esk-mkl [Accessed 31 July 2015]

Exceed. [online]. Available from: http://www.exceed-worldwide.org/ [Accessed 31 July 2015]

footEducation. Available from: http://www.footeducation.com/foot-and-ankle-basics/biomechanics-of-foot-and-ankle/biomechanics-of-walking-gait/ [Accessed 15 June 2015]

Geil, M. D. (2000). Energy storage and return in dynamic elastic response prosthetic feet. *Pediatric Gait: A New Millennium in Clinical Care and Motion* 134-142.

Hafner, B. J., Sanders, J. E., Czerniecki, J. M., Fergason, J. (2002). Transtibial energy-storage-and-return prosthetic devices: A review of energy concepts and proposed nomenclature. *Journal of Rehabilitation Research and Development* **39 (1)**, 1-11.

Holman, N., Young, R. J., Jeffcoate, W. J. (2012). Variation in the recorded incidence of amputation of the lower limb in England. *Diabetologia* **55 (7)**, 1919-1925.

Instron. ElectroPuls E10000 Linear-Torsion All-Electric Test Instrument. [online]. Available from: http://www.instron.com/en-gb/products/testing-systems/dynamic-and-fatigue-systems/electropuls/e10000---linear-torsion [Accessed 25 July 2015]

International Committee of the Red Cross. (2007). Polypropylene technology. [online]. Available from: https://www.icrc.org/eng/assets/files/other/icrc-002-0913.pdf [Accessed 8 August 2015]

International Committee of the Red Cross. (2014). Art from the past: The foot cast that marks a historical fight to ban landmines. [online]. Available from: http://blogs.redcross.org.uk/international/2014/10/art-from-the-past-foot-cast-ban-landmines/ [Accessed 14 July 2015]

International Diabetes Federation. (2013). *IDF Diabetes Atlas*. 6th Ed. Brussels, Belgium: International Diabetes Federation.

Jain, S., Kumar, R., Jindal, U. C. (1992). Mechanical behaviour of bamboo and bamboo composite. *Journal of Materials Science* **27**, 4598-4604.

Jaipur Foot. How we do: Fabrication processes and materials. [online]. Available from: http://jaipurfoot.org/how_we_do/fabrication_processes.html#.VbowT_lViko [Accessed 30 July 2015]

Jensen, J. S., Treichl, H. B. (2007). Mechanical testing of prosthetic feet utilized in low-income countries according to ISO-10328 standard. *Prosthetics and Orthotics International* **31 (2)**, 177-206.

Klasson, B. L. (1995). Carbon fibre and fibre lamination in prosthetics and orthotics: Some basic theory and practical advice for the practitioner. *Prosthetics and Orthotics International* **19**, 74-91.

Kruss. Surface treatment of wood. [online]. Available from: http://www.kruss.de/applications/expertise/surface-treatment/surface-treatment-of-wood/ [Accessed 5 August 2015]

Joint Master Project. (2015). *LegBank Colombia Report.* The Netherlands: TU Delft Design Engineering.

Michael, J. (1987). Energy storing feet: A clinical comparison. *Clinical Prosthetics and Orthotics* **11 (3)**, 154-168.

NBC News. Haiti amputees. [online]. Available from: http://haitiamputees.nbcnews.com/_news/2010/03/19/4040341-limb-loss-a-grim-growing-global-crisis [Accessed 13 July 2015]

Niagara Foot. The Niagara foot: Superior function & energy return at an affordable price. [online]. Available from: http://www.niagarafoot.com/niagarafoot/index.html [Accessed 30 July 2015]

Orthocare Indonesia. Prosthetic cases. [online]. Available from: http://www.orthocareindonesia.com/type-of-prosthesis#.VcXIvyaqqkp [Accessed 8 August 2015]

Össur. [online]. Available from: http://www.ossur.co.uk/ [Accessed 27 July 2015]

Ottobock. [online]. Available from: http://corporate.ottobock.co.uk/cps/rde/xchg/ob_uk_en/hs.xsl/index.html [Accessed 2 July 2015]

Postema, K., Hermens, H. J., Vries, J. de., Koopman, H. F. J. M., Eisma, W. H. (1997). Energy storage and release of prosthetic feet. Part 1: Biomechanical analysis related to user benefits. *Prosthetics and Orthotics International* **21**, 17-27.

ProPortion Foundation. (2014). Social business in prosthetics – Colombia. [online]. Available from: http://proportionfoundation.org/?p=3649 [Accessed 25 June]

Radcliffe, C. W. (1977). Above-knee prosthetics. *Prosthetics and Orthotics International* **1**, 146-160.

Radcliffe, C. W., Foort, J. (1961). *The patellar-tendon-bearing below-knee prosthesis.* Revised Ed. San Francisco: Biomechanics laboratory.

Sadikot, S., Nigam, A., Jain, A. (2007). The Jaipur foot: An effective low-cost prosthesis for people with diabetes. *DiabetesVoice* **52 (3)**, 18-21.

Saheb, D. N., Jog, J. P. (1999). Natural fibre polymer composites: A review. *Advances in Polymer Technology* **18 (4)**, 351-363.

Sam, M., Hansen, A. H., Childress, D. S. (2004). Characterisation of prosthetic feet used in low-income countries. *Prosthetics and Orthotics International* **28**, 132-140.

Schneider, K., Hart, T., Zernicke, R. F., Setoguchi, Y., Oppenheim, W. (1993). Dynamics of below-knee child amputee gait: SACH foot versus Flex foot. *Journal of Biomechanics* **26 (10)**, 1191-1204.

Schuch, C. M. (1992). *Atlas of Limb Prosthetics: Surgical, prosthetic, and rehabilitation principles.* 2nd Ed. Rosemont IL: American Academy of Orthopaedic Surgeons.

Strait, E. (2006). Prosthetics in developing countries. [online]. Available from: http://www.oandp.org/publications/resident/pdf/DevelopingCountries.pdf [Accessed 29 July 2015]

Stewart, C. P. U. (2008). Synopsis of causation: Lower limb amputation. *Ministry of Defence* [online]. Available from: https://www.gov.uk/government/uploads/system/uploads/attachment_data/file/384461/amputations_lower_limb.pdf [Accessed 28 July 2015]

Suherman, R. S. (2012). 41 innovative products selected as finalists in 2012 medical design excellence awards. [online]. Available from: https://infoalkes.wordpress.com/2012/04/05/41-innovative-products-selected-as-finalists-in-2012-medical-design-excellence-awards/ [Accessed 30 July 2015]

The World Bank. Data: Country and lending groups. [online]. Available from: http://data.worldbank.org/about/country-and-lending-groups [Accessed 8 August 2015]

Walsh, N. E., Walsh, W. S. (2003). Rehabilitation of landmine victims – The ultimate challenge. *Bulletin of the World Health Organization* **81 (9)**, 665-670.

Winter, D. A., Sienko, S. E. (1988). Biomechanics of below-knee amputee gait. *Journal of Biomechanics* **21 (5)**, 361-367.

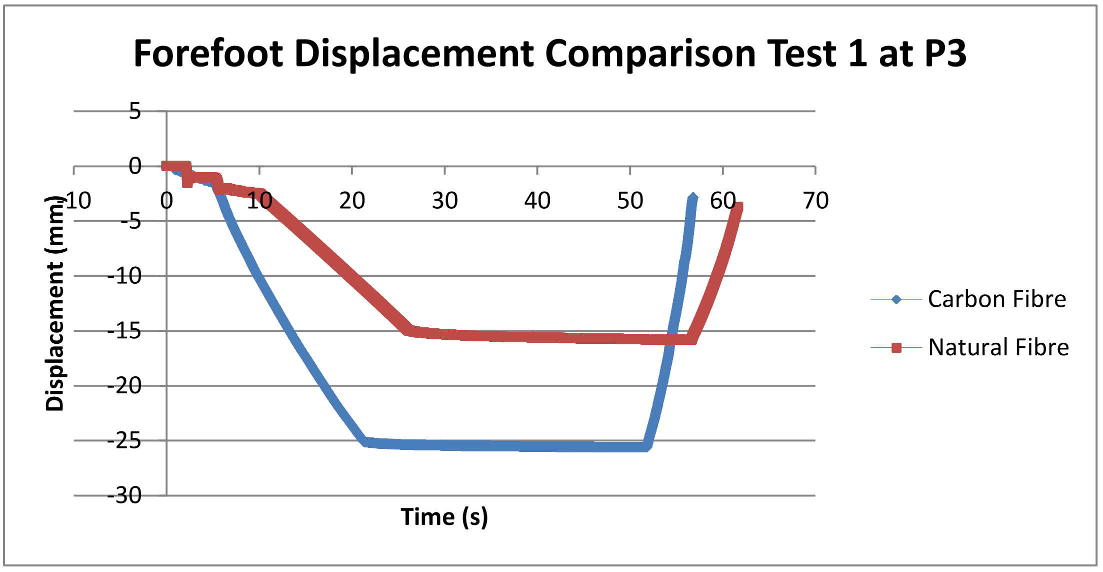
World Health Organisation (WHO). (2005a). World Diabetes Day: Too many people are losing lower limbs unnecessarily to diabetes. [online]. Available from: http://www.who.int/mediacentre/news/releases/2005/pr61/en/ [Accessed 25 June 2015]

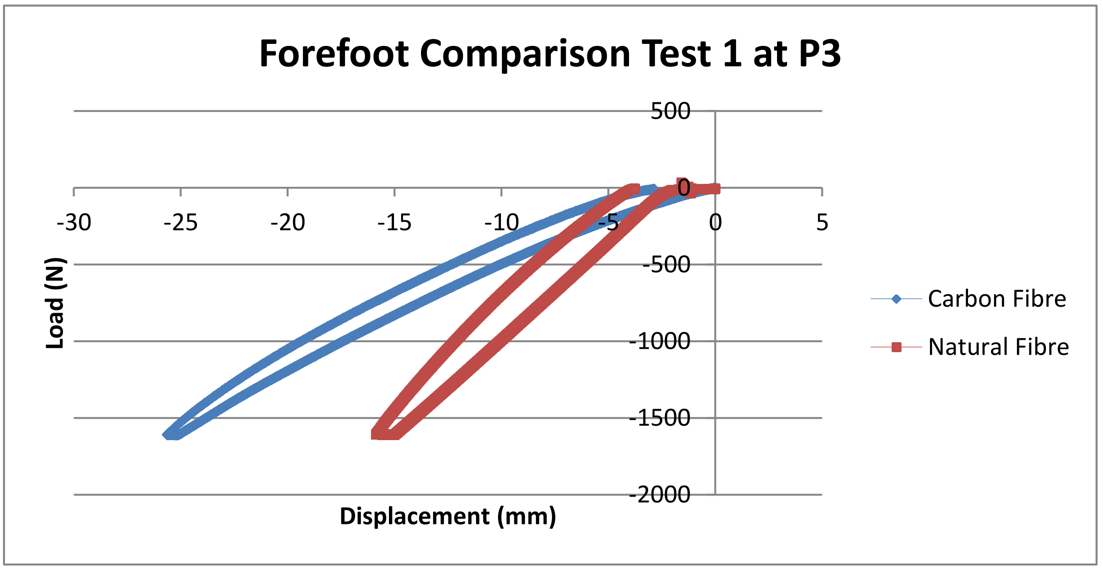
World Health Organisation (WHO). (2005b). Guidelines for Training Personnel in Developing Countries for Prosthetics and Orthotics Services. [online]. Available from: http://whqlibdoc.who.int/publications/2005/9241592672.pdf [Accessed 22 July 2015]

