Comparison of the Mechanical Properties of Different Wound Care Dressings Used in Lower Limb Arthroplasty.

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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.

Signed:

Date:

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ABSTRACT

The population of the UK is, on average, getting older and heavier. The result of this is an increased prevalence of osteoarthritis, and consequently, increased numbers of Total Knee Replacement (TKR) procedures carried out. On completion of TKR surgery, a large dressing is placed over the wound and adhered to the surrounding skin. With the wound being located over the joint, the dressing must have suitable mechanical properties to accommodate a wide range of movement at the joint. A dressing with unsuitable properties can cause many problems, the main one being blistering. The aim of this project is to test the mechanical properties of different orthopaedic dressings to assess their suitability and unsuitability as TKR dressings.

The Bose ElectroForce 3200 test instrument was used to carry out tensile tests on different dressing samples. Samples were strained up to 25% to mimic the wound strain experienced with 90° of knee flexion, and the stress response measured.

The results showed that AQUACEL dressing experienced the lowest stresses and has the lowest calculated Young's Modulus (0.327MPa), compared with Mepore dressing which has a calculated Young's Modulus of 16.33MPa.

It can be concluded from this investigation that from the dressings tested, AQUACEL exhibits the most suitable mechanical properties for the use on TKR wounds.

CONTENTS

1. INTRODUCTION

1.1 BACKGROUND

Total knee replacement (TKR) is a procedure where damaged bone surfaces within the knee joint are completely replaced with artificial components. This damage is usually a result of the degenerative disease, osteoarthritis, where the protective cartilage coating the ends of the bones becomes thinned, causing the individual severe pain. The prevalence of TKR procedures within the UK is relatively high, where 103, 925 procedures were carried out in 2014 in England and Wales alone (Njrcentre.org.uk, 2015). According to Arthritis Research UK, the risk of developing osteoarthritis increases in people aged in their late 40s. This, twinned with the knowledge that the UK is an ageing population suggests that the number of people suffering with osteoarthritis will likely increase with time. Arthritis Research UK predict that by 2035, the number of osteoarthritis sufferers will have increased by 26.4%. This large increase in patients will more than likely result in an increase of TKR procedures being carried out, thus placing more pressure on the National Health Service as well as private services.

There are problems generally associated with total knee replacements, such as infection or deep vein thrombosis (DVT) (Waheed and Dowd, 2013). An issue that is maybe overlooked, however, is problems caused by the post-operative wound dressing. An unsuitable dressing can have various detrimental effects on the patient, with a particularly common one being skin blistering (Johansson et al., 2012). The adhesive section of the dressing produces shear forces that act on the skin when the knee is flexed. With continued knee flexion, the shear forces acting on the skin will damage the skin surface, with the result being painful blisters. Some of the problems caused by the dressing, mainly blistering, are down to the mechanical properties of the dressing. If the mechanical properties do not satisfy the requirements of the wound and its surrounding skin, then the dressing may end up being detrimental, rather than beneficial, to the patient, potentially resulting in an increased hospital stay.

It has been found that a TKR wound increases in length by as much as 51mm when the knee is flexed to 90° (Dillon et al., 2007). It is therefore important that the dressing used to

cover the wound is capable of stretching to accommodate this change in wound length during knee flexion. It is also important that the dressing is able to stretch with as little load as possible, reducing the magnitude of shear forces acting between the dressing and the skin. Section 2.6 discusses the properties required for a dressing to provide optimal healing.

Determining the mechanical properties of orthopaedic wound dressings would be greatly beneficial in assessing whether a specific type of dressing is likely to reduce skin blistering when used on the wound from total knee replacement surgery.

1.2 OBJECTIVES

The objectives of this project are to:

Perform tensile testing on different orthopaedic dressings.

Quantify and compare the loads required to strain different dressings.

Calculate and compare the Young's Modulus among the dressings.

Look at the rate of stress relaxation among the dressings

Compare the findings in this project to previous studies, to determine if any of the dressings are an improvement over previous dressings, and whether their mechanical properties would suggest a potential reduction in the rate of dressing problems reported in patients.

2. LITERATURE REVIEW

2.1 THE KNEE

The knee is a very complex joint, consisting of multiple elements. It is one of the largest joints within the body and joins the largest bone in the body, the femur, to the bone that forms the shin, the tibia, hence the knee is also known as the tibiofemoral joint. Lateral to the tibia, is the slightly smaller bone, the fibula, and anterior to the distal end of the femur, lies the patella. This is more commonly known as the kneecap and is held in place by the tendon of the four muscles that form the quadriceps, and the patellar tendon, a continuation of the quadriceps tendon, which attaches to the tibia (Figure 2.1a). Between, and slightly superior to, the two condyles of the femur, lies a groove with which the patella articulates. This is known as the patellar surface (Figure 2.1b) (Gray, Warwick and Williams, 1973).

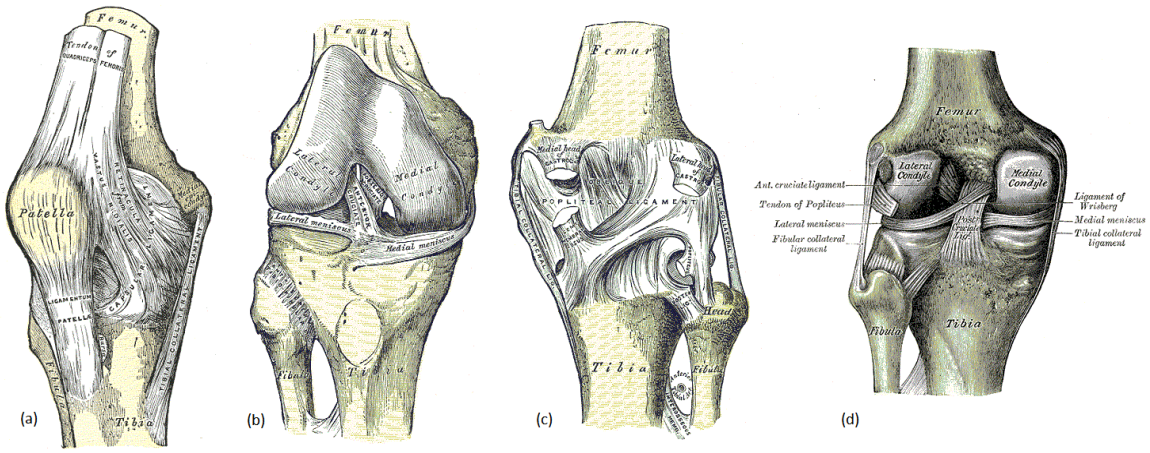

Figure 2.1: Anatomy of the knee. (a) Anterior view of right knee. (b) Anterior view of right knee, showing internal ligaments. (c) Posterior view of right knee. (d) Posterior view of left knee, showing internal ligaments. (Gray and Lewis, 2000)

There are many ligaments in the knee joint contributing to its complexity. Some of these include the Anterior Cruciate Ligament (ACL), Posterior Cruciate Ligament (PCL), Medial Collateral Ligament (MCL) and Lateral Collateral Ligament (LCL). The cruciate ligaments are so called as they form a 'cross' in the middle of the knee joint, between the femur and tibia (Figure 2.1d). These ligaments provide anterior and posterior stability for the femur and tibia, to prevent the bones sliding out of place. The collateral ligaments attach the femur to

the tibia, providing lateral and medial stability to the joint, also preventing the femur and tibia sliding out of place. The MCL prevents lateral rotation of the tibia about the knee, and the LCL prevents medial rotation of the tibia about the knee (Gray, Warwick and Williams, 1973).

Due to the location of the knee in the body, it bears large loads during normal daily activities; standing, walking and going up and down stairs. It has been found that resultant forces at the knee fall within the range of 220-350% of body weight (Kutzner et al., 2010). This puts the knee under a lot of stress with which the anatomy of the knee needs to accommodate. To protect the bones from damage, there is a thin layer of hyaline cartilage coating the ends of each bone, as seen in Figures 2.1b and 2.1c. This cartilage provides some element of shock absorption, as well as aiding smooth movement of the joint (Carticel.com, 2015). This is due to its frictional properties, where the coefficient of friction of this cartilage was found to be between 0.005 and 0.023, approximately 3 times less friction than sliding on ice (Charnley, 1960). To provide the knee joint with extra shock absorption, the menisci are a pair of semilunar fibrocartilages that attach to the head of the tibia and give the femoral condyles a concave pad with which to articulate with the tibia (Gray, Warwick and Williams, 1973). Fibrocartilage is both stronger and more durable than hyaline cartilage. It therefore plays a major role in shock absorption, especially when the load on the knee is increased, during a jump, for example.

The main actions of the knee joint are flexion and extension. These actions are mediated by the muscles within the thigh; the hamstrings and the quadriceps. When extending the leg, the quadriceps muscles contract, pulling the patella up towards the thigh. Since the patellar tendon is a continuation of the quadriceps tendon, the contraction of the muscles pulls the tibia anteriorly, extending the leg. The muscles also provide stability to the knee as well controlling movement of the joint. In a healthy knee, if the quadriceps are activated, an anterior shear force is produced, so the antagonistic hamstrings activate to counteract this force and stabilise the knee (Hortobágyi et al., 2005).

2.2 OSTEOARTHRITIS

2.2.1 WHAT IS IT?

Osteoarthritis is a degenerative disease that damages the joints of the body resulting in pain when the affected joints are used. There are many types of arthritis, each one affecting the body in different ways, but osteoarthritis is the most prevalent type of arthritis in the United Kingdom (Nhs.uk, 2015). In the UK alone, 8.75 million people have sought treatment for osteoarthritis, with higher numbers of women than men having sought treatment (Arthritisresearchuk.org, 2015). The knee is the most commonly affected joint, with approximately half of the total number of sufferers having osteoarthritis of the knee (Arthritisresearchuk.org, 2015). The risk of developing osteoarthritis starts to increase in people aged in their late 40s (Arthritisresearchuk.org, 2015), so with a currently ageing population, the numbers of people suffering from osteoarthritis will likely increase with time.

Osteoarthritis affects the cartilage within the knee and subsequently, the underlying bone surfaces. As a person gets older, the cartilage protecting the bones in the joint weakens, and begins to wear away (Niams.nih.gov, 2014). As this cartilage starts to wear away, its surface becomes rough and friction increases within the joint. Increased friction within the joint gives rise to an increased rate of cartilage wearing, making it thinner and thinner, eventually exposing bone. The body then responds and tries to repair the problem. In this situation, its response is to form new bone at the site of damage. Osteophytes, or bony spurs, start to grow in the area (Figure 2.2b), and with continued use of the damaged knee, the bony spurs will cause huge amounts of pain to the sufferer (Lozada, 2015). Figure 2.2 highlights the differences between a healthy knee and an osteoarthritic knee.

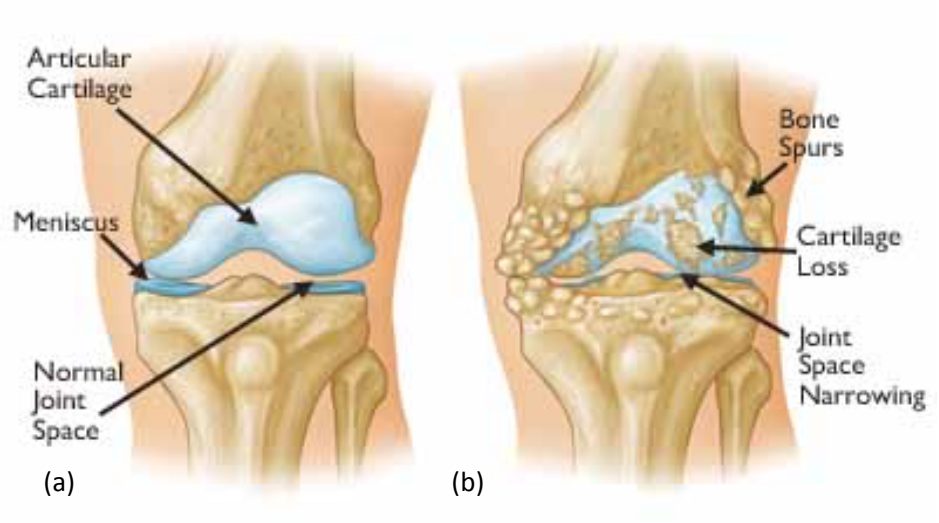

Figure 2.2: (a) Healthy knee joint. (b) Osteoarthritic knee joint (Orthoinfo.aaos.org, 2014)

2.2.2 INCREASED RISKS & SOLUTIONS

Unfortunately, there are factors that may increase the risk of someone developing osteoarthritis. A study carried out determined that females who participate in weight bearing sports, like running or tennis, over a prolonged period of time, have a 2 to 3 fold increased risk of developing osteoarthritis in the knees or hips than those who don't participate in the sports (Spector et al., 1996).

Obesity is a major factor in osteoarthritis, and many researchers have carried out studies looking into the reasons for this. The assumed hypothesis states that obesity increases the load on the joints, therefore increasing the risk of osteoarthritis. Griffin and Guilak, 2008, believe that there may be other factors of obesity contributing to osteoarthritis than just increased loads. They state that, "the ultimate influence of obesity on OA may involve a complex interaction of biomechanical and inflammatory factors" (Griffin and Guilak, 2008). Weight loss where and when possible is the obvious choice to help reduce the risk of developing osteoarthritis, or at least delay the onset of the disease.