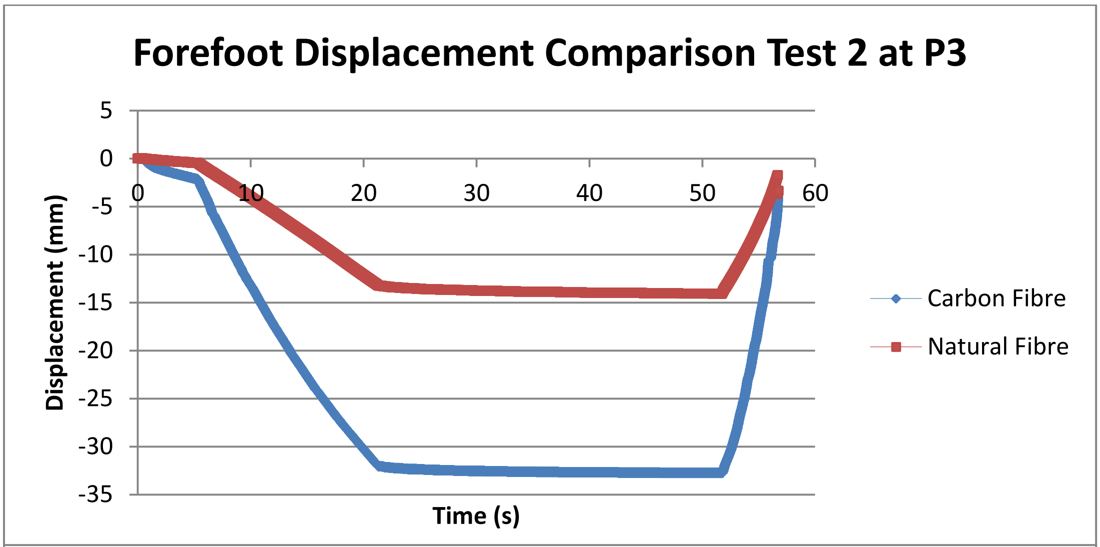
Appendix A

A.1 Static Proof Test Results Continued

A.1.1 Comparison of Natural Fibre with Carbon Fibre



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-10
-15
-20
-25
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-35
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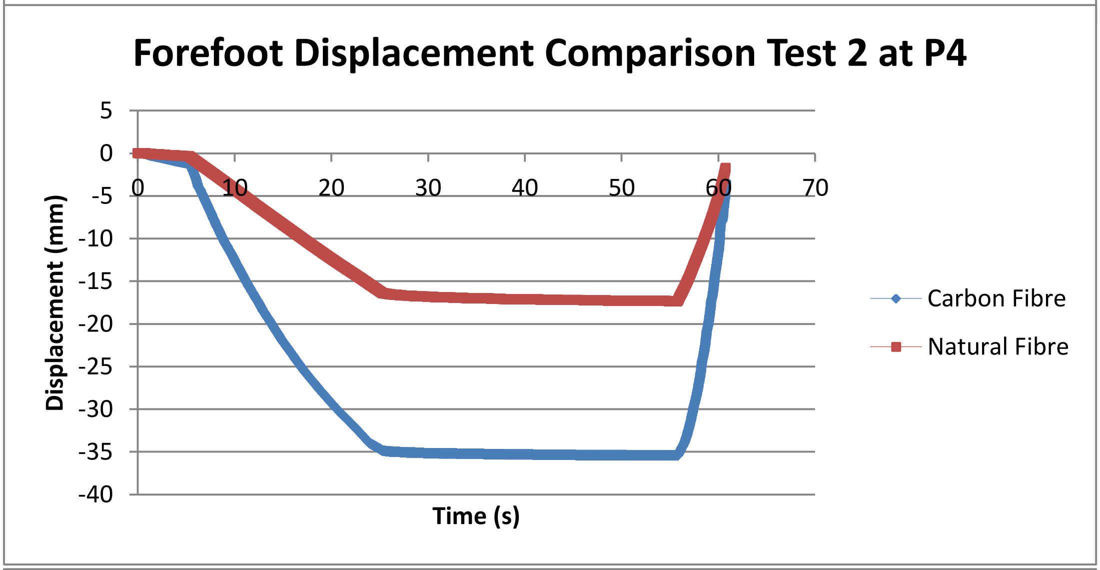
10
20
30
40
50
60
Displacement (mm)
Time (s)
Carbon Fibre
Natural Fibre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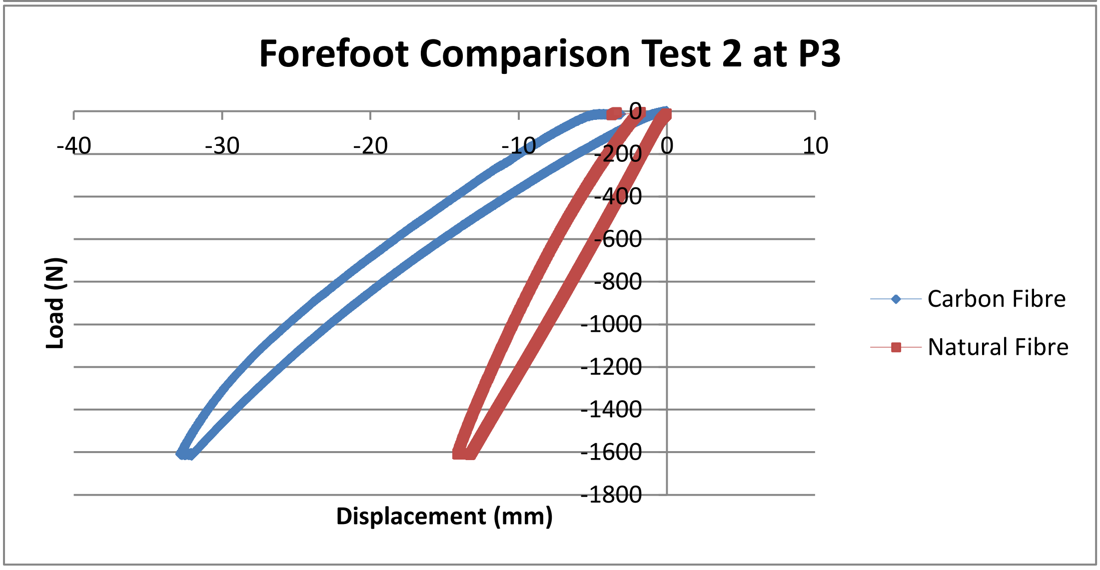
30
40
50
60
70
Displacement (mm)
Time (s)
Carbon Fibre
Natural Fibre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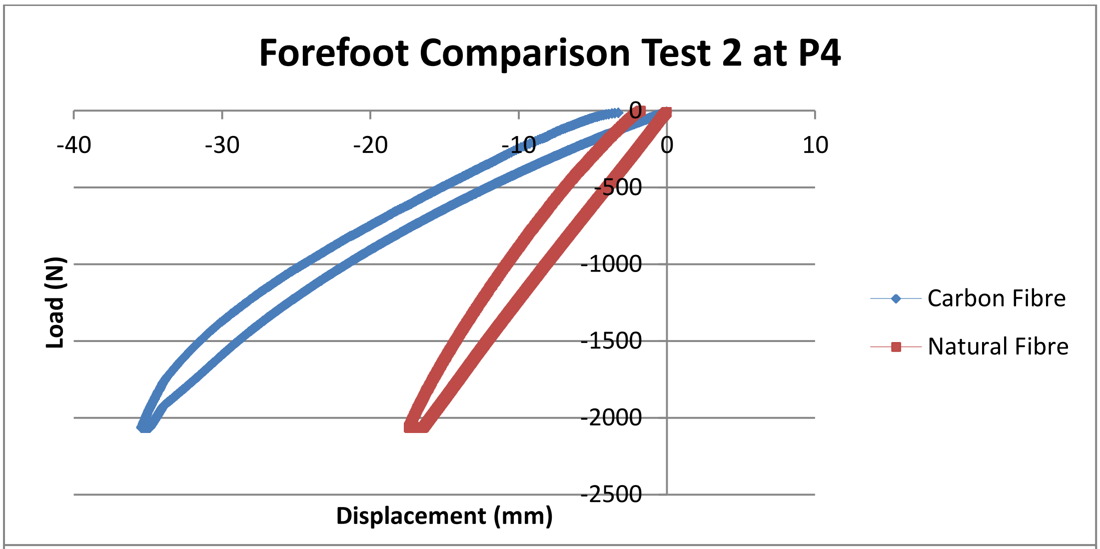
-20
-10
0
10
0
-200
-400
-600
-800
-1000
-1200
-1400
-1600
-1800
Load (N)
Displacement (mm)
Carbon Fibre
Natural Fibre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10
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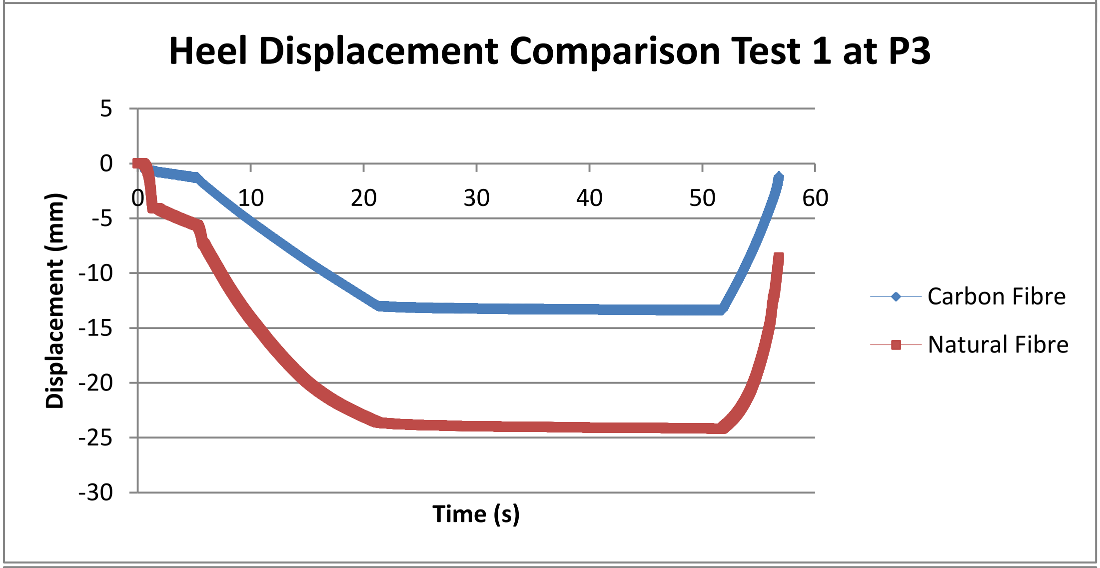
Load (N)
Displacement (mm)
Carbon Fibre
Natural Fibre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-10
-15
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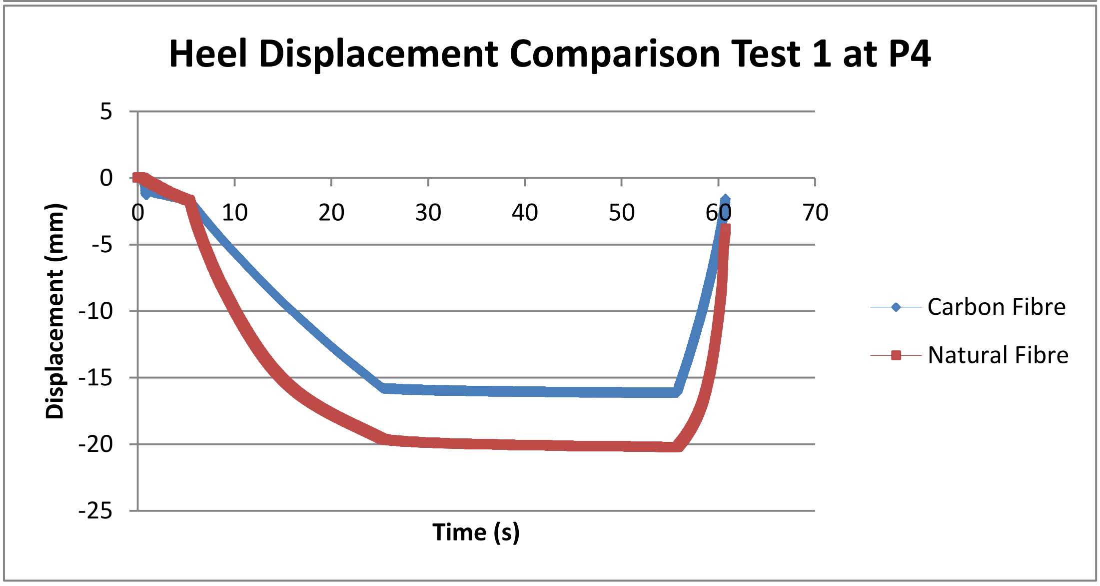
-25
-30
0
10
20
30
40
50
60
Displacement (mm)
Time (s)
Carbon Fibre
Natural Fibre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-10
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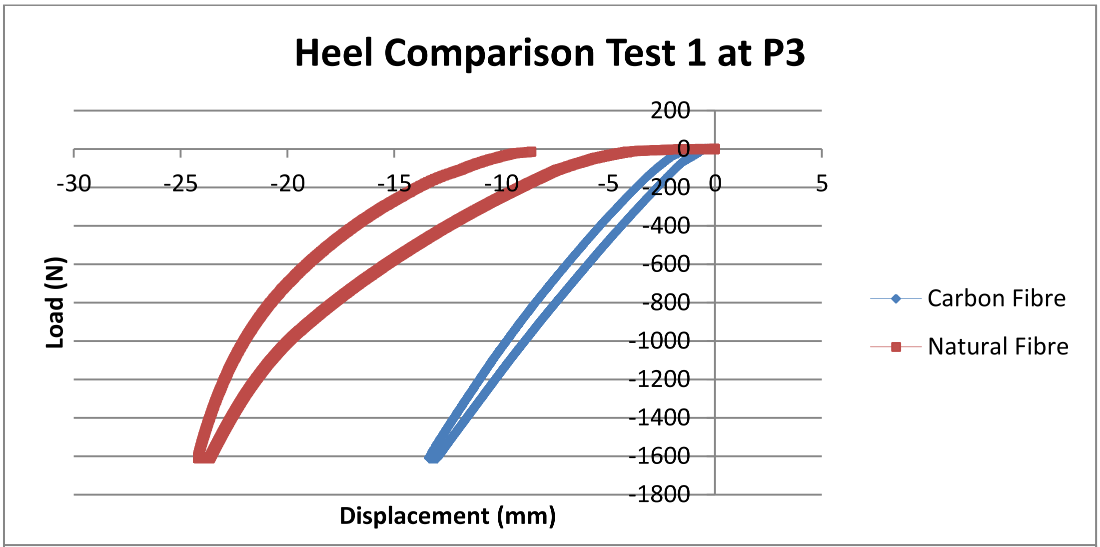
20
30
40
50
60
70
Displacement (mm)
Time (s)
Carbon Fibre
Natural Fibre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Displacement (mm)
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-400
-600
-800
-1000
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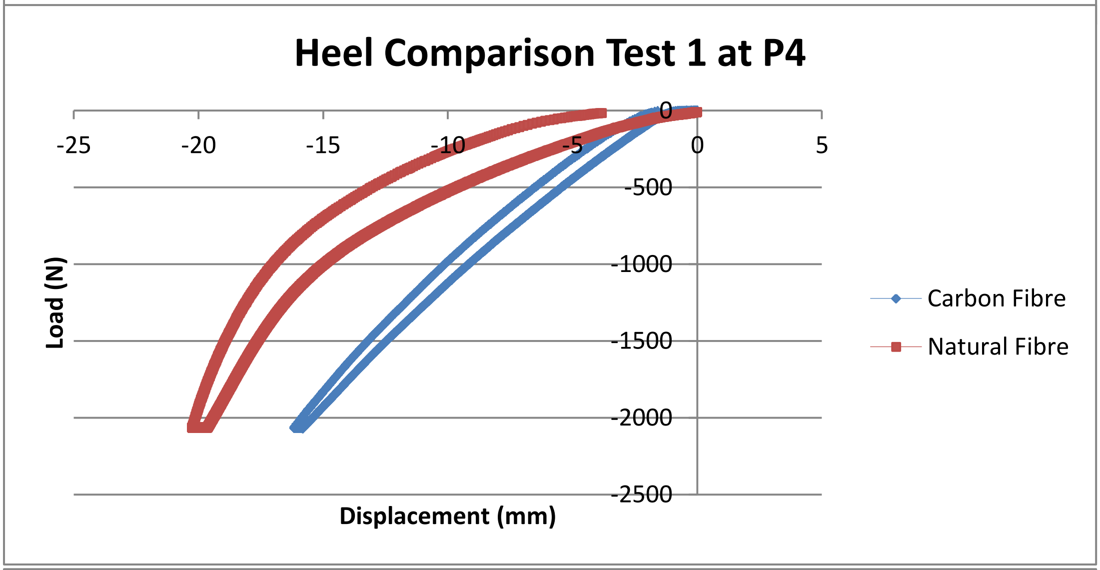
-1600
-1800
-30
-25
-20
-15
-10
-5
0
5
Carbon Fibre
Natural Fibre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Displacement (mm)
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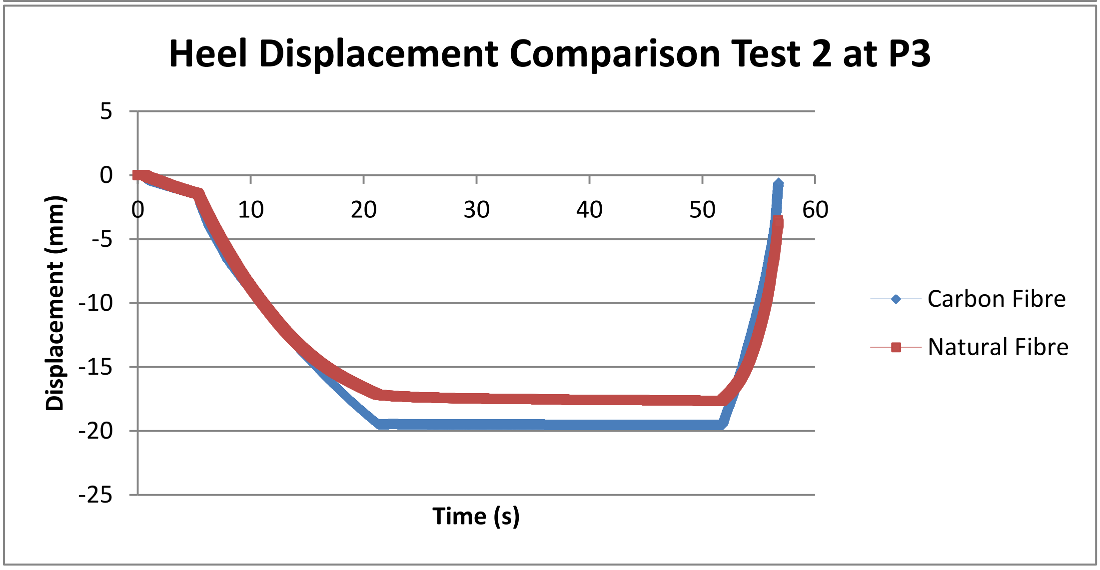
-2500
-25
-20
-15
-10
-5
0
5
Carbon Fibre
Natural Fibre