These risks, alongside the knowledge that the UK is an ageing population, give rise to the possibility of osteoarthritis becoming even more prevalent than it is currently. Arthritis Research UK, have predicted the number of people who will be suffering from osteoarthritis in 2020 and 2035 (Figure 2.3).

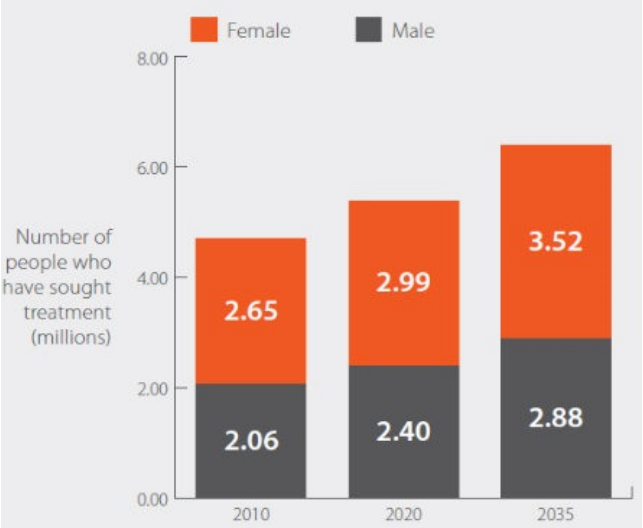

Figure 2.3: Number of osteoarthritis sufferers in 2010, predicted number of sufferers in 2020 and predicted number of sufferers in 2035 (Arthritisresearchuk.org, 2013).

Depending on the severity of the arthritis, different treatment options are available, both surgical and non-surgical. As mentioned previously, obesity is a major risk factor in osteoarthritis, so losing weight and reducing the load placed on the knee is an easy starting point for relieving symptoms of early stage arthritis. Another non-surgical option is medication, where anti-inflammatory drugs can be taken to reduce the inflammatory response within the joint. This, however, is a short-term method of treatment, as will only relieve symptoms for the duration that the drug is in the patients system.

The obvious choice of treatment would be surgery either to just remove small areas of damage and limit further damage, or to replace whole areas of the joint with artificial implants. Different surgeries have varying degrees of invasiveness with the most invasive treatment being a total knee replacement. An example of a minimally-invasive surgical option is arthroscopic chondroplasty. This option is used in early stage osteoarthritis and involves the smoothing of damaged cartilage within the joint to reduce frictional effects, allowing smoother movement of the knee. A slightly more invasive treatment is an Osteochondral Autograft Transplant (OATS) procedure. This involves the removal of a plug of bone where the cartilage has worn away, and the replacement with a plug of bone with healthy surface cartilage from another area of the body, ideally a non-weight bearing part of the knee (Sherry and Baer, 2011). If the area needing replaced is more than 20mm in size, bone and cartilage tissue must be taken from a cadaver; an allograft (Pamf.org, 2015).

2.3 TOTAL KNEE REPLACEMENT (TKR)

2.3.1 WHAT IS IT?

Total knee replacement is an invasive surgery, which completely replaces damaged surfaces within the knee joint. Three artificial surfaces are implanted in the knee, a femoral component, a tibial component and a patellar component. The aim is to give the patient a totally artificial, but fully functioning knee joint. The first knee replacement carried out in 1968, but many improvements since then, both methods and materials used for the implants, have significantly increased the success and effectiveness of the surgery (Orthoinfo.aaos.org, 2011). In an ideal situation where: the surgical procedure is carried out accurately, appropriate care is taken with the new joint and activity levels are modified, the knee replacement could last for 20 years. Ritter, 2009, found that there is a 98.6% survival of total condylar type implants 20 years post-surgery.

In 2014, the number of total knee replacements carried out in England and Wales alone was 103,925. 66.7% of those were carried out on the NHS and the remaining 33.3% were carried out privately (Njrcentre.org.uk, 2015). 84% of knee replacements were carried out on people aged over 60 in 2007/08 in Scotland (Arthro.scot.nhs.uk, 2014). 7169 knee replacements were carried out on the NHS in Scotland in 2013 (Arthro.scot.nhs.uk, 2014). Osteoarthritis accounts for over 94-97% of all TKR operations carried out worldwide (D. Van Manen, Nace and Mont, 2012)

2.3.2 REASON FOR SURGERY

Total knee replacement is a surgical treatment for severe osteoarthritis of the knee. It is only carried out if other treatment options have failed, if other treatment options are unsuitable, or if the patient displays certain symptoms, some of which are discussed below. On examination of the patients knees through radiography, if any of the following changes are observed, the patient will likely have to undergo TKR: Narrowing of the joint space between the femur and tibia, bone spurs forming on the end surfaces of the bones or if the condyles of the femur or tibia begin to appear as square (D. Van Manen, Nace and Mont, 2012).

If the patient exhibits any of the following symptoms, TKR is recommended (D. Van Manen, Nace and Mont, 2012):

- Severe knee pain, especially at night
- Difficulty carrying out daily tasks without pain
- Problems with mobility
- Knee deformity due to arthritis (Orthoinfo.aaos.org, 2011)
- No response to alternative, non-surgical treatments.

To undergo the surgery, the patient must conform with the following factors (D. Van Manen, Nace and Mont, 2012):

- No infection present
- Suitability for surgery must be addressed, ie. Age, weight, underlying health conditions
- Musculature within the thigh must be intact, allowing full participation in the rehabilitation process
- Patient must be aware of all benefits, risks and alternative treatment procedures available, other than TKR.

2.3.3 TYPES OF IMPLANT USED

Different types of implant used for different patients, and the surgeon carrying out the procedure will match an implant best suited to that individual patient. A typical knee implant has three components: a femoral component, a two-part tibial component, and a patellar component. These come together to form a new knee joint for the patient. Figure 2.4 shows a complete knee implant and its individual components.

The materials used for knee implants must be suitable for being implanted in the body, and must be capable of withstanding the internal biological environment and stress that the knee is put under every day. The most common materials used are titanium metal, cobalt chromium based alloys and polyethylene (BoneSmart®, 2015). Titanium is an inert biomaterial (BoneSmart®, 2015), meaning the material will not change its properties once implanted within the body. It is widely used for implantable devices due to its biocompatibility, but for knee implants cobalt chromium is utilised more often than titanium. This material has several features, including biocompatibility, resistance to

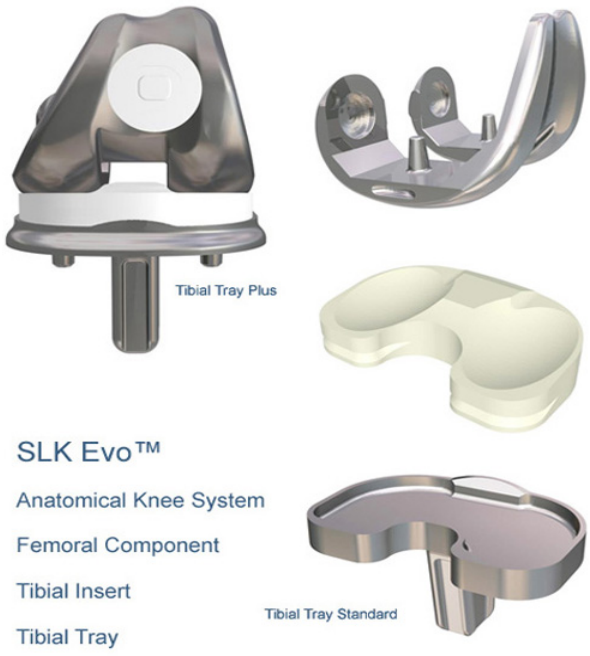

Figure 2.4: SLK Evo knee implant showing the different components making up the artificial knee joint (Implantsinternational.com, 2006)

corrosion and excellent toughness (BoneSmart®, 2015), making it an obvious choice for a knee implant. Titanium and cobalt chromium metals are used for the femoral component and one part of the tibial component of the knee implant. These are the parts that are put directly onto the bone, replacing the damaged surfaces. The other part of the tibial component is a plastic insert that provides the cushion and shock absorption of the menisci, as well as providing a smooth gliding surface for the metal femoral component to slide over. This plastic insert is usually made from polyethylene, either Ultra Highly Cross Linked Polyethylene or Ultra High Molecular Weight Polyethylene (UHMWPE). Navarro et al., 2008, state that UHMWPE is a very appealing choice for tibial inserts due to its optimal properties, such as low friction, high impact stress, low density and biocompatibility. This creates an artificial surface very similar to the native cartilage found within a healthy knee joint.

OXINIUM Oxidized zirconium is a newer material that has been in use since 2001 (BoneSmart®, 2015). It reduces the wear rate over standard cobalt chrome implants and the material exhibits improvements in hardness, smoothness and scratch resistance when

compared to the same standard implants: (Smith-nephew.com, 2015). When oxygen is absorbed into the material, the surface changes from metal to ceramic. This ceramic surface is 4900 time more abrasion resistant than the cobalt chrome, and therefore, reduces friction between the metal and plastic components (Smith-nephew.com, 2015).

There are three main types of knee implant, a fixed bearing implant, a mobile bearing implant or a posterior cruciate ligament (PCL) retaining/substituting implant. With the fixed bearing implant, the polyethylene tibial insert is fixed to the metal tibial component, prohibiting any movement between the two parts. The mobile bearing implant differs in that the tibial insert can rotate on the surface of the metal tibial component, giving the patient an extra few degrees of rotation at the knee. PCL retaining/substituting implants either have space to accommodate the PCL, a PCL retaining implant, or have a design feature to act in place of the PCL, a PCL substituting implant. This provides stability to the joint, which would be otherwise lacking without the PCL.

2.3.4 SURGICAL METHODS

There are two main surgical approaches to total knee replacement, one of which is classed as minimally invasive.

The medial parapatellar approach is the most recognised method of gaining access to the patients internal knee anatomy. This technique involves a large incision, 6-10 inches in length (Minitotalknee.com, 2009), spanning from the thigh, over the kneecap and part of the way down the shin. It also involves an incision of the quadriceps tendon (Mukherjee, Press and Hockings, 2009), which could potentially have detrimental effects on muscle action. The alternative, less invasive approach is termed the midvastus approach. To be deemed minimally invasive, the incision must be less than 14cm in length (Bonutti et al., 2004). Other than the size of the incision, there are benefits of the minimally invasive approach. Firstly, there is no incision in the quadriceps tendon (Mukherjee, Press and Hockings, 2009), instantly reducing the risk of muscular damage. White et al., 1999, carried out a clinical comparison between the parapatellar approach and the midvastus approach. It was found that in the first 8 days post-surgery, patients having undergone the minimally invasive surgery experienced less pain than those having undergone the parapatellar approach. There were also increased numbers of patients being able to straighten the leg 8

days post-operatively in the minimally invasive knees. White et al., 1999 also compared the two approaches 6 months after surgery and found that the same clinical parameters were equal between the two. This highlights that altering the surgical approach only improves the short term recovery, since the parameters were found to be equal within 6 months.

Surgery itself consists of: preparing the ends of the femur and tibia with a special bone cutting device (Stillwell, 1984), to create the correct shape for the metal implants to fit onto the bone, as exactly as possible; implanting and carefully aligning the new components to prevent the patient having a squint leg; and then securing the implants permanently into the leg.

2.4 REHABILITATION & ENHANCED RECOVERY PROGRAMME

Any patient having undergone a total knee replacement procedure will usually remain in hospital for between three and five days (Nhs.uk, 2014) to achieve the best start to their recovery as possible. On average, patients are discharged on the fourth day after surgery, with only 29 per cent of patients requiring further outpatient physiotherapy (Audit Scotland, 2010). The rehabilitation process will vary between patients, depending on their circumstances, so each patient will be assigned sets of exercises and tips to help them optimise their recovery time in the hospital. Patients who are fit and able will be encouraged to move around as much as possible on the same day as their surgery. If patients are showing promise, they may then be placed on an enhanced recovery programme, where they will likely be up and walking within the same day of surgery. The first week of post-operative rehabilitation can be crucial in the success of a patients recovery. Seventy per cent of patients are able to move around on the day of surgery (Audit Scotland, 2010).

Studies indicate that the numbers of patients on the enhanced recovery programme who are mobile within 24 hours of lower limb arthroplasty, are higher than patients who are not on the enhanced recovery programme. These studies also found that patients on the enhanced recovery programme experienced decreased length of hospital stays, post-operatively (McDonald et al., 2012), (Scott et al., 2012).

There are various factors in regard to wound healing that will be affected by the enhanced recovery programme, where the demands of the wound, and the dressing need to match the demands of the rehabilitation. The demands of the rehabilitation, however, are inevitably too high for the demands of the dressing, such that the dressings can then cause additional problems for the patient. Some of the problems caused by the dressing are discussed in section 2.5.1.

2.5 THE WOUND & WOUND DRESSINGS

TKR incisions are commonly closed using either sutures or staples. These methods both provide a watertight wound, as well as a cosmetically acceptable scar (Newman et al., 2011). However, it has been found that on comparison, suturing a TKR incision significantly increases wound complications over stapling (Newman et al., 2011). Hlubek et al., 2014, state that the use of "continuous suture technique may produce skin strangulation at wound margins, which will complicate healing." They then state that the use of staples to close TKR incisions minimises the risk of this skin strangulation, however, few studies on the comparison of suturing and stapling have been published (Hlubek et al., 2014).

On completion of total knee replacement surgery, an appropriate dressing is placed over the wound, consisting of an absorptive central section and an adhesive border, which adheres to the skin surrounding the wound. The purpose of a dressing is to maintain a suitable level of moisture for the wound, prevent bacteria from infiltrating the site of the wound, causing infection, and to absorb any fluids that may leak from the wound as it goes through the healing process (NICE Clinical Guidelines, No. 74). Due to the location of the wound, being the knee, there is a wide range of movement with which the dressing must accommodate. This often is not the case, with the result being additional pain and extra problems for the patient.

There are different types of dressings, ranging from traditional, to novel. Each dressing will vary in its mechanical properties, which will alter the effect that each dressing has on the skin it is adhered to. Some dressings result in more problems than others, but various types of dressing are described below.

Traditional adhesive: This dressing has limited stretch capabilities and has a high incidence of painful blistering on the periwound area when worn over the knee area (Ravenscroft, Harker and Buch, 2006). Dillon et al., 2007, found that for less than 5% strain of the dressing, the loads required to achieve this small amount of strain were very high, compared to other dressings tested (Figure 2.5). This would result in excess strain and friction on the skin, creating large shear forces and subsequent skin blistering (Ravenscroft, Harker and Buch, 2006).

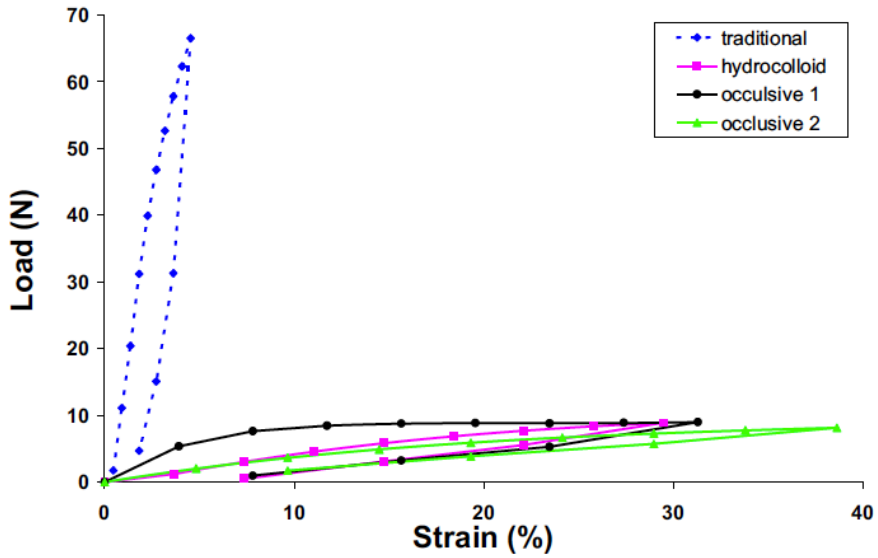

Figure 2.5: Loads required to achieve certain strains in four different dressings (Dillon et al., 2007)

Occlusive: This is the general term for various different dressings that form an airtight seal with the skin surrounding the wound, meaning the wound is completely sealed of oxygen, water and bacteria. There is however, nothing built in to the dressing to absorb fluid leakage from the wound.

Hydrocolloid (DuoDERM® Extra Thin): An adhesive dressing made from natural or synthetic polymers (Purser, 2010). The absorbant section of this dressing is separate from the adhesive section, and is placed on the wound prior to the hydrocolloid. The occlusive nature of the hydrocolloid dressing provides a complete seal from external factors like bacteria, thus reducing the risk of infection, or water, meaning the dressing is waterproof. This allows the patient to bathe and shower normally whilst still wearing their dressing.

Hydrocolloid with Hydrofiber technology (AQUACEL® Ag Surgical): This dressing consists of a hydrocolloid adhesive and a Hydrofiber absorbent section. These two sections are bonded during manufacture, unlike the DuoDERM Hydrocolloid. The hydrofiber section is made up of 100% sodium carboxymethyl-cellulose, which converts to a gel when it comes into contact with any fluid exuded from a wound (AQUACEL™ Hydrofiber Wound Dressing, 2009), keeping the wound surroundings moist for optimal healing (Ravenscroft, Harker and Buch, 2006), while retaining wound exudates by vertical absorption (G. Richetta et al., 2011). This type of dressing has also shown decreased adhesion to the wound itself (Purser, 2010), potentially a factor in the reduction of wound healing time.

"Hydrofiber dressings absorb up to 30 times their weight and provide less risk of maceration because of their vertical fluid absorption properties" (Gibbs, no date.). This highly absorptive nature of hydrofibre dressings makes them suitable for heavily exuding wounds (Purser, 2010).

"The hydrofiber/hydrocolloid dressing combination showed significant clinical improvement compared to the adhesive dressing" (Burke et al., 2012).

2.5.1 PROBLEMS ASSOCIATED

The dressing used to cover a TKR wound plays a major role in the healing process. The effects each dressing has on the skin will vary between patients. This will be down to skin changes between patients. Older skin is less elastic than younger skin (Lifshitz and Tomecki, 2015), male skin may vary from female skin, and other factors, such as medication, may affect the patients skin and the healing ability of the wound (Ousey, Gillibrand and Stephenson, 2011). Different dressings can have different adverse effects on the wound and surrounding skin. These adverse effects can prolong the healing process and potentially increase the risk of infection. Some of these problems are discussed below, where all information is taken from the American Journal of Nursing (Bryant, 1988).

SKIN STRIPPING

Due to the adhesive nature of the wound dressings used for TKR patients, problems can arise if the adhesive is too strong, or if the dressing is taken off and replaced multiple times on the same section of skin. Skin stripping is a 'superficial injury' where the top layer of the skin surface is damaged or even removed by the adhesive parts of the dressing. Figure 2.6

shows electron micrographs of the damage to the skin surface after three different dressings were removed from the skin (Waring, Bielfeldt and Brandt, 2009).

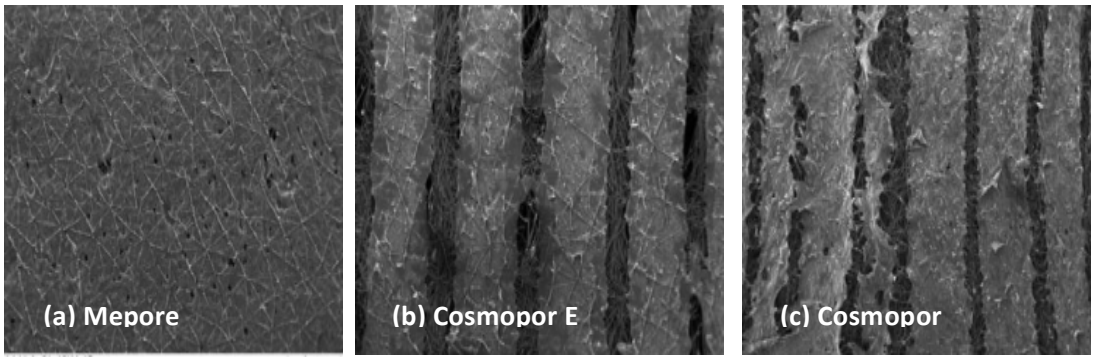

Figure 2.6: Samples of electron micrograph images showing damage to the skin surface after the removal of three different dressings
(Waring, Bielfeldt and Brandt, 2009)

CHEMICAL INJURY

This occurs when a chemical becomes 'trapped' between the skin and the adhesive part of the dressing. If the skin is in contact with this chemical for an extended period of time, irritation will occur.

TENSION BLISTERS

Dressings are normally applied under tension to create an area of compression around the wound. Blistering can occur on the skin surface when the dressing is applied under too much tension and the tension in the dressing is then transferred to the skin, which, if not changed often enough, will result in blistering, similar to those seen in Figure 2.7.

FOLLICULITIS

Folliculitis is an inflammation of the hair follicles. This can occur if the hair on the skin surface is shaved using a razor, to allow the dressing to adhere to the skin, or if a chemical becomes 'trapped' in the hair follicle itself. If adhesive from the dressing is transferred to the skin, it could also play a part in causing folliculitis.

MACERATION

This can also be referred to as 'skin pruning' where the resultant skin resembles the wrinkly surface of a prune. When the skin underneath the dressing becomes saturated with moisture, it becomes wrinkly and can lose structural integrity.

ALLERGIC REACTION

This is the least common problem caused by adhesive wound dressings, but can vary in severity from person to person. An allergic reaction is the result of skin sensitisation to a specific element within the adhesive portion of the dressing. The length of time that the skin is exposed to the specific allergen determines the severity of the allergic reaction, where increased exposure will likely increase the severity of the reaction.

BLISTERING

Skin blistering is a very common post-operative problem (Johansson et al., 2012). The blisters caused by inappropriate dressings can vary in size and severity. It can be seen in Figure 2.5, that to strain the traditional dressing by 5%, hugely high loads are required compared to the other dressings tested in the study. This limit of stretchiness within the dressing is then transferred to the skin, as a shear force, where it is pulled with the dressing as the knee joint moves, for instance.

"Increased friction and/or tension at the interface between the skin and the wound dressing creates shear forces, loosening the connections between the epidermis and dermis, which causes the separation of the two skin layers and allows interstitial fluid to seep into the newly created space to form blisters." (Johansson et al., 2012)

An example of this type of blister can be seen in Figure 2.7.

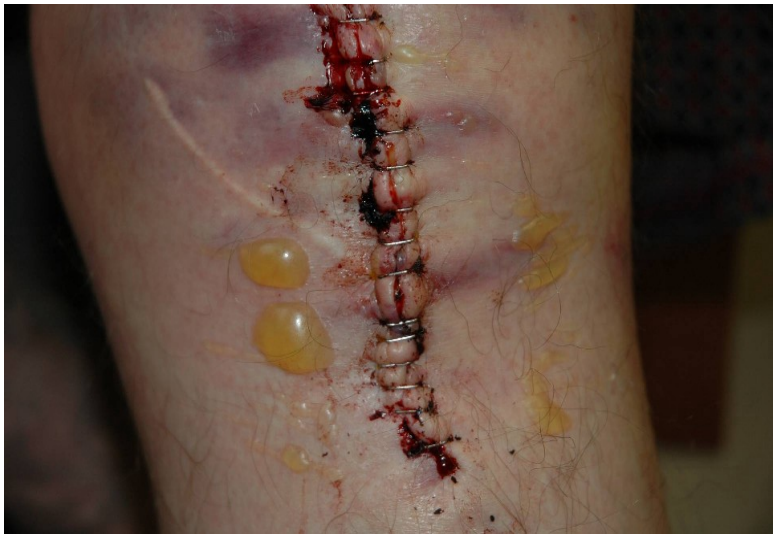

Figure 2.7: Photo of blistering caused by an adhesive dressing

Various studies have been carried out looking into the occurrence and rate of blister formation with different post-operative wound dressings.

Cosker et al., 2005 looked into the occurrence of blisters, the presence of infection and the number of dressing changed required with three different types of dressing (Primapore™, Tegaderm™ with pad and OPSITE™ PostOp). Three hundred orthopaedic patients were included in this study, and the three dressings were assigned randomly, but equally, to the patients. It was found that the OPSITE™ PostOp produced the lowest rate of blistering with only 6% of subjects showing blisters, compared to 16% with Tegaderm™ and 24% with Primapore™. It was also observed that patients using the OPSITE™ PostOp dressing and lower levels of wound exudate when compared to the other two dressings. They additionally stated that a dressing with a clear film and a high vapour transmission rate results in reduced blistering and decreased wound discharge (Cosker et al., 2005).

Another study (Bhattacharyya et al., no date) looking into the prevalence of blisters focused on patients having undergone minimally invasive arthroscopy of the knee. The patients in this study had much smaller incisions than the standard total knee replacement incision, but since the dressings are still being applied around the knee, the dressings are still required to be flexible enough to accommodate large amounts of knee flexion. Bhattacharyya et al., compared Smith & Nephew's OPSITE™ PostOp dressing with Mölnlycke's Mepore® dressing for blister formation. 116 patients were involved in this study, half were given OPSITE™ PostOp and half were given Mepore®. The dressings were adhered along the longitudinal axis of the leg whilst making sure no tensile force was created and a layer of wool and crepe bandage was then applied to the leg. No blisters were observed in the OPSITE™ PostOp group, whereas 6% of the Mepore® group developed blisters (Bhattachayya et al., no date).

The ability of an orthopaedic dressing to stretch is a very important factor in the wound care regime. With joint surgery, the resulting wound will likely lie over the joint where the surgery has been carried out. This means that as the joint moves, the wound will also move. Knowing this, it is important that the dressing selected has enough stretch and flexibility to accommodate this movement. Koval et al., 2003, looked into the effect a

stretchy and non-stretchy dressing has on the skin; the findings of which are discussed below.

Koval et al., 2003, studied the prevalence of blistering in hip surgery patients where a non-stretchy dressing was used, compared to where a stretchy dressing was used. One hundred hips in ninety nine patients were studied with the application of either a non-stretch silk tape, or a perforated stretch cloth tape. Each dressing was applied along the longitudinal axis of the leg, with the hip in extension, to avoid producing any tensile forces within the dressing. It was observed that there was a 41% risk of developing blisters with the non-stretch silk tape, compared to a 10% risk of blister formation with the perforated stretch cloth tape. It was stated that the resistance to stretch in the silk tape is a major contributor to the formation of skin blisters. This is due to the increase in shear forces acting on the skin at the ends of the tape, when the hip is both flexing and extending, creating constantly changing tensile forces within the tape dressing (Koval et al., 2003).