Heel Displacement Comparison Test 2 at P3
Displacement (mm)
Time (s)
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-5
-10
-15
-20
-25
0
10
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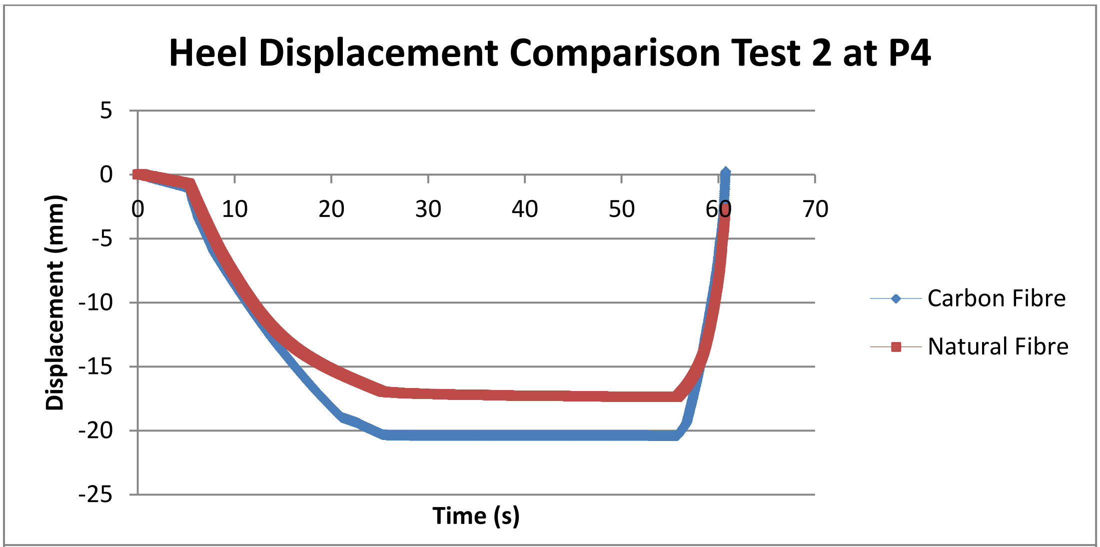
30
40
50
60
Carbon Fibre
Natural Fibre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Displacement (mm)
5
0
-5
-10
-15
-20
-25
0
10
20
30
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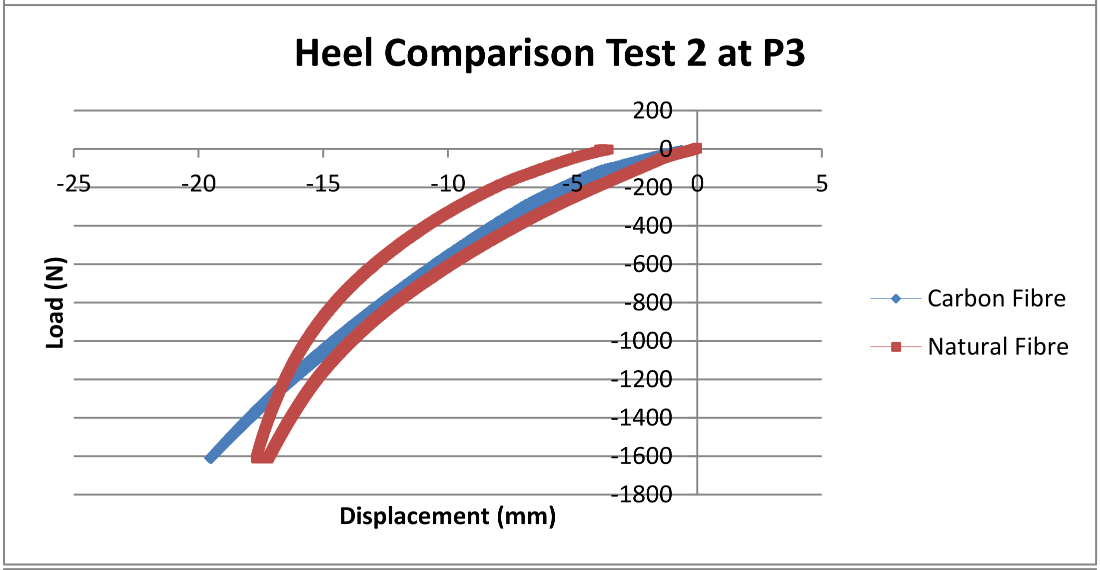
60
70
Time (s)
Carbon Fibre
Natural Fibre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Load (N)
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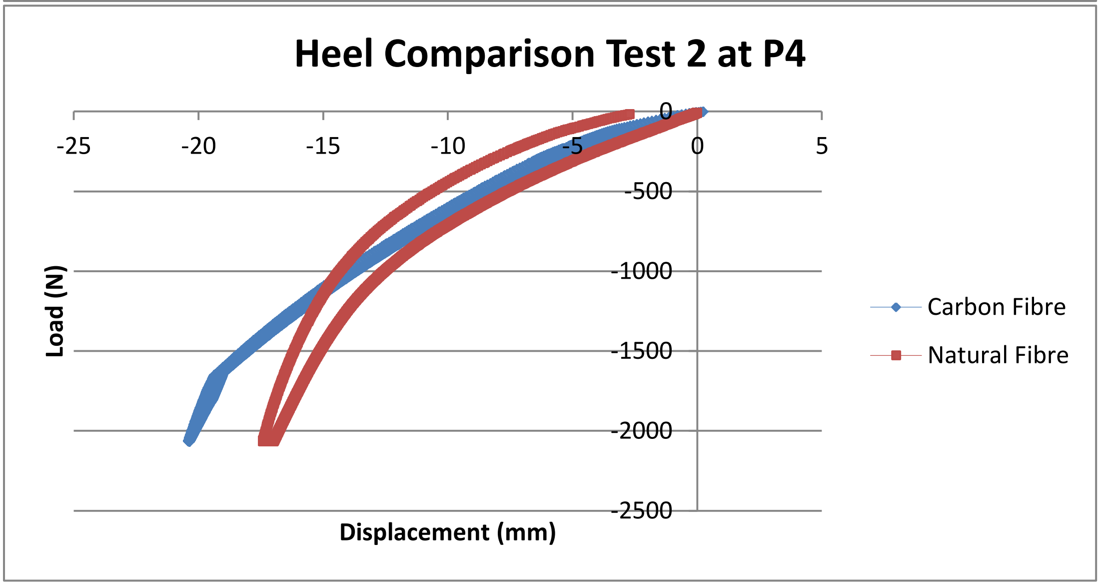
-200
-400
-600
-800
-1000
-1200
-1400
-1600
-1800
-25
-20
-15
-10
-5
0
5
Displacement (mm)
Carbon Fibre
Natural Fibre