It is crucial that the correct dressing is selected for TKR wounds. Wound problems are a major issue, which can slow down the rehabilitation process, and ultimately could result in patients spending longer in hospital.

2.6 IDEAL PROPERTIES

The properties of a wound dressing are very important in the healing of a wound. Having suitable biological as well as mechanical properties is crucial to enable the dressing to provide optimal healing. The following list of ideal properties is taken and adapted from Chapter 6 of Surgical Dressings and Wound Management (Thomas, 2010).

The dressing must keep the wound and surrounding skin in an optimal state of hydration. A moist wound environment promotes wound healing, and occlusive dressings have been proven to contribute to faster acute wound healing.

In addition to keeping the wound well hydrated, it is important to for the dressing to maintain the wound at an optimal temperature and pH, as these factors can have a large impact, both positively and negatively, on the healing process.

The dressing should provide protection to the periwound skin from potentially irritant wound exudate and excess moisture. Whilst keeping the wound in an optimal state of hydration is very important, excess moisture can hinder the healing process, therefore extending the healing time.

Ideally the dressing will form an airtight, and water-resistant seal to the wound, forming an effective bacterial barrier to reduce the risk of infection, but also remains easily removable, causing minimal trauma to the skin surface. This airtight and water-resistant seal also allows the patient to bathe normally whilst wearing the dressing, an activity that was not possible with older, non water-resistant dressings.

A suitable dressing will conform well to the wound and the respective limb. This is important for the patients post-operative comfort, since rehabilitation can begin as early as the same day of surgery, meaning the dressing must be suitable for mobility shortly after application.

The application, wearing, and removal of the dressing will ideally be as pain free as possible for the patient, as any pain in addition to the pain from the surgery will be unwelcome.

The dressing should be free of toxic or irritant substances that could result in major complications for the patient, such as infection or a painful skin irritation.

It is very important for a TKR dressing to have suitable stretch capabilities to allow the joint to flex and extend smoothly, without the dressing hindering this movement. Ideally, the dressing will create minimal shear force on the skin, reducing the risk of blister formation; a problem that is very much a result of poor dressing selection.

2.7 RESEARCH SUMMARY

The studies discussed in this chapter provide evidence to show that there are many complications that can arise following orthopaedic surgery, and that the choice of dressing used for post-operative wounds can have a huge impact on both the healing of the wound and the patient rehabilitation regime.

Many problems are associated with the dressing alone, but it can be seen from the reviewed literature that the most reported problem is skin blistering. This can prolong the patients recovery and cause them serious discomfort in addition to any post-operative pain. This can, therefore, result in hospital stays being extended, which ultimately increases the costs; a significant disadvantage with the entire TKR procedure.

There are very few studies characterising the mechanical properties of orthopaedic wound dressings. This is maybe an aspect that is overlooked slightly, since the majority of research carried out on wound dressings is clinical. As important as the clinical research is, by carrying out mechanical testing prior to the clinical testing, the use of mechanically unsuitable dressings could be avoided. This, ideally, would reduce the problems caused by dressings that are unsuitable for the demands of the wound and the location of that wound.

3. METHODOLOGY

3.1 INITIAL EXPERIMENT

3.1.1 INSTRON ELECTROPULS E10000

Figure 3.1: Instron ElectroPuls E10000

The Instron ElectroPuls E10000 is an "all-electric test instrument designed for dynamic and static testing on a wide range of materials and components" (ElectroPuls™ E10000 Linear-Torsion All-Electric Dynamic Test Instrument, 2014). It carries out tension and compression tests, as well as torsion tests. The Instron has two mechanical grips, as seen in Figure 3.1, which feature serrated gripping surfaces. This provides a high friction environment, limiting the risk of the sample slipping in the grips during testing. The opening and closing of these two grips is controlled by compressed air, powered from the mains supply. The linear load capacity of the Instron is 10000N (10kN), with a 10kN load cell, and for smaller load measurements, a more sensitive 1kN load cell can be used instead. The load cell can be placed in either the upper or lower position, depending on personal preference. The lower

position is fixed vertically, so cannot move up and down. This creates a large range of distance between the two grips. The whole top section of the Instron can be moved up and down the two pole frame, via the large screws on either side of the instrument (Figure 3.1). This changes the distance between the grips, meaning a large variety of sample sizes can be tested in the Instron. The upper crosshead has a total vertical movement capacity of 60mm, which is split into +30mm and -30mm, and this vertical movement is controlled by a linear electric motor within the crosshead. This is essentially a large electromagnet where at 0 Volts, the grip is in the middle position, classed as 0mm. If a positive voltage is applied, the crosshead moves up, by a maximum of +30mm, and if a negative voltage is applied, the crosshead moves down, by a maximum of -30mm.

Two types of software are used in conjunction with the Instron: one to set up the limits and environment for the testing, Instron Console, and one to set up and carry out the testing method, Blue Hill.

3.1.2 TRIAL TEST

Before the testing was carried out, a suitable method had to be established and trialed to ensure all tests ran accordingly and errors were limited. During this trial session, the method of operation of the Instron was made familiar, and trials were carried out. The details of the trials are detailed below, along with any problems that were encountered.

One sample of each dressing, DuoDERM and AQUACEL was tested in the trial. A 10kN load cell was used in the Instron to perform a linear tensile test. Each sample was strained to 50% at a rate of 6.25mm/second. The sample was then held at 50% strain to allow for any stress relaxation to be recorded. The backing paper that covers and protects the adhesive side on the hydrocolloid dressing was left on during the trial to protect the grips of the Instron from the adhesive surface. Whilst held in the grips, however, the adhesive was pressed through the paper backing, and on attempting to ungrip the samples, the dressing adhesive had glued the Instron grips together. This then required for the grips to be carefully separated and cleaned of the adhesive. To prevent further issues with the adhesive sticking the grips together, masking tape was applied round the ends of each sample, protecting the Instron grips from the dressing adhesive. When the sample was then removed from the Instron grips, the end sections were examined for any signs of the

sample having slipped whilst being strained. There was a neat pattern from the grips imprinted onto each samples with no evidence of the sample slipping in the grips, suggesting the grips create the friction required for the tensile testing of orthopaedic dressings.

For the testing of the first sample, the data was captured at 10Hz, meaning the software collects data every 100ms. On examining the resultant data, it was evident that capturing data once every 100ms was too slow, as not enough data was produced for the period of the test. The frequency of data capture was then changed to 100Hz for the second sample, so every 10ms, and this decrease in time interval produced a lot more data that the initial test. On attempting to analyse the data produced from trial one, it was observed that the data file was in the wrong format. A setting within the software was altered to ensure the raw data was exported as a Comma Separated Values (.csv) file. The second test was then carried out with this altered setting, producing a file of data compatible with Microsoft Excel (Microsoft Corp, Redmond, USA). Graphs were then drawn to allow analysis of the samples load response to the prescribed displacement. By inspection of the data, it was made evident that the 10kN load cell was not sensitive enough for the small loads being recorded. It was therefore necessary to change the load cell to the more sensitive 1kN one for the requirements of this testing.

3.1.3 PROCEDURE

25mm x 5mm dumbbell shape samples were cut from each of the dressings, DuoDERM hydrocolloid and AQUACEL Ag Surgical. These samples can be seen in Figure 3.2. Masking tape was applied to the gripping ends of each sample before being loaded into the Instron grips.

A tensile test method was created on the Blue Hill software. Three different strain rates were used: 6.25mm/sec, 12.5mm/sec and 25mm/sec. Samples from each dressing were strained to 50%, 75% and 100% and the software recorded the loads required to carry out the specified displacement. Samples were then held at the 50%, 75% and 100% strains for 10 seconds to allow for stress relaxation within the samples. The Blue Hill software recorded the load as the sample relaxed.

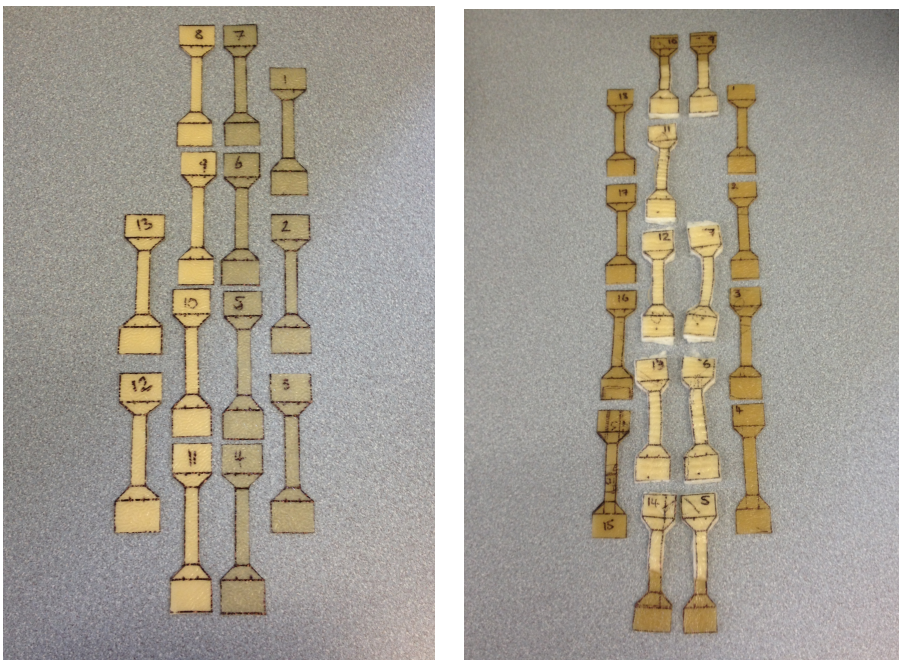

Figure 3.2: DuoDERM and AQUACEL dressings cut into samples for use in the Instron

Each sample was strained and relaxed three times at the fixed strain rate to determine whether or not the samples required higher or lower loads to carry out the displacement once the sample had already been strained.

The raw data from the testing was exported to a Comma Separated Values (.csv) file for use in Microsoft Excel (Microsoft Corp, Redmond, USA).

Results from the Instron testing were poor (see Section 4.1 – INSTRON ELECTROPULS E10000), so an alternative testing machine was used. The testing methods were altered slightly to accommodate the limits of the alternative machine. The following section describes the machine used instead, and the required change in methods.

3.2 ALTERNATIVE EXPERIMENT

3.2.1 BOSE ELECTROFORCE 3200 TEST INSTRUMENT

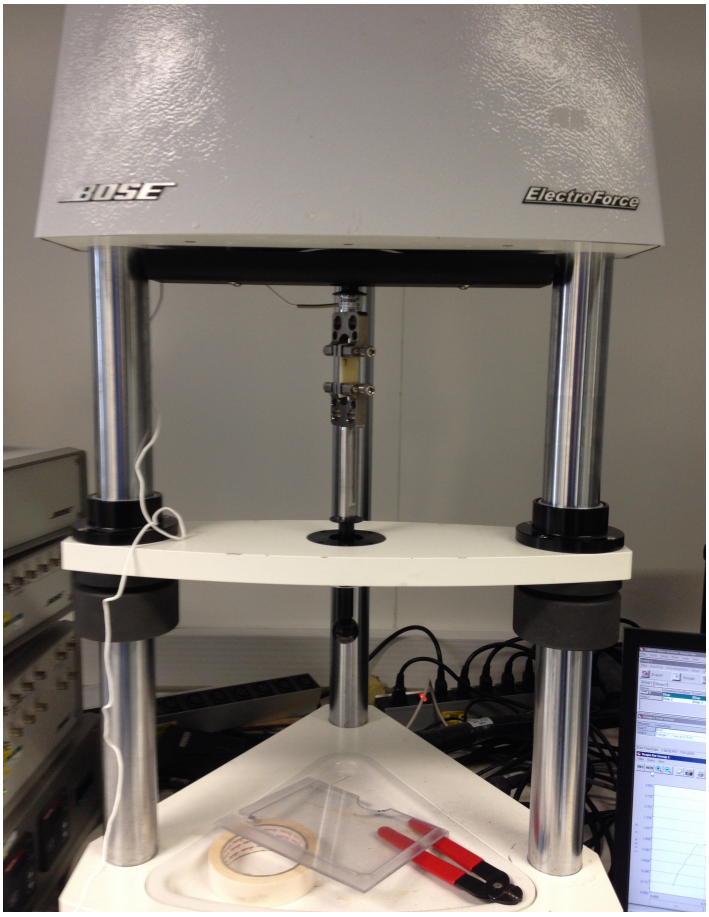

Figure 3.3: Bose ElectroForce 3200 testing instrument

The Bose ElectroForce 3200 is a highly sensitive instrument used in the mechanical testing of various different materials (Figure 3.3).

"Bose® test instruments incorporate proprietary linear motion technologies and WinTest® controls to provide a revolutionary approach to dynamic mechanical testing" (ElectroForce Test Instruments, 2015).