Heel Comparison Test 2 at P4
Load (N)
0
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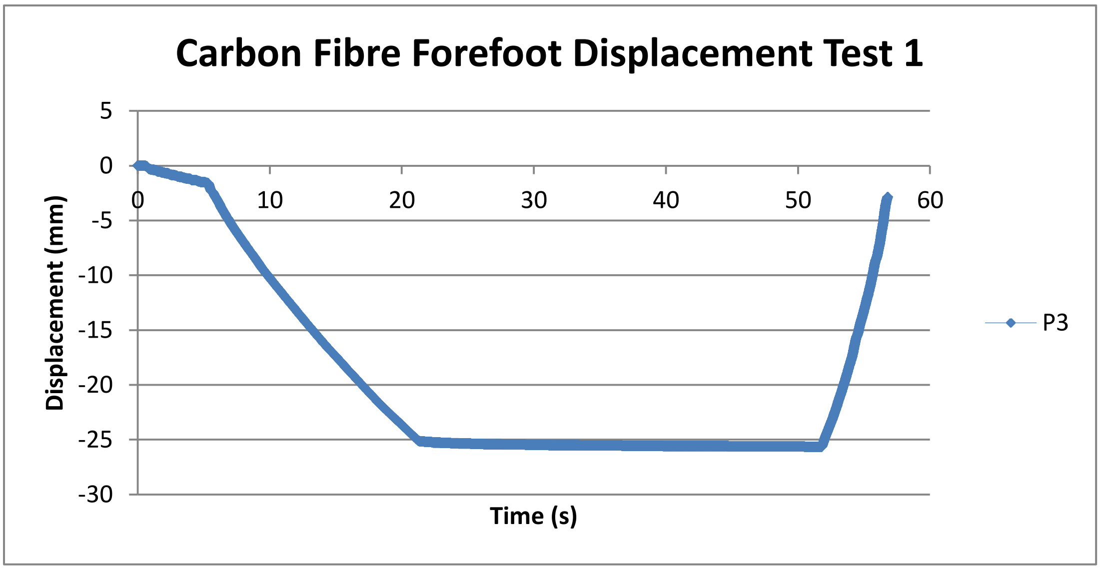
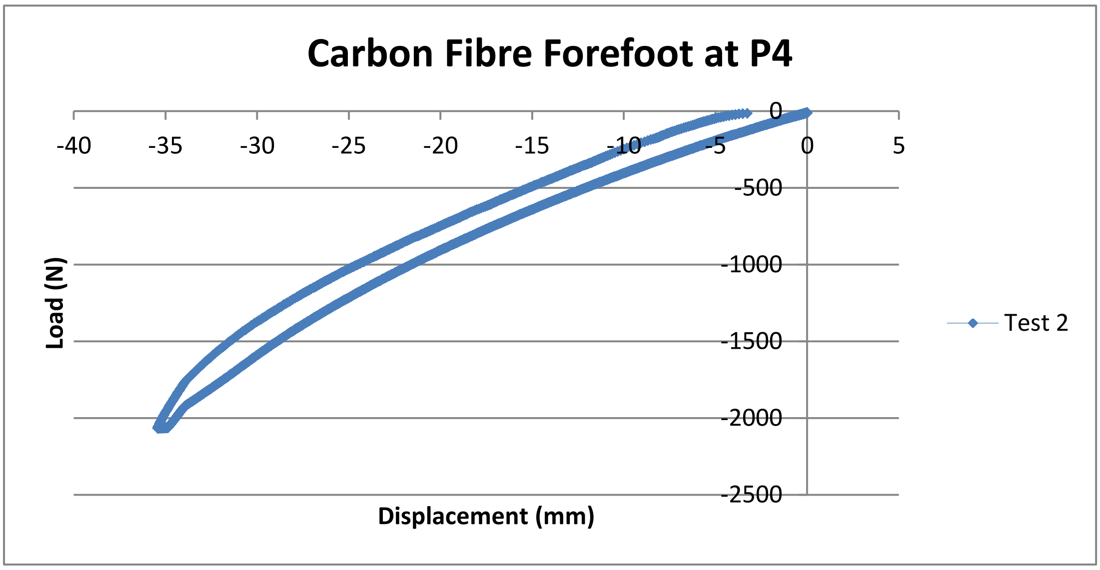
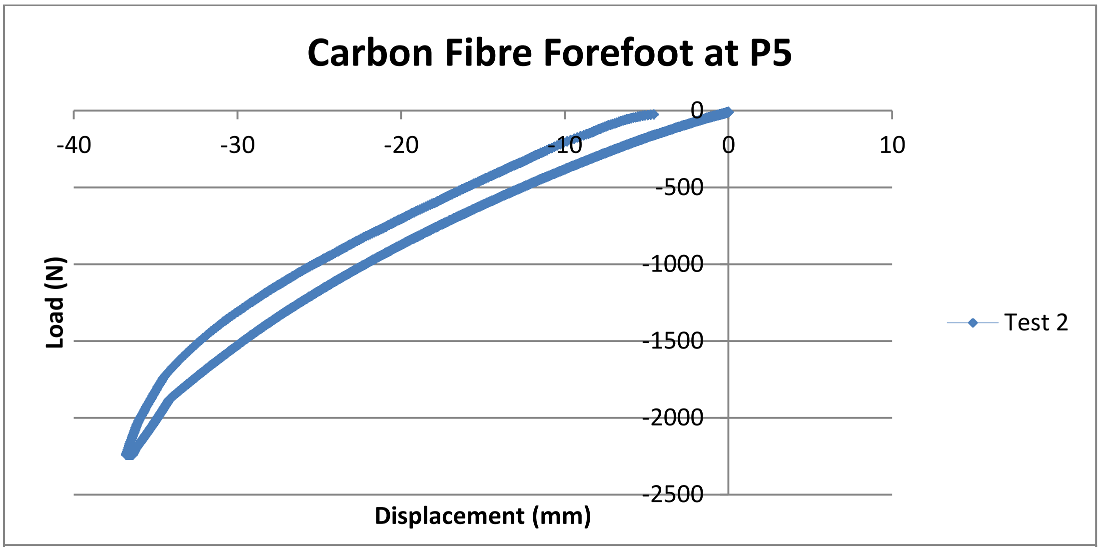
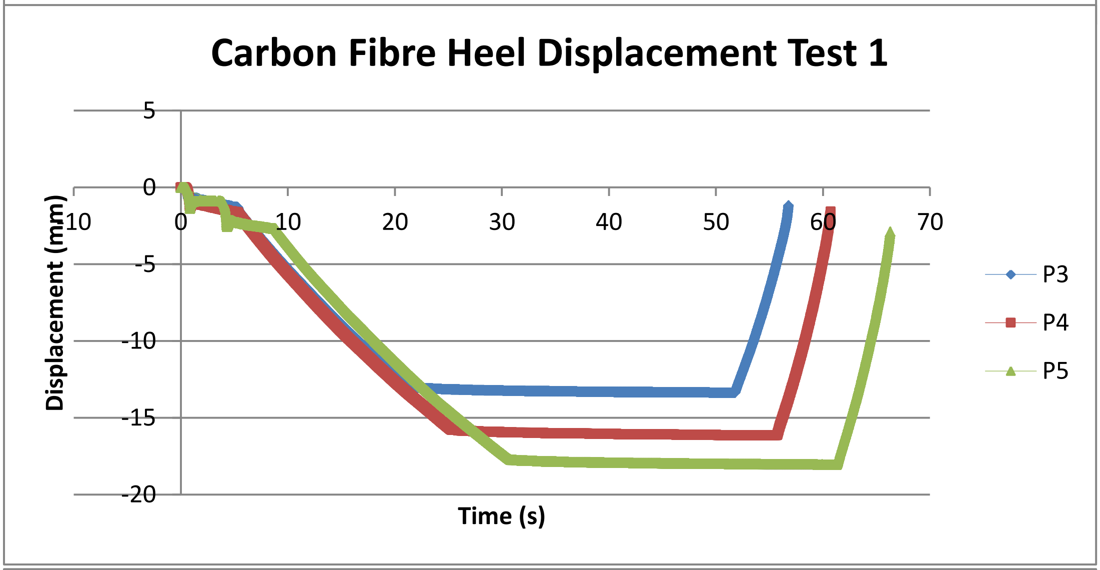
-5
0
5
Displacement (mm)
Carbon Fibre
Natural Fibre