The capability of the Bose ranges from basic static tests to obtain tensile, compressive or bending data, to more advanced dynamic fatigue or fracture testing. These mechanical tests can be carried out on both engineered materials and biological materials, making this instrument highly versatile. The Bose is powered electrically via the mains electricity supply, therefore no additional power supplies or infrastructure is required (Bose Corporation, 2014). The electrical linear motor runs without friction, a crucial factor in low

force testing (Bose Corporation, 2014). The slightest frictional force during a test would be picked up by the sensitive load cell and incorporated into the resultant data, meaning the test results would be inaccurate and the mechanical properties of a material would be incorrectly characterised.

There are two ways in which the Bose can be configured, either with a 450N load cell, or a much more sensitive 22N load cell. Each load cell has a dry and wet version, for use with either dry or wet specimens. The load cell chosen is therefore dependent on the material that is being tested and how large or small the test loads are likely to be. These load cells are very delicate and must be handled with care to avoid damage. Figure 3.4 shows the small 22N load cell screwed onto the crosshead at the top. Another factor that is dependent on the material being tested, and the type of test being carried out, is the selection of grips or platens. The Bose has a range of grips and platens to suit multiple different testing methods. For the requirements of this project, a set of 25mm T/C Titanium flat, knurled face grips were chosen as the most suitable for gripping the soft, thin samples (Figure 3.4). These grips are simply screwed into place.

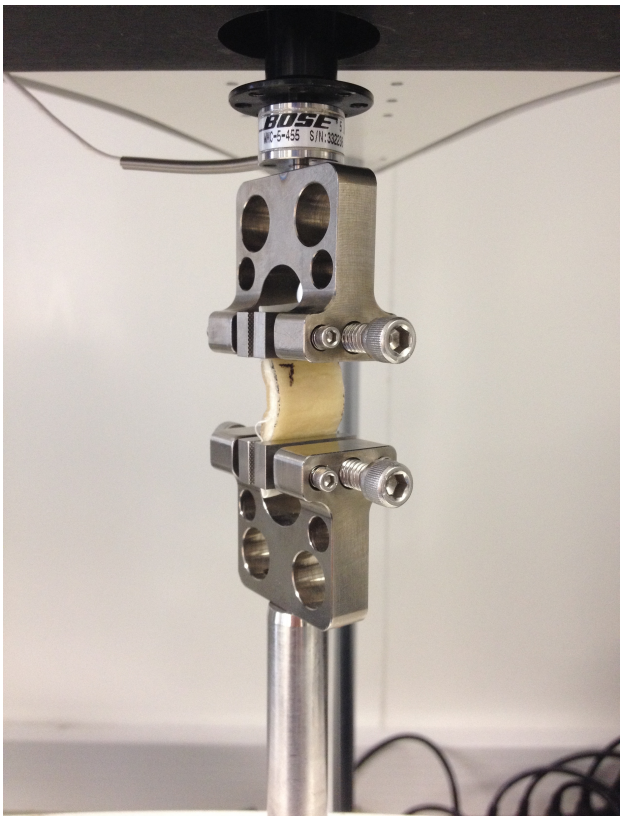

Figure 3.4: 25mm grips of Bose 3200 testing instrument

The linear displacement of the crosshead is limited to ±6.5mm, and this displacement is controlled by the electric motor, which has a velocity range of static to 3.2m/s. The load limit is dependent on the configuration, where the 450N load cell has a limit of ±450N, and the 22N load cell has a limit of ±22N. The Bose testing space can be adjusted by manually moving the white platform, seen in Figure 3.3, up and down to provide the desired test space.

To set up and carry out tests, appropriate software is required. WinTest 4.1 software is used in conjunction with the Bose 3200 test instrument. This software allows the user to set test limits to ensure the load cell is protected for the duration of the testing, design a test with as few or as many phases as required and run the test. The software then collects the raw data throughout the tests and produces a real time graph of the results, on which the user can choose the data they would like to see on the X and Y axes.

3.2.2 TRIAL TEST

Before the testing was carried out, a suitable method had to be established and trialed to ensure all tests ran accordingly and errors were limited. The details of this trial are detailed below, along with any problems that were encountered.

One sample from the AQUACEL® dressing was tested in the trial. The 450N load cell was used in the Bose to perform a linear tensile test. Before loading the sample into the Bose, the backing paper that covers and protects the adhesive side on the dressing was removed, and the ends of each sample were covered with masking tape. This prevented the adhesive surface of the dressing sticking to the grips whilst loaded in the Bose. The sample was then loaded into the Bose and gripped with the 25mm, knurled face grips, as seen in Figure 3.4, and strained to approximately 25% at a rate of 0.5mm/second. The sample was then held at 25% strain to allow for any stress relaxation to be recorded, before being restored back to its original length.

When the sample was then removed from the Bose grips, the end sections were examined for any signs of the sample having slipped whilst being strained. There was a neat pattern from the grips imprinted onto the sample, showing no evidence of the sample slipping in the grips

The raw data from the trial was exported as a Comma Separated Values (.csv) for use in Microsoft Excel (Microsoft Corp, Redmond, USA), and a corresponding graph was produced (Figure 3.5). On examining the graphical data, it is evident that the smaller, more sensitive 22N load cell would be better suited to the requirements of this test due to the low loads recorded in the trial.

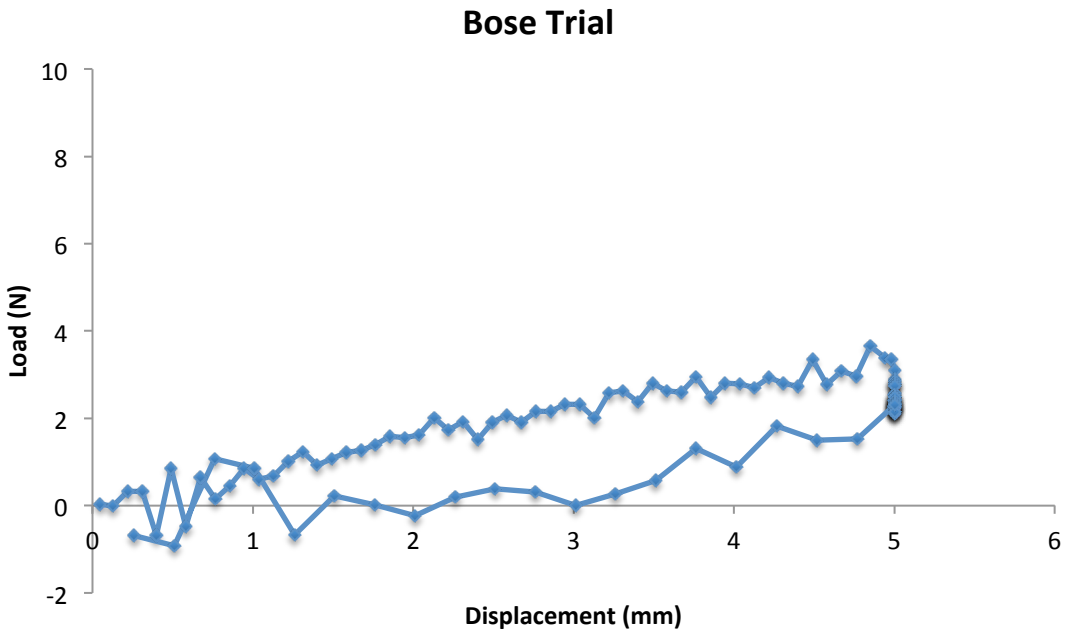

Figure 3.5: Load v extension graph for trial test on Bose configured with 450N load cell

3.2.3 PROCEDURE

Five different dressings were selected for the testing: ConvaTec AQUACEL® Ag Surgical Dressing, DuoDERM® Extra Thin Hydrocolloid, Coloplast Contreet Hydrocolloid, Mölnlycke Mepore® and Coloplast Biatain® Adhesive.

40mm x 20mm samples were cut from each of the five dressings. The dressings featuring an absorbant section required extra samples cut due to the non-uniformity of the entire dressing. These samples can be seen in Figure 3.6.

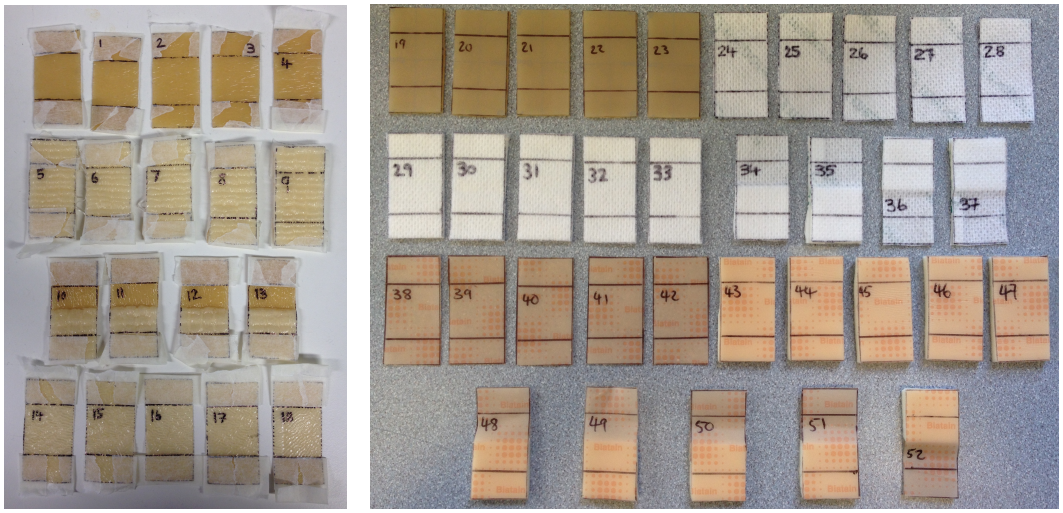

Figure 3.6: Samples of the five dressings being tested. Samples 1-13 – ConvaTec AQUACEL®. Samples 14-18 - DueDERM® Extra Thin Hydrocolloid. Samples 19-23 - Contreet Hydrocolloid. Samples 24-37 - Molnlycke Mepore®. Samples 38-52 - Biatain® Hydrocolloid

Two of the hydrocolloid dressings (DuoDERM® and Contreet) are purely adhesive, so are assumed to be uniform throughout. Five samples were cut from each of these dressings. The other three dressings, however, are non-uniform. They consist of both an adhesive and an absorbant section, both of which are assumed to have varying properties. For the non-uniform dressings, four samples were taken from AQUACEL adhesive, AQUACEL half and half and Mepore half and half, and five samples were taken from AQUACEL absorbent, Mepore adhesive, Biatain adhesive, Biatain absorbent and Biatain half and half.

The width and thickness of each sample was accurately measured using a Digital Vernier Caliper and a Digital Micrometer. These devices provide highly accurate readings, an important factor, as these measurements are crucial for calculating the stress within the sample when strained.

As mentioned in Section 3.2.2, masking tape was applied to the gripping ends of each sample before being loaded into the Bose grips. The Bose testing instrument was carefully set up with the 22N load cell and the 25mm knurled face grips. The gauge length was fixed at 19.1mm.

WinTest 4.1 software was loaded to set up and control the testing procedure. The limits within the software were adjusted to protect the 22N load cell, and ultimately the entire test instrument. A ±6mm displacement limit was set, alongside a ±20N load limit. A ramp and hold test method was then created. The ramp phase was set to strain the samples to 5mm at a rate of 1mm/second. The next phase was set to hold the samples at 5mm displacement for 60 seconds to allow the material to relax, before the sample was then restored back to its initial position at a rate of 1mm/second. Each sample was loaded into the Bose grips and the test carried out. Raw data for time, displacement and load was exported to a Comma Separated Values (.csv) file for use in Microsoft Excel (Microsoft Corp, Redmond, USA).

3.2.4 DATA ANALYSIS

The cross sectional area for each sample was calculated using the measurements from the Digital Vernier Caliper and Digital Micrometer. The calculated cross-sectional areas for each sample can be found in Table B1, Appendix B. Stress and strain was then calculated for each set of data using Equations 3.1 and 3.2.

$$Stress\ (Pa) = \frac{Load\ (N)}{Cross - sectional\ area\ (m^2)}$$

Equation 3.1: Equation for the calculation of stress

$$Strain = \frac{Displacement\ (mm)}{Original\ length\ (mm)}$$

Equation 3.2: Equation for the calculation of strain

$$Young's\ Modulus\ (E) = \frac{Stress\ (Pa)}{Strain}$$

Equation 3.3: Equation for the calculation of Young's Modulus

Stress (y-axis) vs strain (x-axis) was plotted on Microsoft Excel (Microsoft Corp, Redmond, USA) for each sample. Linear trendlines were added to the ramp phase on the graphs, and the equations of these trendlines were displayed (Appendix D). The gradient of each trendline represents stress over strain, the basic calculation for Young's Modulus, as seen in Equation 3.3. The Young's Modulus values from the trendlines were compiled into a table

and colour coded for each dressing. The mean Young's modulus value was then calculated for each dressing and presented on a bar graph. Again, standard deviation and standard error were calculated for the addition of error bars on the Young's modulus graph.

Mean stress and strain values for each type of sample (AQUACEL adhesive, absorbent and half & half; DuoDERM adhesive; Contreet adhesive; Mepore adhesive, absorbent and half & half; Biatain adhesive, absorbent and half & half) were calculated and presented on graphs. The standard deviation was also calculated to allow the standard error to be calculated for the mean values. The standard error values were then used in the addition of error bars.

Statistical analysis was carried out using Minitab 16 software.