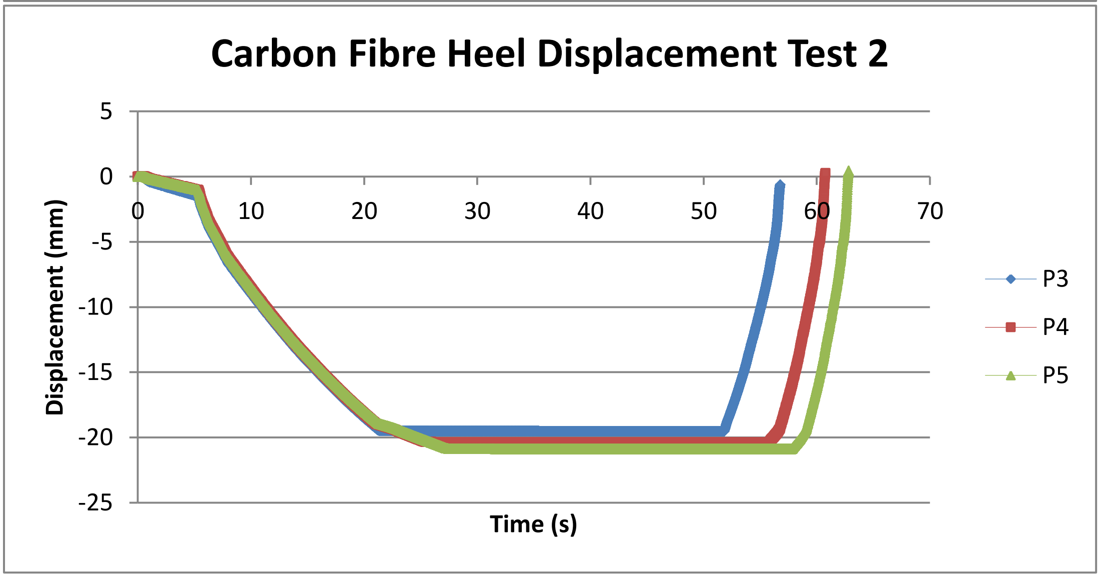
A.1.2 Carbon Fibre Results

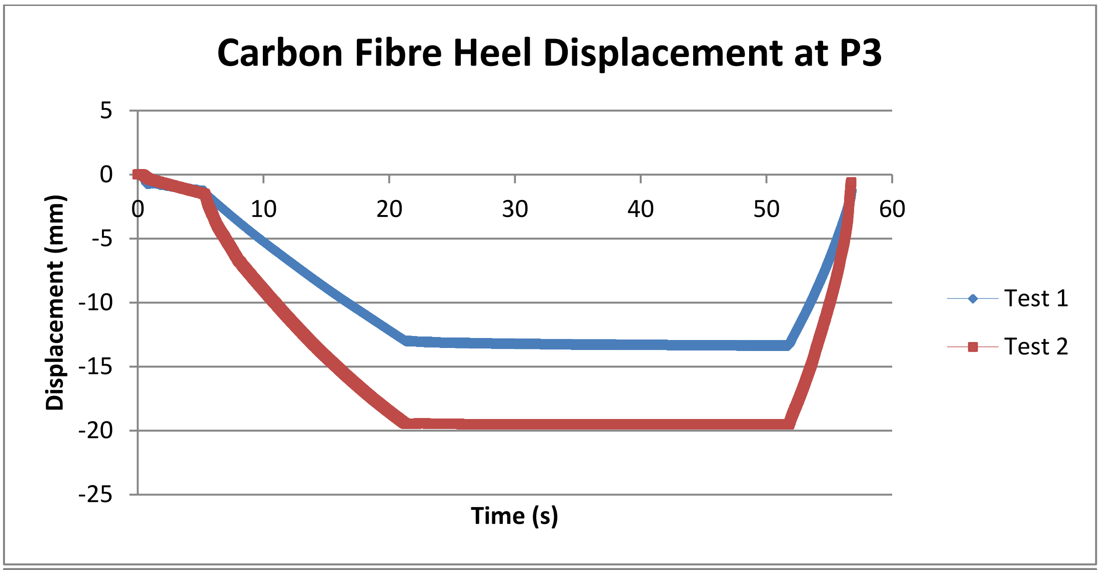


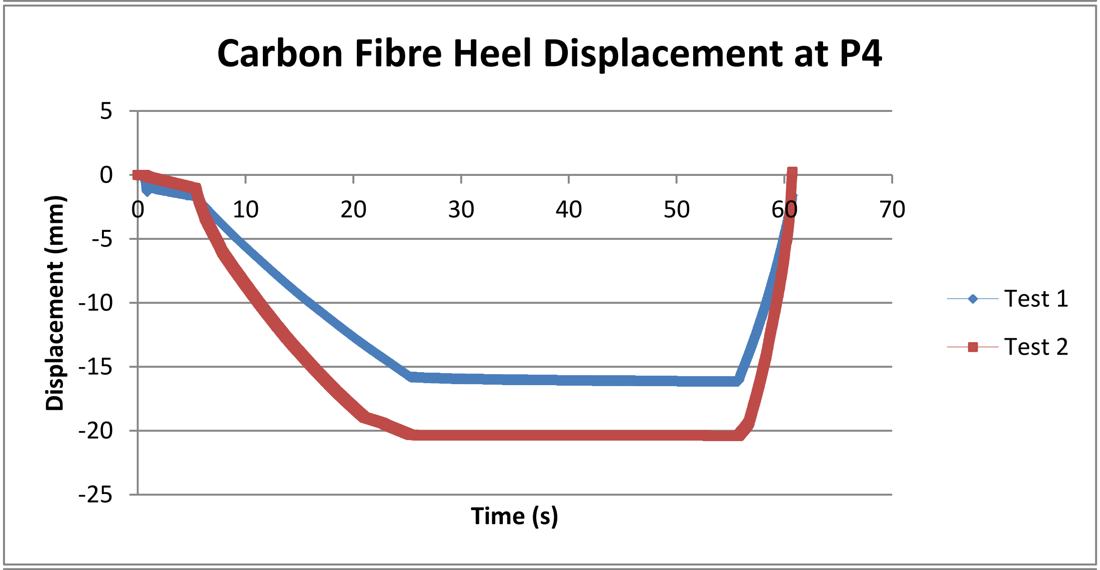


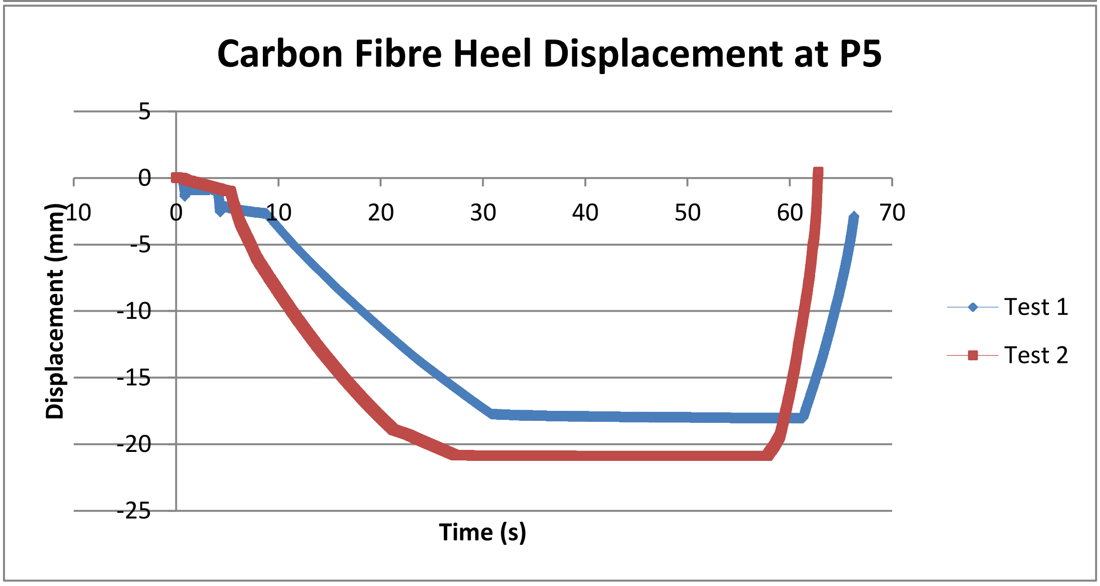

Carbon Fibre Forefoot at P5
Load (N)
Displacement (mm)
-40
-30
-20
-10
0
10
0
-500
-1000
-1500
-2000
-2500
Test 2

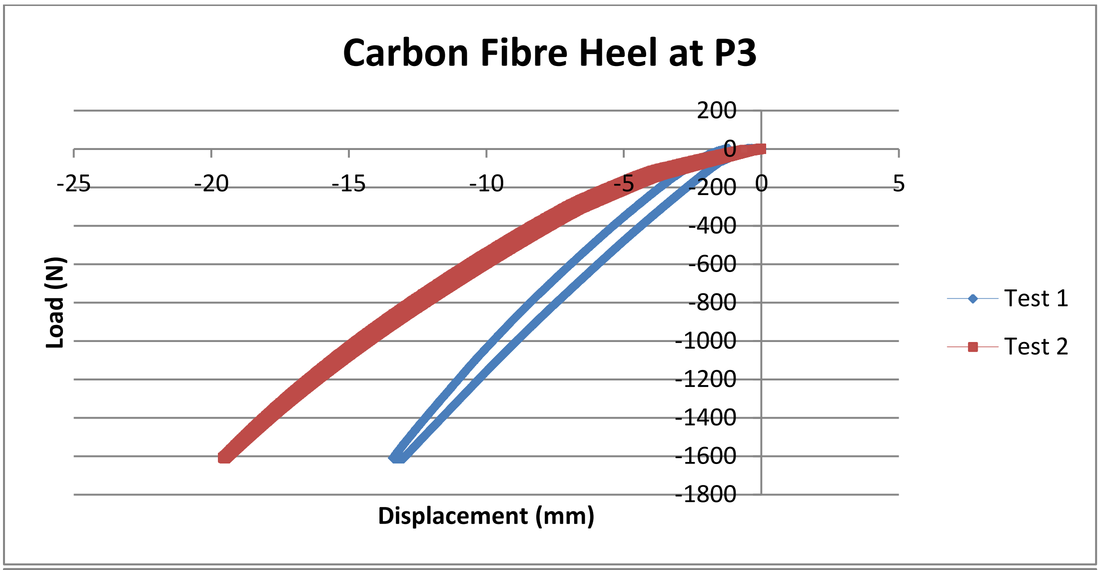

Carbon Fibre Heel Displacement Test 1
Displacement (mm)
Time (s)
5
0
-5
-10
-15
-20
-10
0
10
20
30
40
50
60
70
P3
P4
P5

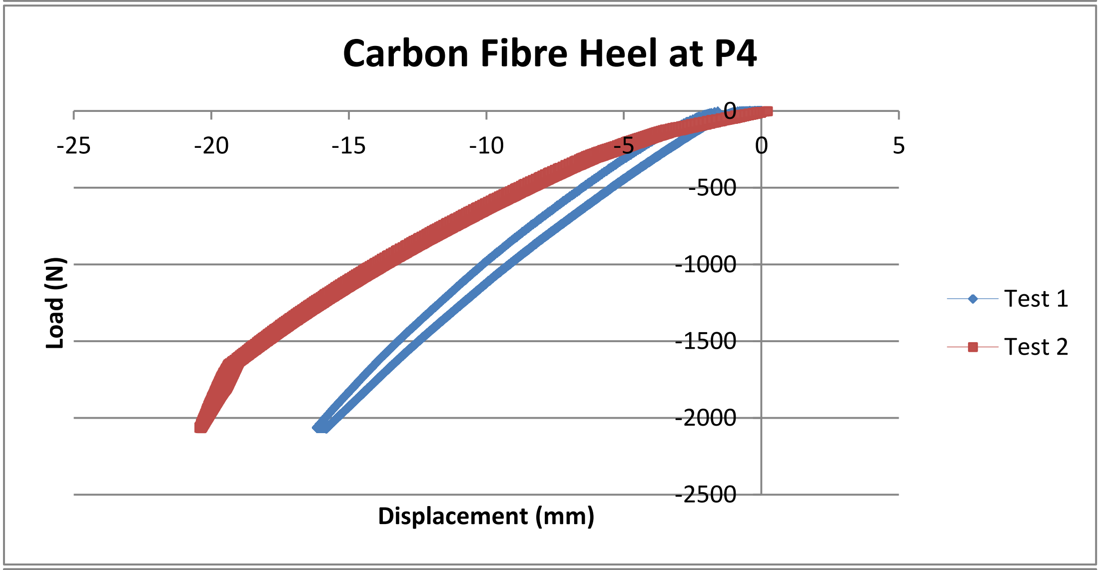

Carbon Fibre Heel Displacement Test 2
Displacement (mm)
Time (s)
5
0
-5
-10
-15
-20
-25
0
10
20
30
40
50
60
70
P3
P4
P5

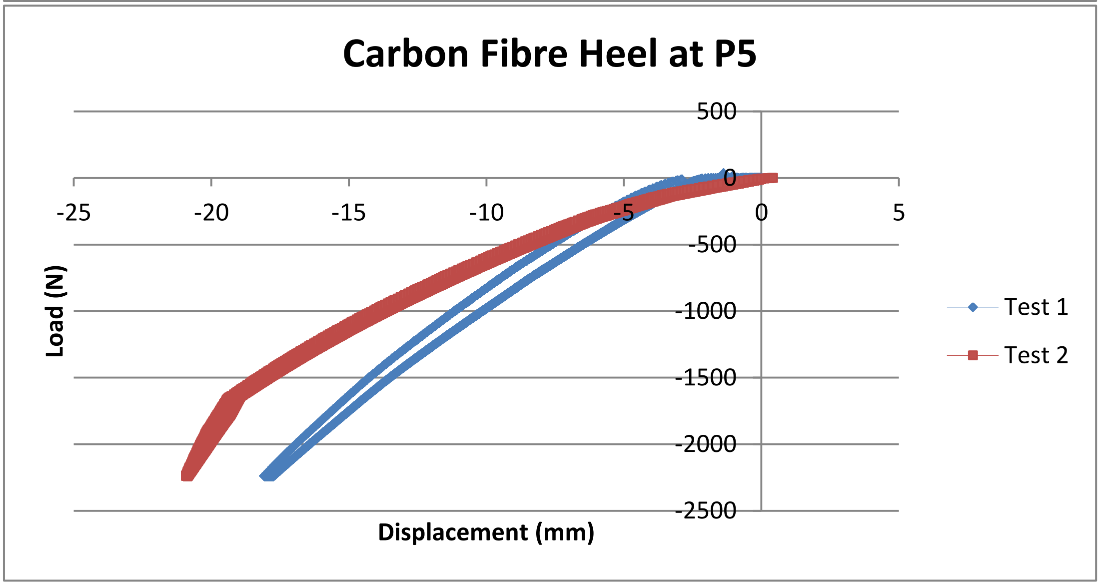

Carbon Fibre Heel Displacement at P3
Displacement (mm)
Time (s)
5
0
-5
-10
-15
-20
-25
0
10
20
30
40
50
60
Test 1
Test 2


Carbon Fibre Heel Displacement at P4
Displacement (mm)
Time (s)
5
0
-5
-10
-15
-20
-25
0
10
20
30
40
50
60
70
Test 1
Test 2

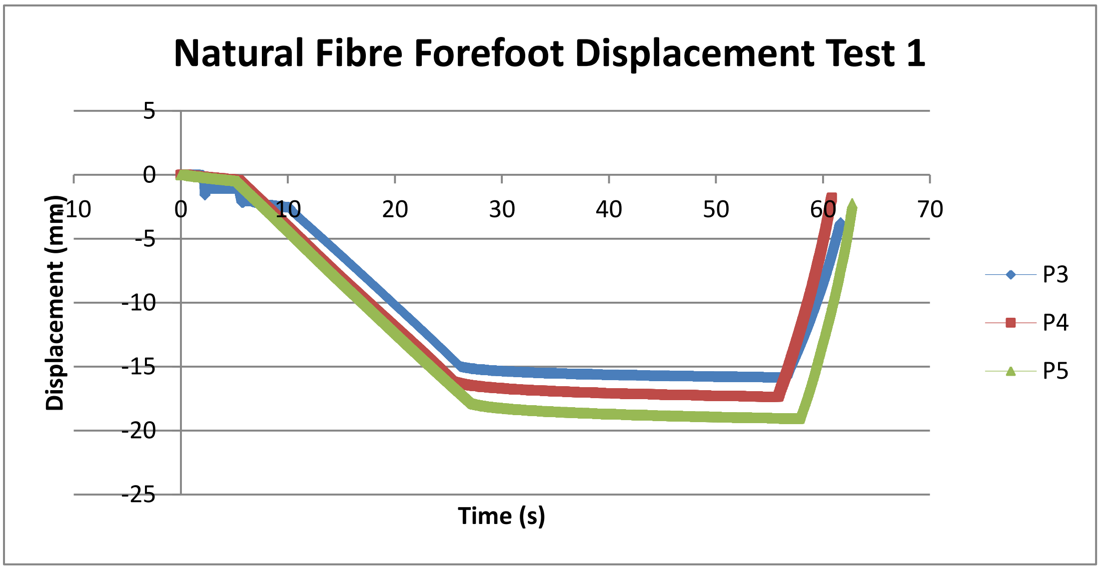

Carbon Fibre Heel Displacement at P5
Displacement (mm)
Time (s)
5
0
-5
-10
-15
-20
-25
-10
0
10
20
30
40
50
60
70
Test 1
Test 2

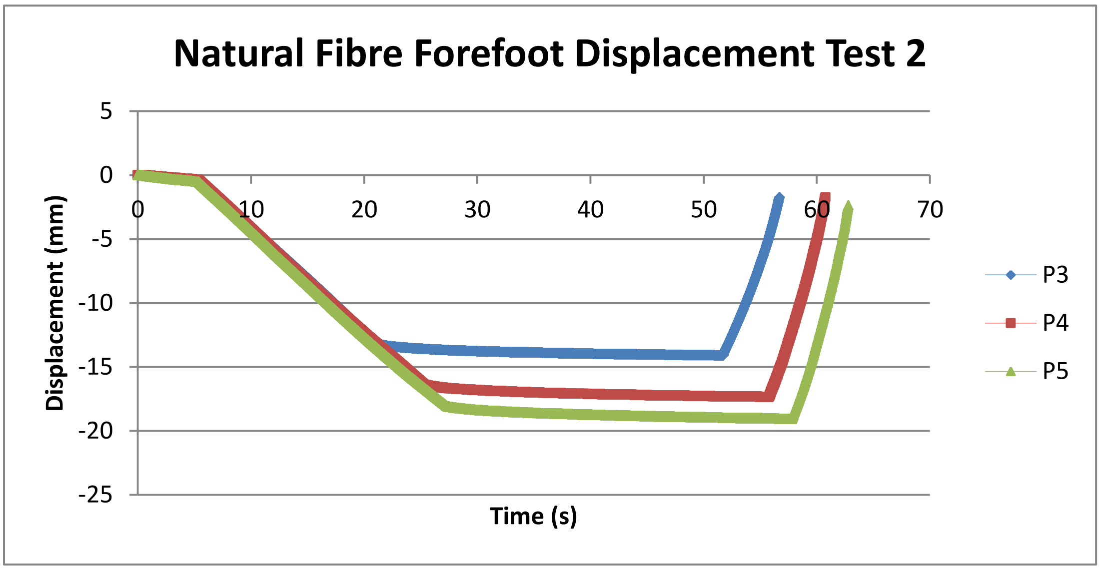

Carbon Fibre Heel at P3
Load (N)
Displacement (mm)
200
0
-200
-400
-600
-800
-1000
-1200
-1400
-1600
-1800
-25
-20
-15
-10
-5
0
5
Test 1
Test 2

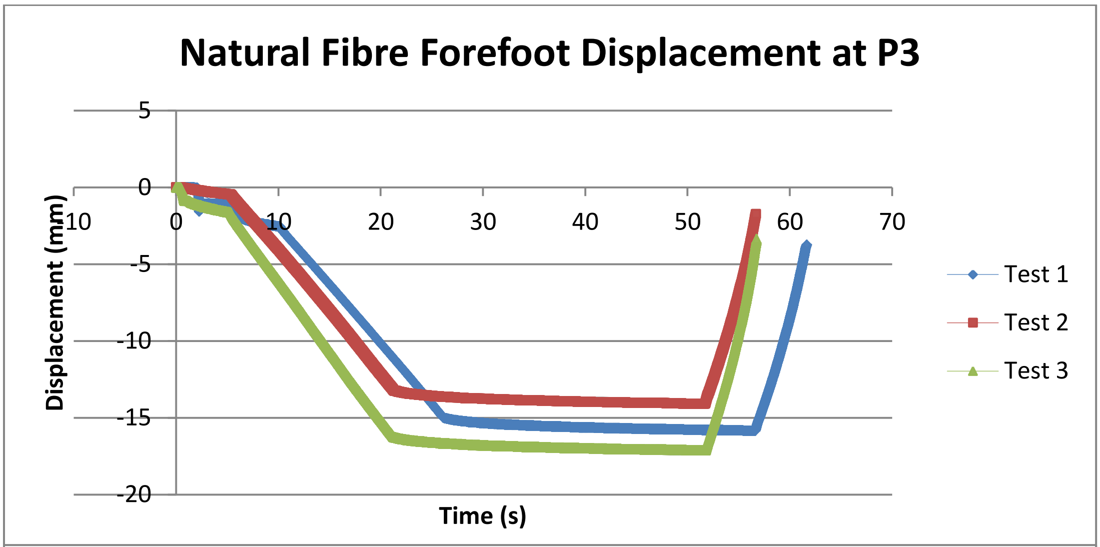

Carbon Fibre Heel at P4
Load (N)
Displacement (mm)
0
-500
-1000
-1500
-2000
-2500
-25
-20
-15
-10
-5
0
5
Test 1
Test 2

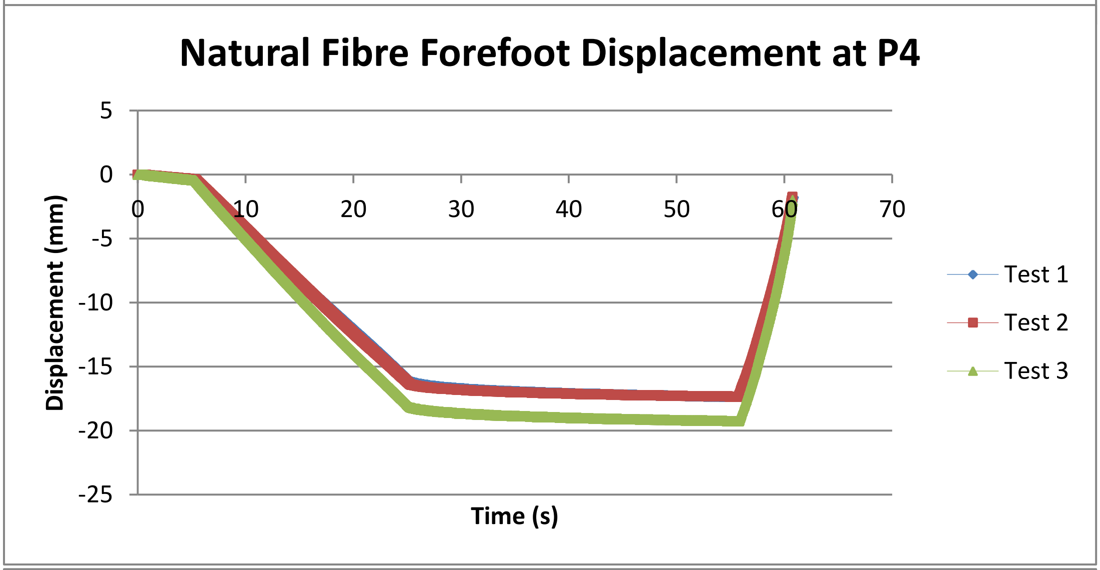

Carbon Fibre Heel at P5
Load (N)
Displacement (mm)
500
0
-500
-1000
-1500
-2000
-2500
-25
-20
-15
-10
-5
0
5
Test 1
Test 2

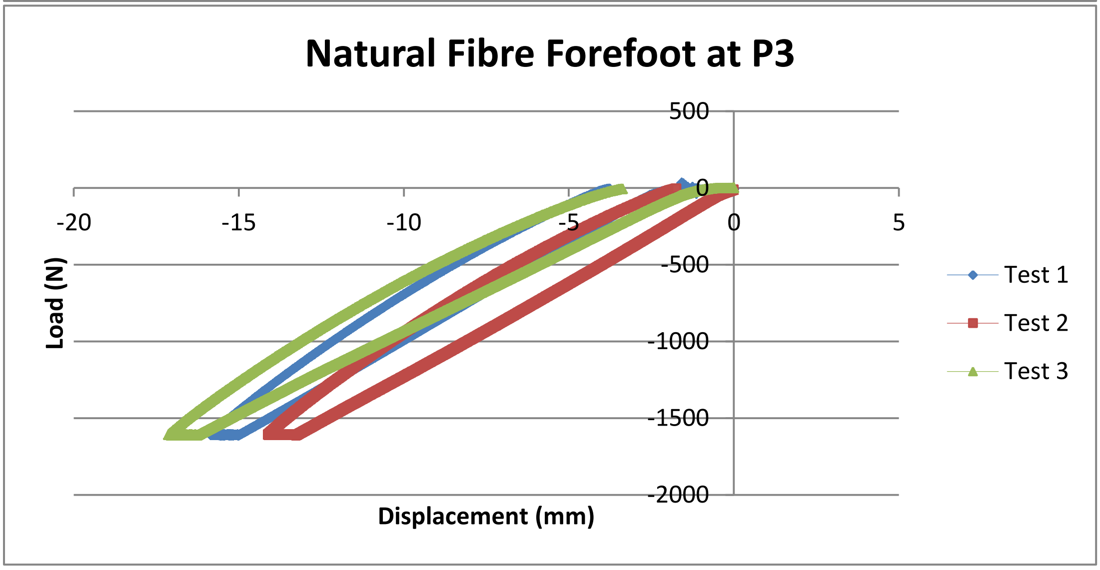
A.1.3 Natural Fibre Results

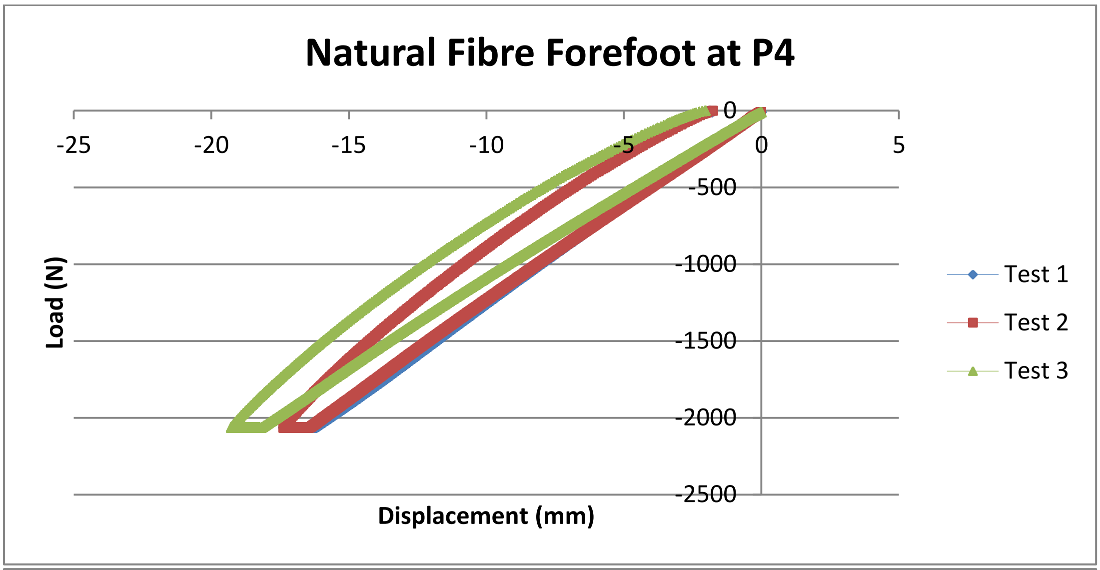


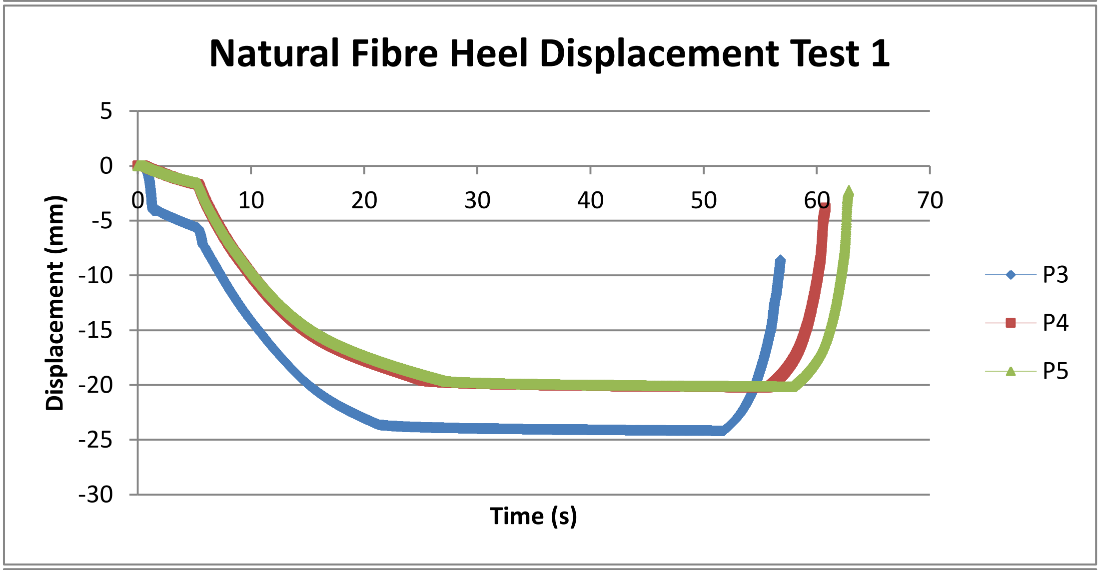


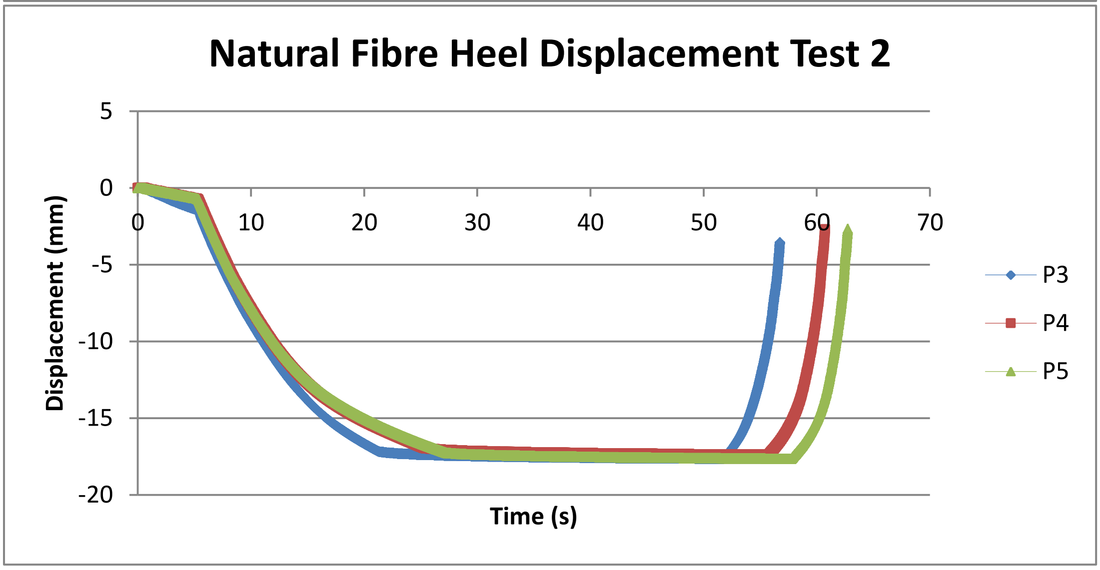

Natural Fibre Forefoot Displacement at P3
Displacement (mm)
Time (s)
5
0
-5
-10
-15
-20
-10
0
10
20
30
40
50
60
70
Test 1
Test 2
Test 3