A one way ANOVA test was carried out on the Young's Modulus results to compare the means among the four groups. A one way ANOVA test was carried out on the adhesive samples Young's Modulus data. 2-sample t-tests were carried out on the absorbent and half & half Young's Modulus data. Mepore data was excluded from statistical analysis due to considerably larger (x20) Young's Modulus values.

4. RESULTS

4.1 INSTRON ELECTROPULS E10000

The following section highlights the results obtained from testing three different samples from the AQUACEL dressing on the Instron ElectroPuls E10000.

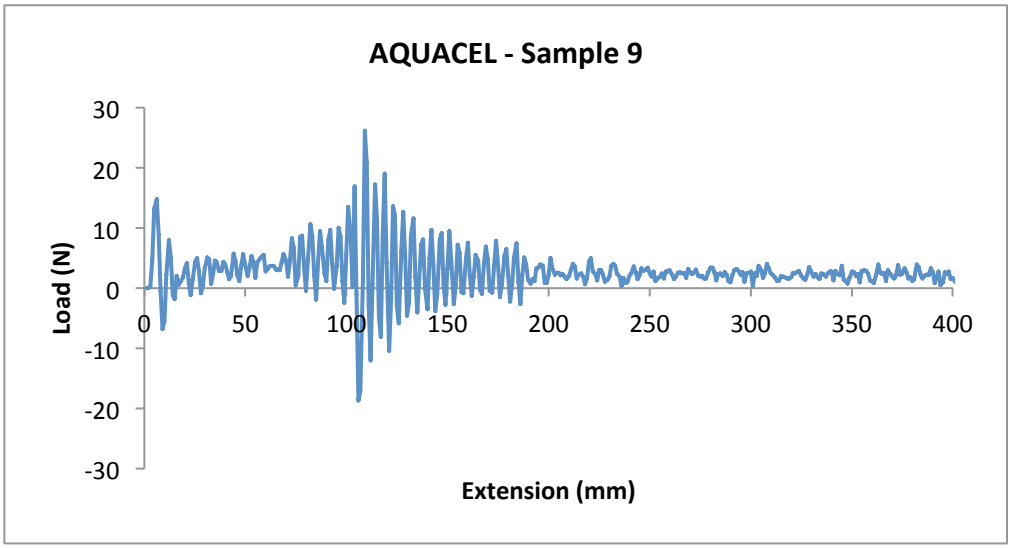

Figure 4.1: Load vs extension result when AQUACEL sample 9 was tested in Instron

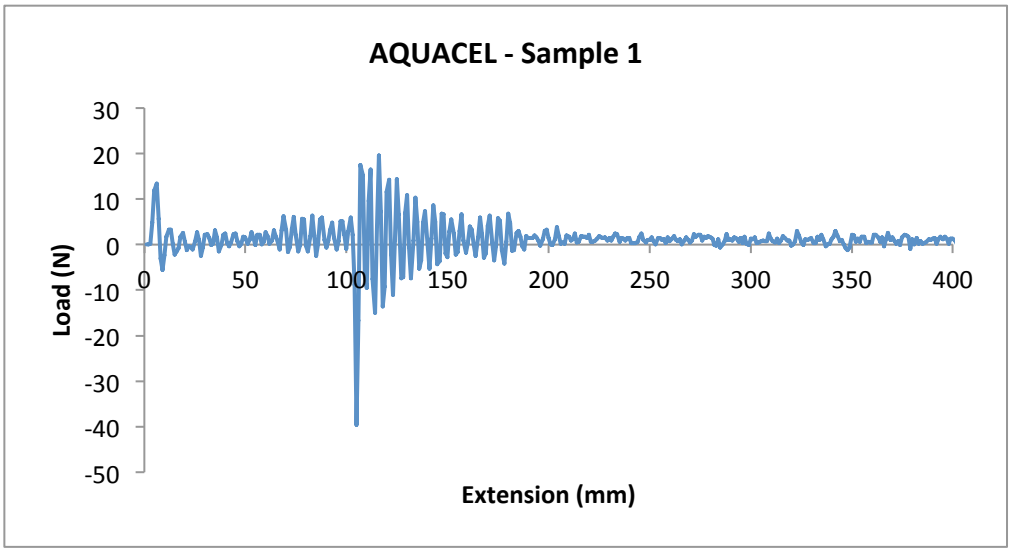

Figure 4.2: Load vs extension result when AQUACEL sample 1 was tested in Instron

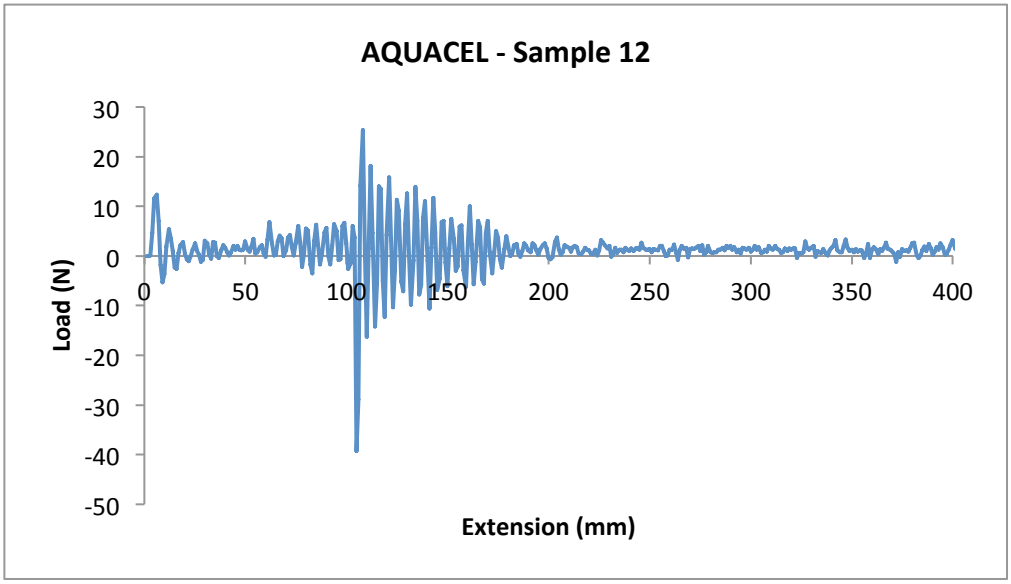

Figure 4.3: Load vs extension result when AQUACEL sample 12 was tested in Instron

Figures 4.1-4.3 indicate the unsuitability of the 1kN load cell used in the Instron to carry out tensile testing on the dressing samples. As strain increases, the load is expected to also increase, but it can be seen that as the AQUACEL samples are strained, the load response is very erratic. These responses are not characteristic of a tensile test, so it was necessary to develop another method of testing. As described in chapter 3, the Bose ElectroForce testing instrument was used for alternative testing, the results of which are presented below.

4.2 BOSE ELECTROFORCE 3200 TESTING INSTRUMENT

The following section involves references to sample numbers. A table with these sample numbers and the corresponding type of sample can be found in Appendix B.
Individual sample graphs can be found in Appendix C.

4.2.1 STRESS-STRAIN RESPONSE

Four of the five wound dressings involved in this project were stretchy by inspection, with the fifth dressing (Mepore) being very inelastic in comparison. It was therefore expected that the Mepore dressing would experience significantly higher stresses when strained.

Figure 4.4 compares the mean stress-strain response of the purely adhesive samples from four of the five dressings. Figure 4.5 compares the mean stress-strain response of the purely absorbent samples from the AQUACEL, Mepore and Biatain dressings. Figure 4.6 compares the mean stress-strain response of the samples from AQUACEL, Mepore and Biatain that consist of both an adhesive section and an absorbent section.

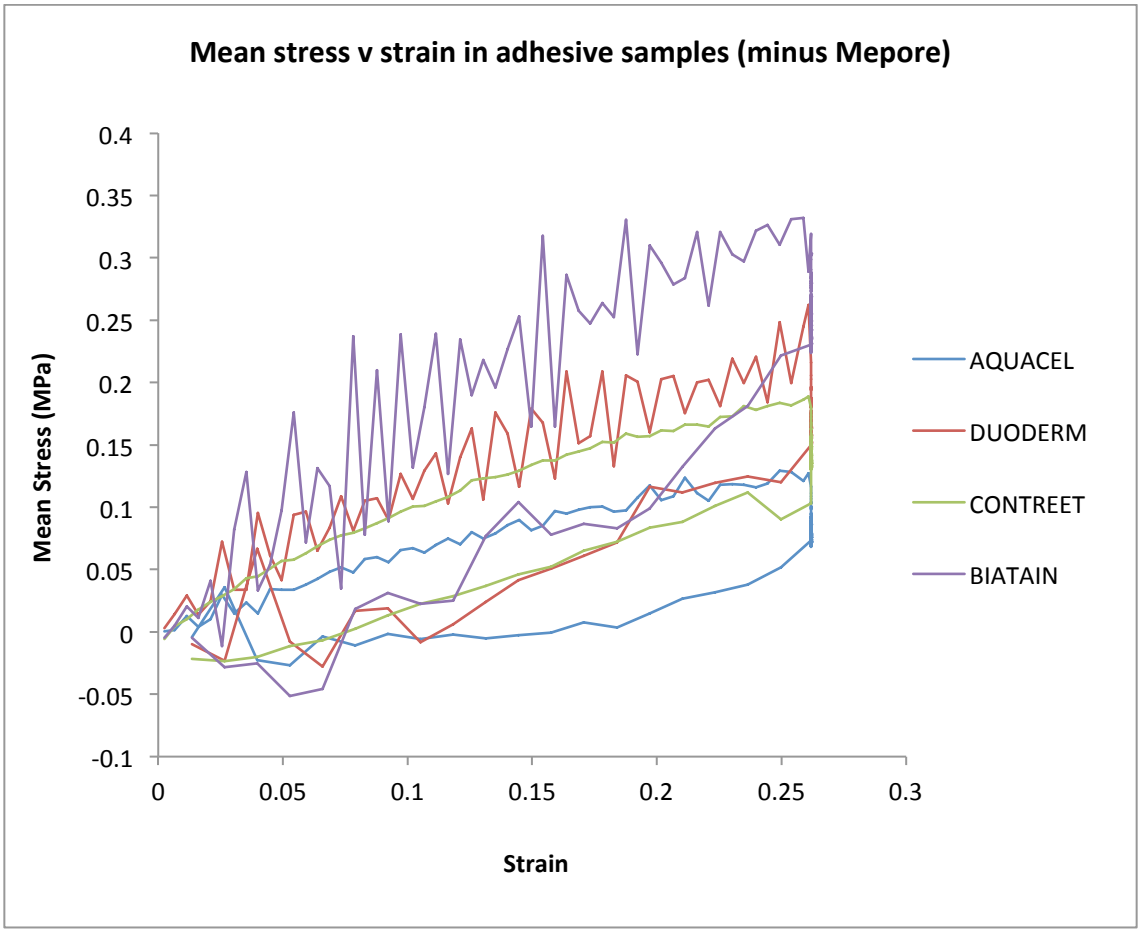

Figure 4.4: Mean stress-strain response of the adhesive samples from four of the five dressings

Standard error bars were making the graph hard to read due to variability of some of the results, so were removed.

Biatain adhesive samples have the largest stress response (maximum = 0.332MPa), DuoDERM and Contreet samples have similar stress responses, and AQUACEL adhesive samples show the lowest stress response (maximum = 0.129) (Figure 4.4).
Biatain absorbent samples show a larger stress response than AQUACEL absorbent samples with maximum stress values of 0.078MPa and 0.050MPa respectively (Figure 4.5).

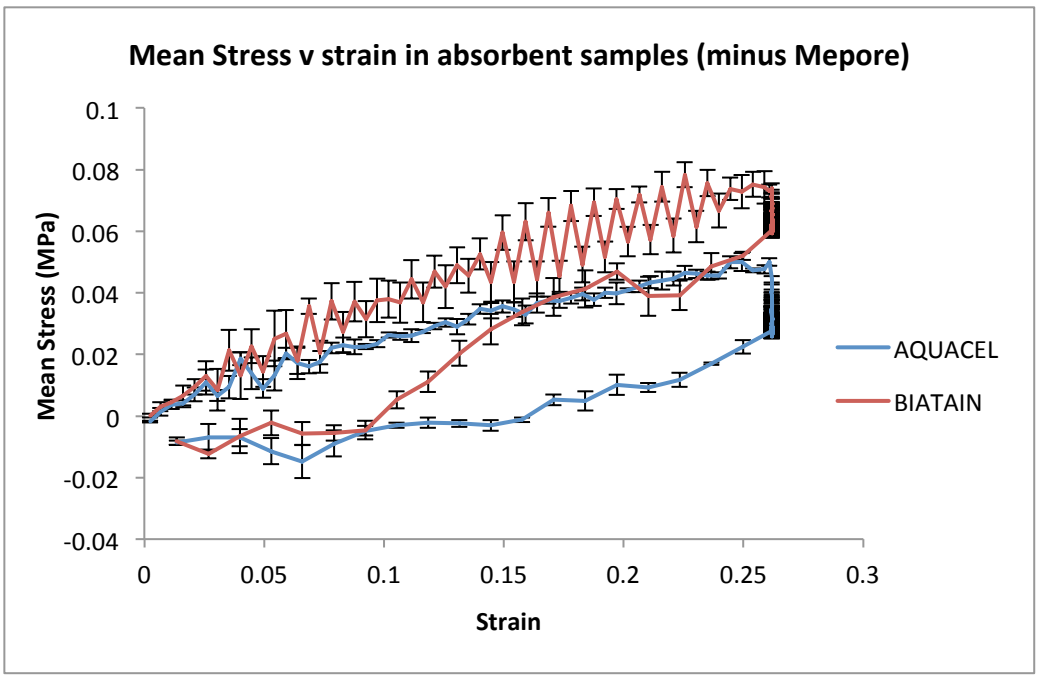

Figure 4.5: Mean stress-strain response of absorbent samples from AQUACEL and Biatain

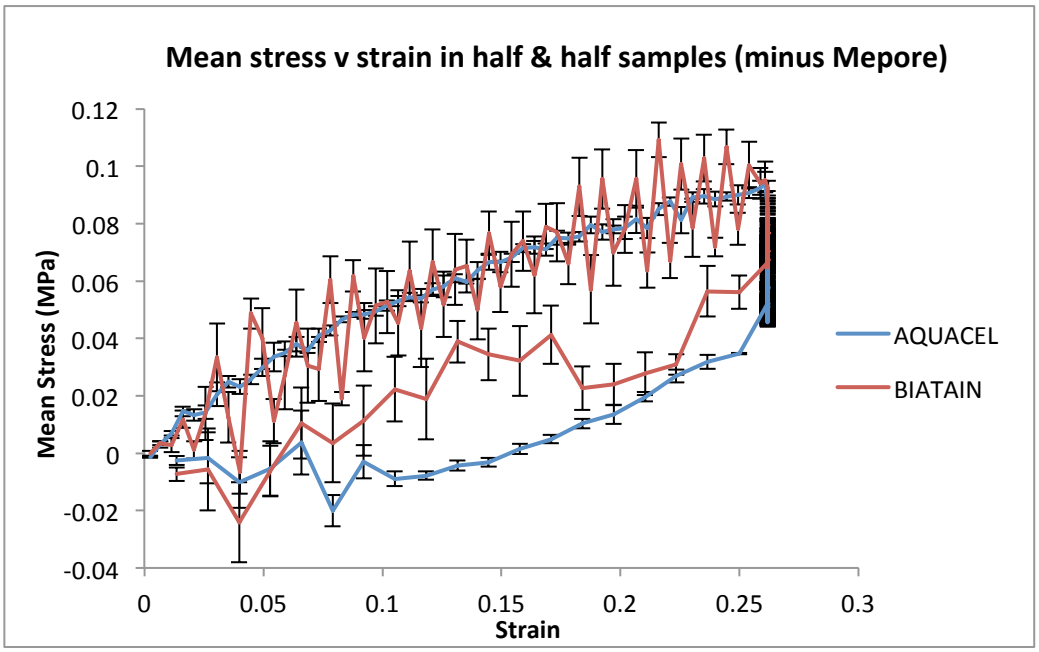

Figure 4.6: Mean stress-strain response for AQUACEL and Biatain samples containing both adhesive and absorbent sections

Biatain and AQUACEL half & half samples have a very similar stress response, with maximum values reaching 0.109MPa and 0.094MPa.

Figures 4.7 and 4.8 show the mean stress and strain responses for each group of identical samples. Figure 4.7 shows the mean stress and strain response for adhesive, absorbent and half & half AQUACEL samples, adhesive DuoDERM samples, adhesive Contreet samples and adhesive, absorbent and half & half Biatain samples. Figure 4.8 shows the mean stress and strain response for adhesive, absorbent and half & half Mepore samples.

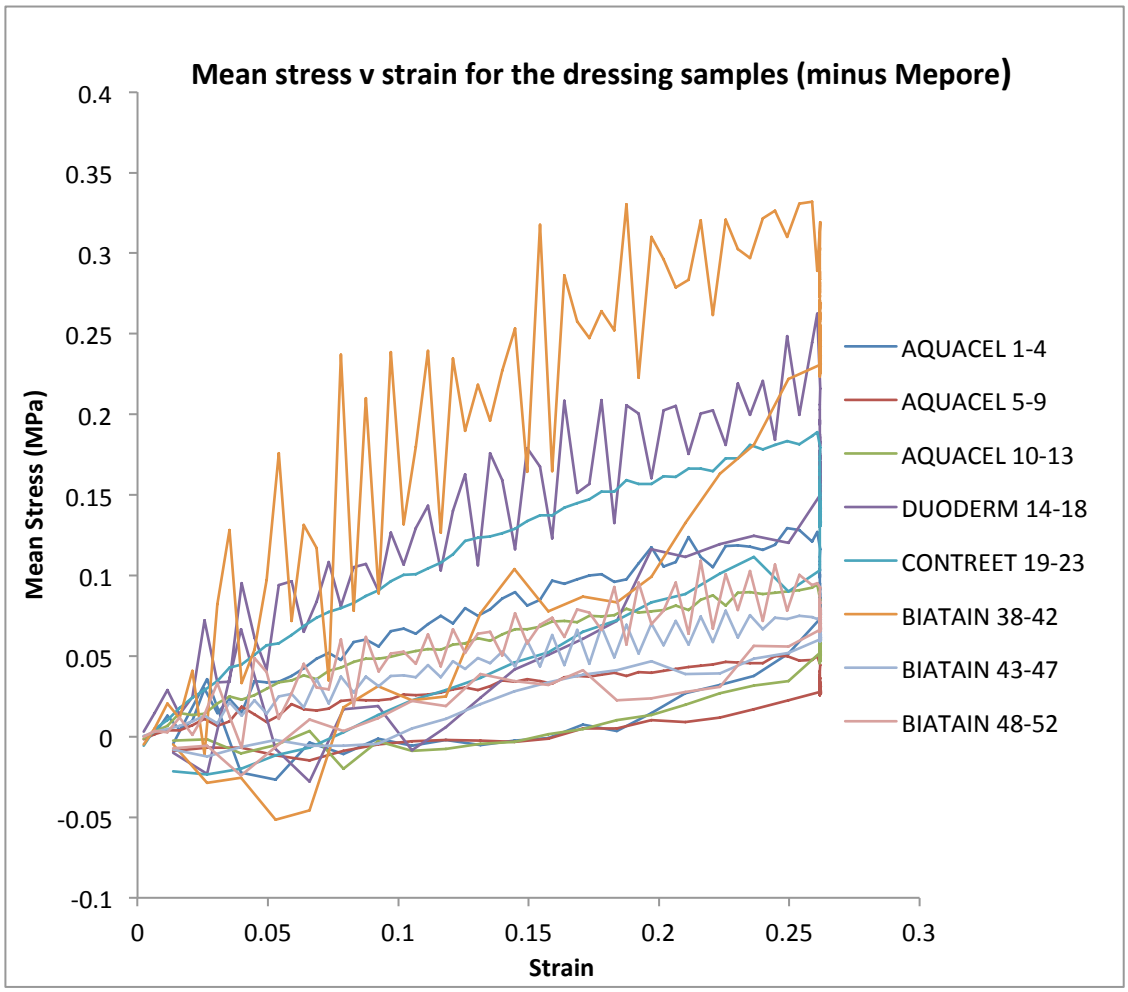

Figure 4.7: Mean stress-strain response for each type of sample (Absorbent, Adhesive or half & half) for four out of the five dressings

It can be seen in Figure 4.7 that AQUACEL 5-9 (AQUACEL absorbent), on average, experience the least amount of stress (0.05MPa). Conversely, BIATAIN 38-42 (Biatain adhesive), on average, experience the largest amounts of stress (0.3MPa).

Error bars were, again, removed due to variability of some results.

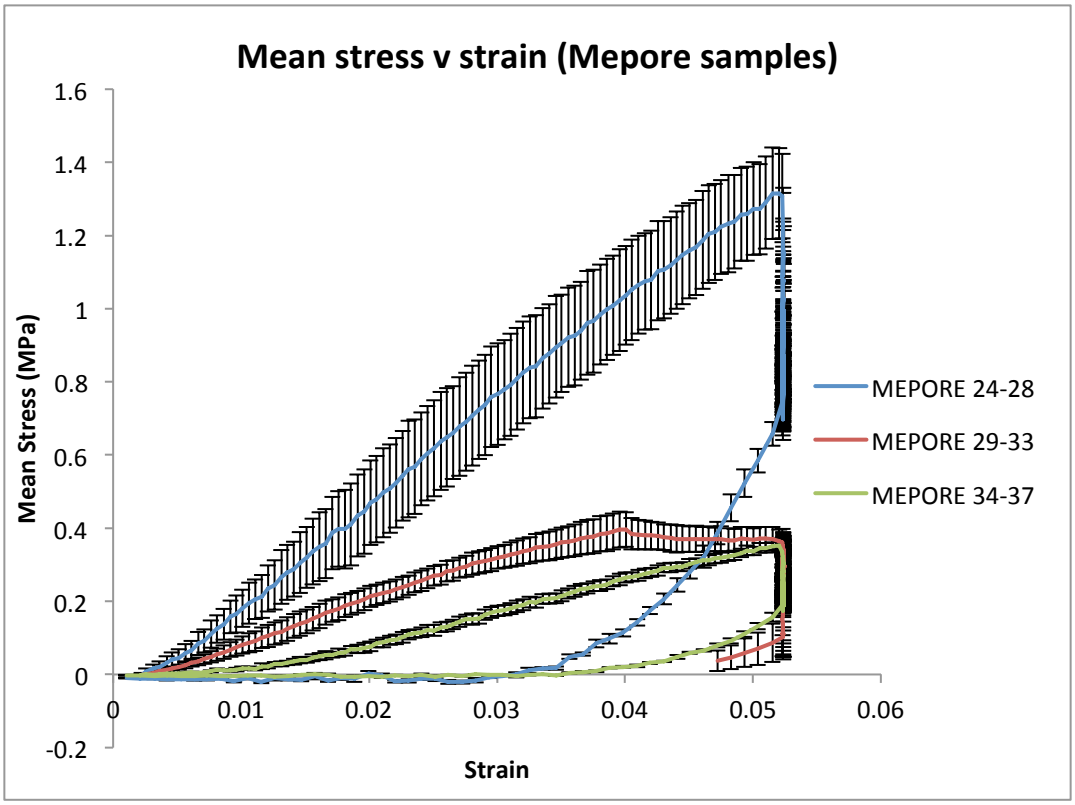

Figure 4.8: Mean stress-strain response of the three types of sample from Mepore dressing

The mean stress experienced in the adhesive Mepore samples (Mepore 24-28, Figure 4.8) is notably higher than the absorbent or half & half samples, reaching as much as 1.3MPa, compared with only 0.4MPa for the absorbent and half & half samples.

Figures 4.9 and 4.10 show a total mean stress-strain relationship for each dressing, taking all samples into consideration.

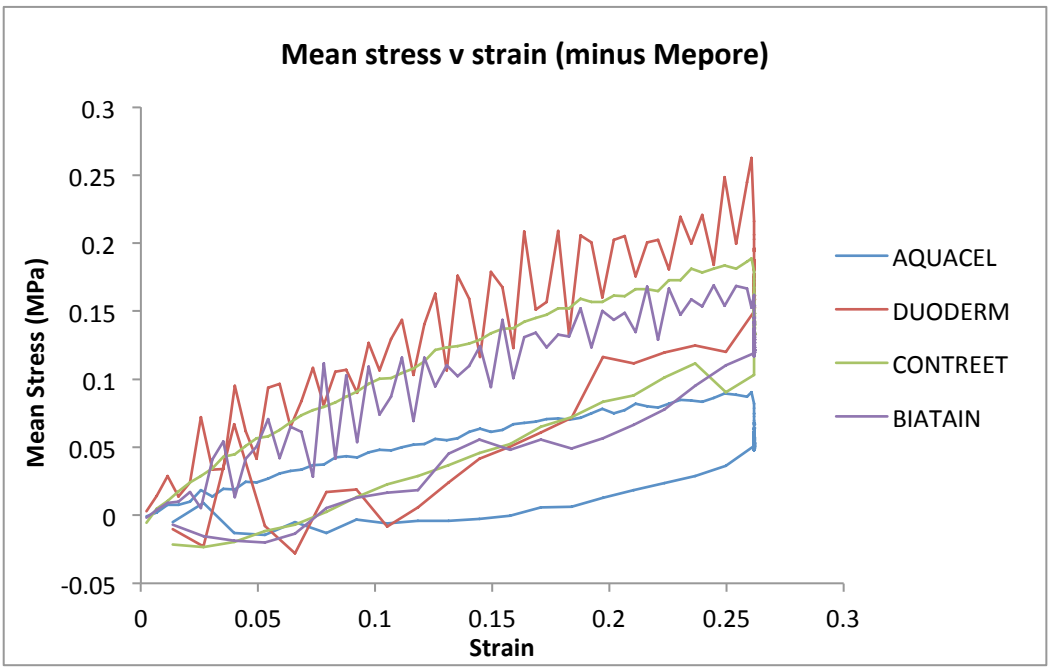

Figure 4.9: Mean stress-strain response of all samples from each dressing (minus Mepore)

The maximum mean stresses experienced by each dressing are seen in Figures 4.9 and 4.10, and are listed as follows: AQUACEL: 0.091MPa, DuoDERM: 0.262MPa, Contreet: 0.189MPa, Biatain: 0.169MPa and Mepore: 0.679MPa.