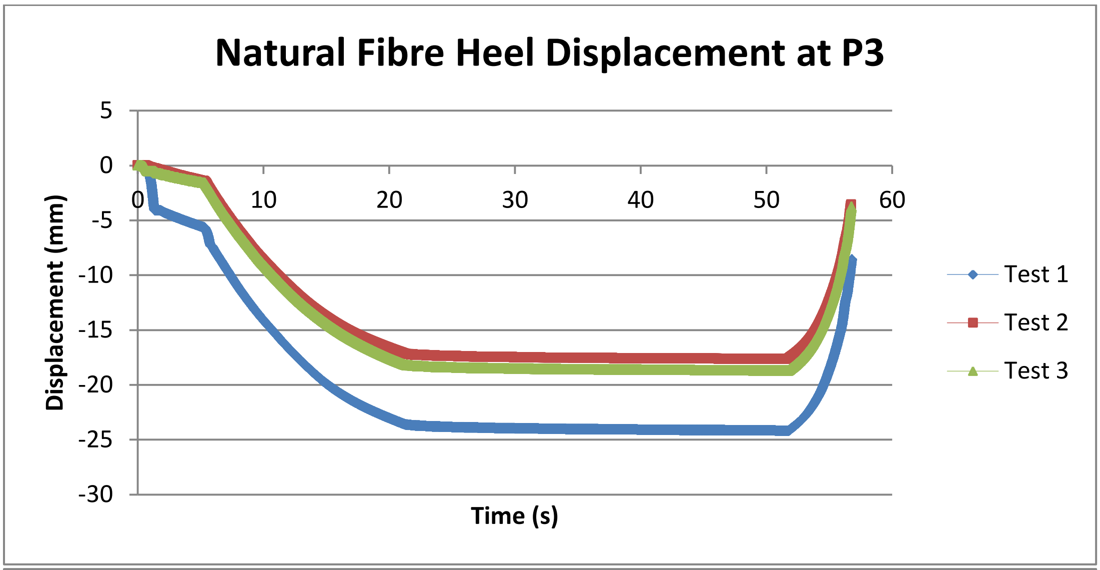

Natural Fibre Forefoot Displacement at P4
Displacement (mm)
Time (s)
5
0
-5
-10
-15
-20
-25
0
10
20
30
40
50
60
70
Test 1
Test 2
Test 3

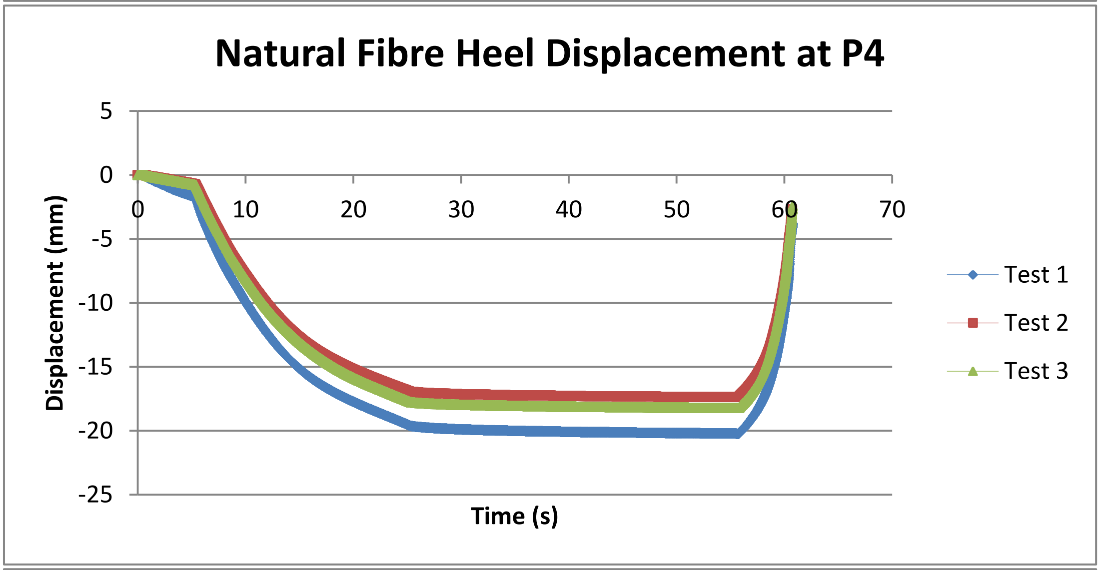

Natural Fibre Forefoot at P3
Load (N)
Displacement (mm)
500
0
-500
-1000
-1500
-2000
-20
-15
-10
-5
0
5
Test 1
Test 2
Test 3

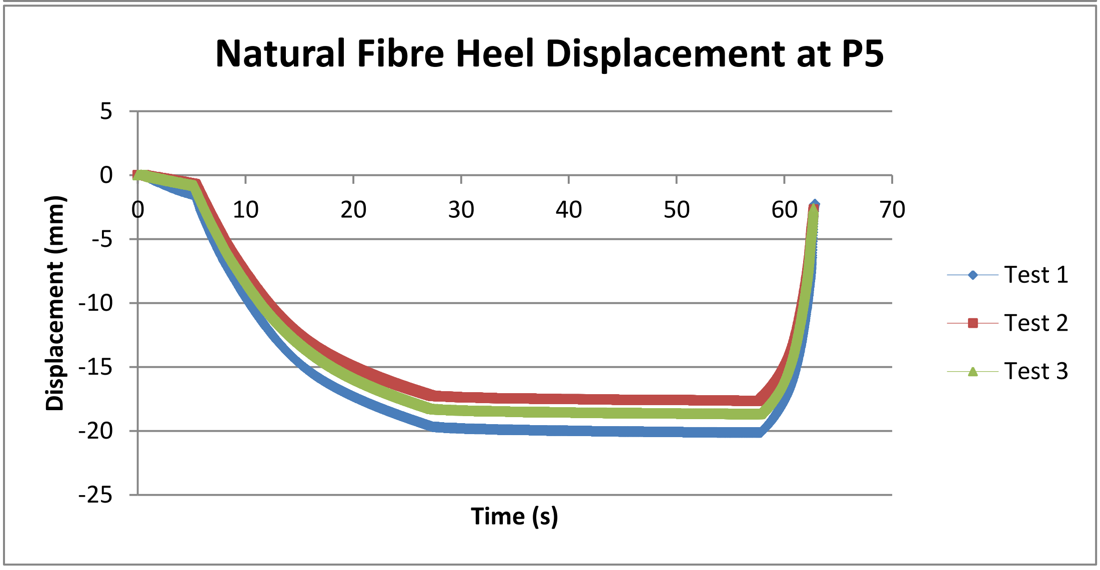

Natural Fibre Forefoot at P4
-25
-20
-15
-10
-5
0
5
0
-500
-1000
-1500
-2000
-2500
Load (N)
Displacement (mm)
Test 1
Test 2
Test 3

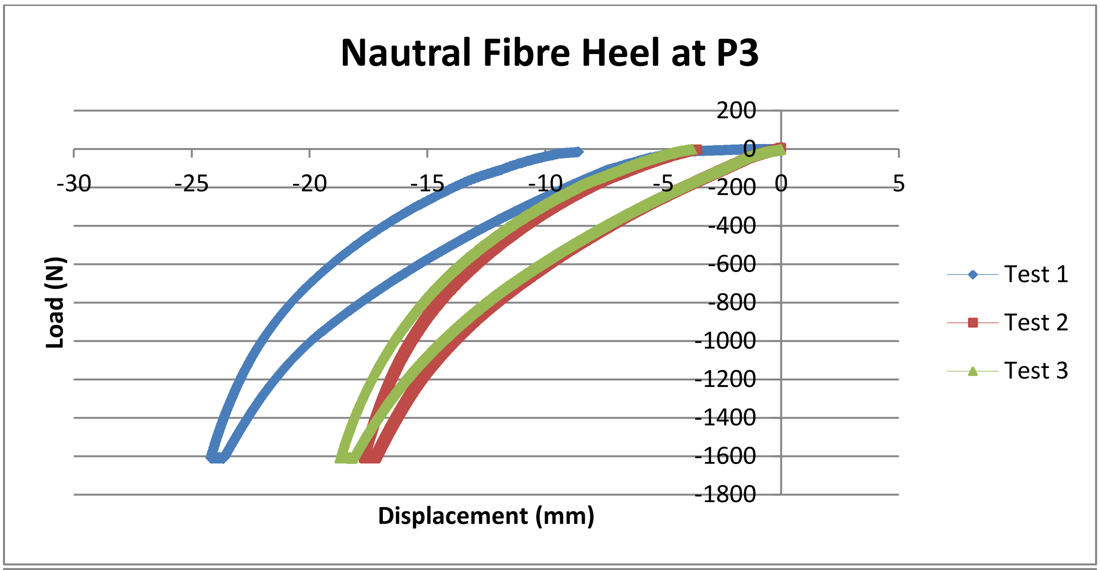

Natural Fibre Heel Displacement Test 1
5
0
-5
-10
-15
-20
-25
-30
0
10
20
30
40
50
60
70
Displacement (mm)
Time (s)
P3
P4
P5

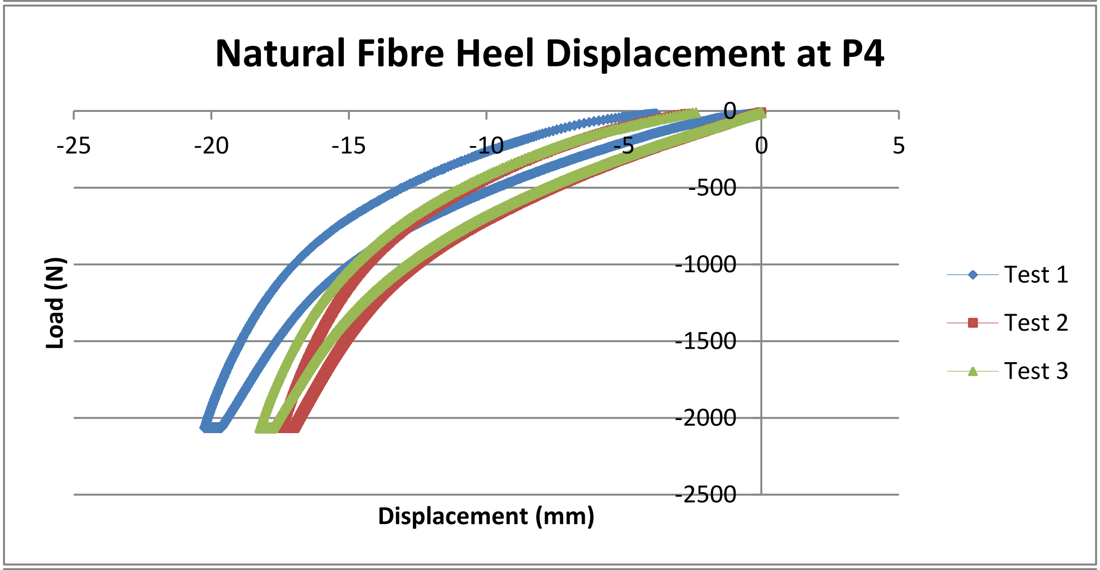

Natural Fibre Heel Displacement Test 2
5
0
-5
-10
-15
-20
0
10
20
30
40
50
60
70
Displacement (mm)
Time (s)
P3
P4
P5

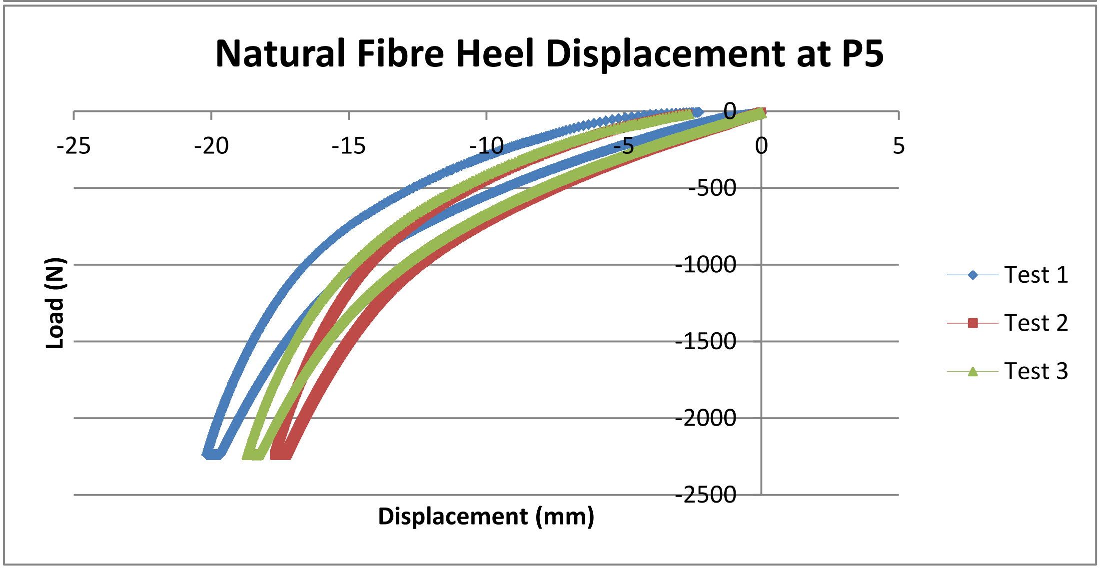

Natural Fibre Heel Displacement at P3
Displacement (mm)
5
0
-5
-10
-15
-20
-25
-30
0
10
20
30
40
50
60
Time (s)
Test 1
Test 2
Test 3


Natural Fibre Heel Displacement at P4
Displacement (mm)
5
0
-5
-10
-15
-20
-25
0
10
20
30
40
50
60
70
Time (s)
Test 1
Test 2
Test 3


Natural Fibre Heel Displacement at P5
Displacement (mm)
5
0
-5
-10
-15
-20
-25
0
10
20
30
40
50
60
70
Time (s)
Test 1
Test 2
Test 3


Nautral Fibre Heel at P3
200
0
-200
-400
-600
-800
-1000
-1200
-1400
-1600
-1800
-30
-25
-20
-15
-10
-5
0
5
Load (N)
Displacement (mm)
Test 1
Test 2
Test 3

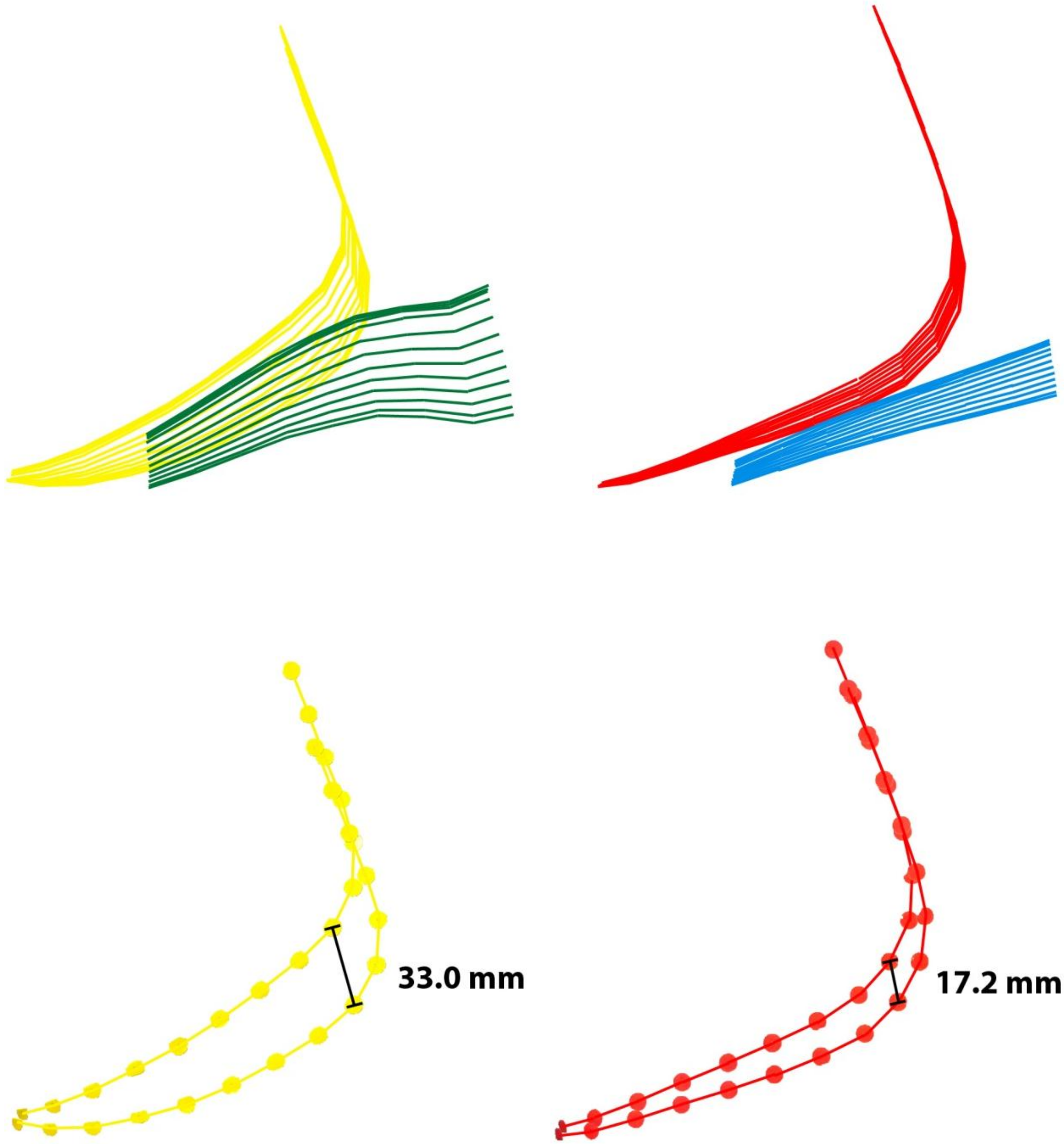

Natural Fibre Heel Displacement at P4
0
-500
-1000
-1500
-2000
-2500
-25
-20
-15
-10
-5
0
5
Load (N)
Displacement (mm)
Test 1
Test 2
Test 3


Natural Fibre Heel Displacement at P5
0
-500
-1000
-1500
-2000
-2500
-25
-20
-15
-10
-5
0
5
Load (N)
Displacement (mm)
Test 1
Test 2
Test 3

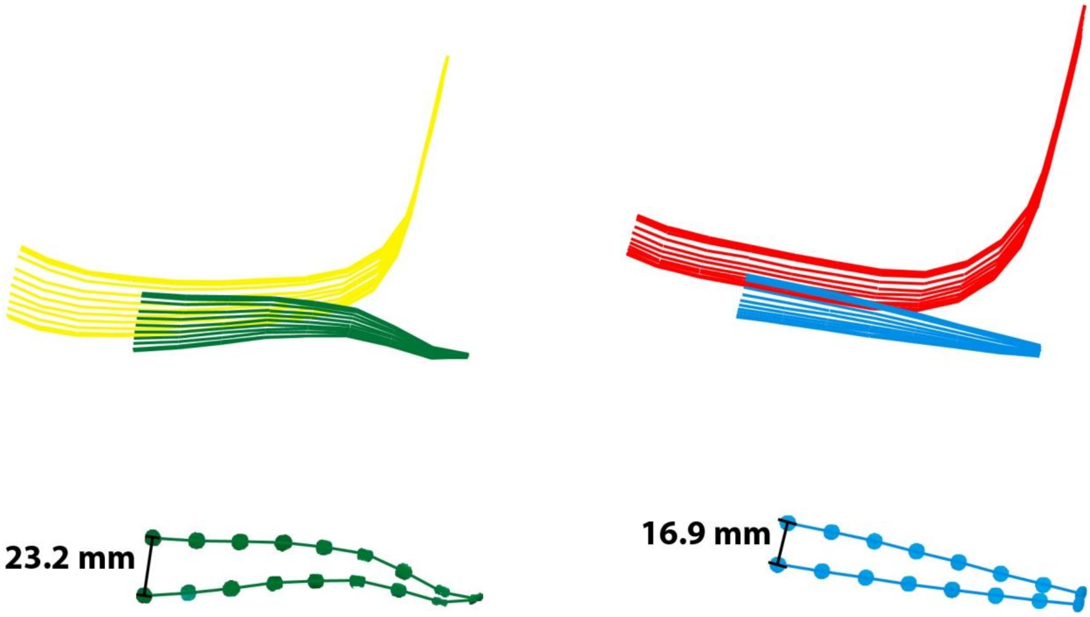
Appendix B

B.1 Deformation Tracking Results Continued

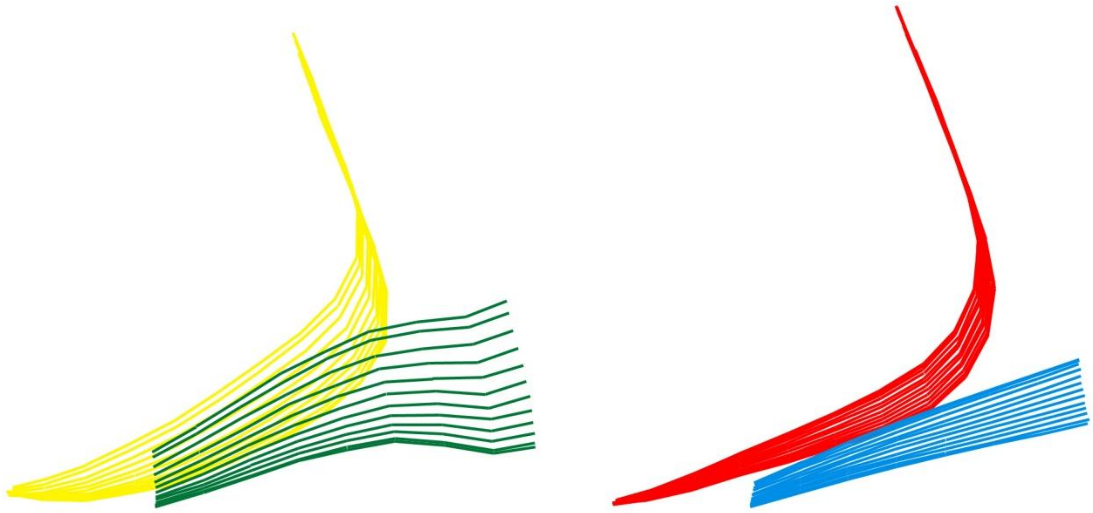
B.1.1 Forefoot Loading at Test Loading Level P3

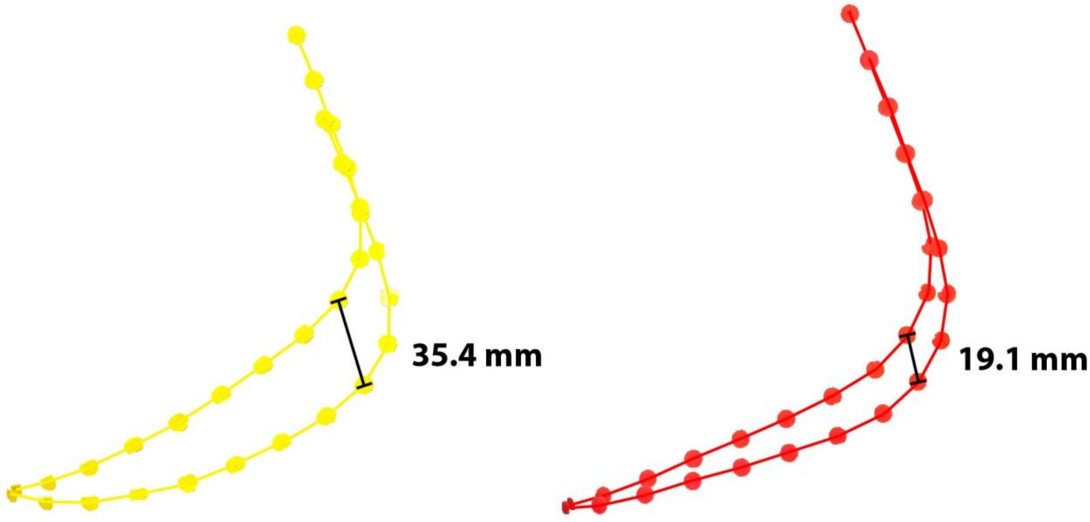


B.1.2 Heel Loading at Test Loading Level P3



B.1.3 Forefoot Loading at Test Loading Level P4





B.1.4 Heel Loading at Test Loading Level P4