Standard error bars were removed in Figure 4.9 to improve readability of graph.

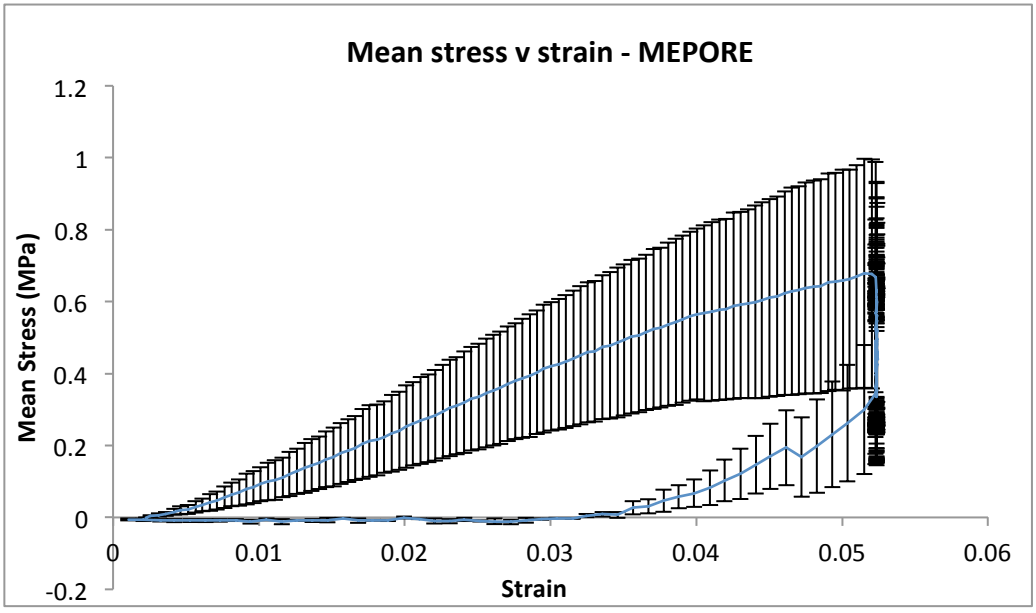

Figure 4.10: Mean stress-strain response for all samples from Mepore dressing

4.1.2 YOUNG'S MODULUS

Table 4.1 summarises the estimated Young's Moduli for every sample tested. Corresponding graphs highlighting the stress-strain response of each sample can be found in Appendix C.

Table 4.1: Young's Moduli obtained from trendlines in Figures B1-B5

Sample No.	Young's Modulus (MPa)	Sample No.	Young's Modulus (MPa)
1	0.508	27	33.147
2	0.4792	28	- *
3	0.4868	29	5.2154
4	0.4313	30	8.2982
5	0.1803	31	- *
6	0.1694	32	14.684
7	0.1726	33	14.009
8	0.1766	34	7.3689
9	0.2006	35	8.3743
10	0.3095	36	7.3366
11	0.3412	37	8.5169
12	0.3164	38	1.2132
13	0.3523	39	1.2203
14	0.7762	40	1.2705
15	0.7594	41	1.2117
16	0.8914	42	1.2132
17	0.8174	43	0.2785
18	0.7733	44	0.2827
19	0.7544	45	0.2811
20	0.7199	46	0.2644
21	0.7008	47	0.2833
22	0.6943	48	0.3407
23	0.5846	49	0.3668
24	28.711	50	0.37
25	27.975	51	0.3574
26	32.31	52	0.334

KEY:	= AQUACEL	= DuoDERM
= Contreet	= Mepore	= Biatain

*Straining samples 28 and 31 required a load greater than the 20N limit set on the Bose testing machine, so the test was stopped automatically when this limit was reached. The data for these tests wasn't useable, so was discarded.

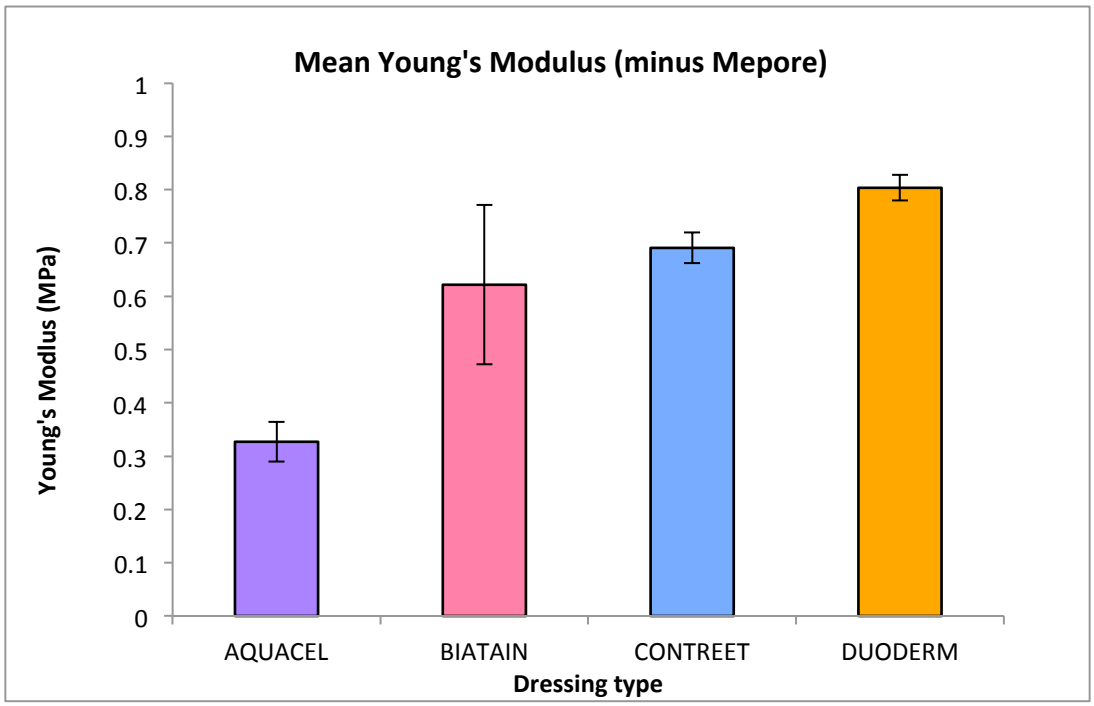

Figure 4.11: Bar graph comparing the mean Young's Modulus ±SE for four out of the five dressings

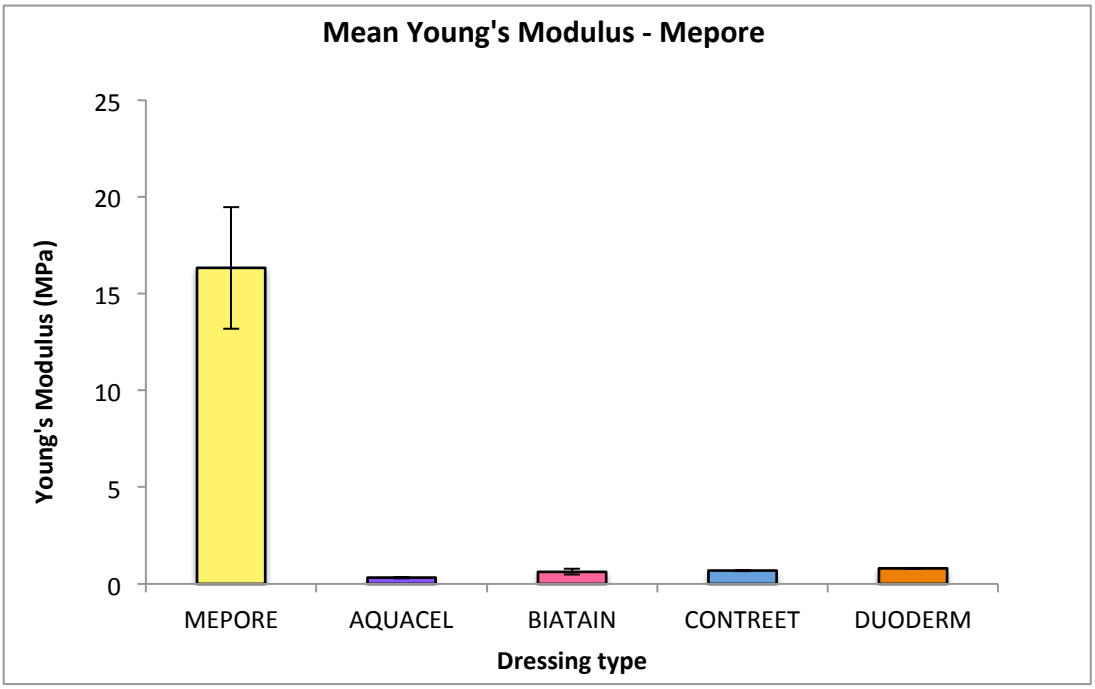

Figure 4.12: Bar graph comparing the mean Young's Modulus ±SE for all five dressings

Figures 4.11 and 4.12 show that the five different dressings have a variety of mean Young's Moduli ranging from 0.327 ± 0.04MPa (AQUACEL) to 16.33 ± 3.15MPa (Mepore). The Young's modulus of AQUACEL significantly differs from Biatain, Contreet and DuoDERM (One-way ANOVA) (p=0.009).

Mepore data was omitted from all statistical testing.

When comparing the Young's moduli between the purely adhesive samples (APPENDIX C), all four dressing types differed significantly from one another (p<0.001). Table 4.2 shows the Post-hoc Fisher test result carried out as part of the ANOVA test. Means that do not share a letter are significantly different.

Grouping Information Using Fisher Method

	N	Mean	Grouping
BIATAIN	4	1.23802	A
DUODERM	4	0.81037	B
CONTREET	4	0.71735	C
AQUACEL	4	0.47633	D

Table 4.2: Post-hoc Fisher test results for comparing the adhesive samples Young's Moduli (Minitab)

As none of the four dressings share a letter in the Fisher test (Table 4.2), the Young's Moduli of all four groups differ significantly.

2-sample t-test results indicate that the Young's Moduli for the purely absorbent AQUACEL and Biatain samples (APPENDIX C) are significantly different (p =0.001).

2-sample t-test results prove that the Young's Moduli for the half & half AQUACEL and Biatain samples (APPENDIX C) are significantly different (p=0.015).

5. DISCUSSION

Due to the location of joint replacement surgery, the wound tends to lie directly over the joint. An ideal wound dressing, therefore, requires to have sufficient stretch ability to allow the dressing to move with the joint as it moves, limiting shear forces acting on the skin. Keeping shear forces as low as possible is very important in the reduction of post-operative skin blistering; a commonly reported problem in the wound care regime following joint replacement surgery.

The dressings tested in this project varied in material composition, and the mechanical testing results, therefore, varied among the dressings. This chapter explores the findings from this project and discusses the relevance of these results.

5.1 DRESSING PROPERTIES

Figures 4.4-4.6 exhibit the mean stress vs strain responses for the adhesive samples from each dressing (Figure 4.4), the absorbent samples from each dressing (Figure 4.5) and the half & half samples from each dressing (Figure 4.6). The mean results from the Mepore testing were plotted on a separate graph (Figure 4.8) as the stress values were up to four times larger than the largest mean stress value for any of the other dressings.

When comparing the adhesive samples from each dressing, the stress experienced for 0.25 strain, ranges from 0.129MPa for AQUACEL, to 0.33MPa for Biatain (Figure 4.4). This low stress value in the AQUACEL dressing is a benefit when looking from a clinical aspect. If the adhesive section of the dressing requires a reduced magnitude of load to be strained, then the stress transferred to the skin as a shear force will also be reduced; a crucial factor in the prevention of skin blistering. As a contrast, to strain the Mepore adhesive samples to as little as 0.05, the mean stress response was 1.32MPa (Figure 4.8). This much larger value for Mepore is expected due to its longitudinally inelastic nature. Inelasticity is a highly undesirable property for an orthopaedic wound dressing, as the dressing is required to stretch and accommodate a wide range of movement, as mentioned at the start of this chapter.

Figure 4.5 compares the stress-strain response for the absorbent samples from the AQUACEL and Biatain dressings. The observed maximum mean stresses for AQUACEL and

Biatain are 0.05MPa and 0.078MPa respectively. Again, the mean stress for the absorbent Mepore section, 0.396MPa, is almost ten times larger than that of the other dressings. Since the absorbent sections of the dressing do not adhere to the skin, their stretch capability does not impact directly on the skin. If the absorbent section has no stretch ability, however, as soon as a load is applied and the dressing is required to increase in length (during knee flexion), the absorbent section will resist any strain and consequently, the elastic adhesive sections would take the majority of the strain. If testing whole dressings had been an option in this investigation, the way in which the adhesive and absorbent sections strain when a load is applied, could have been studied.

It is important that the absorbent section has an elastic element, allowing it to stretch and share the strain with the adhesive section. The samples consisting of both adhesive and absorbent parts are crucial in this investigation to determine how the dressing behaves at the point where the adhesive section joins the absorbent section. It can be seen in Figure 4.6 that the mean stress response of the half & half samples for both AQUACEL and Biatain are very similar, both reaching 0.1MPa at 0.25 strain. When comparing the half & half Mepore samples (Figure 4.8) to the half & half samples from AQUACEL and Biatain, it is observed that the mean stress response of Mepore is almost four times larger than the aforementioned.

Dillon et al., 2007 carried out similar tensile testing on four different wound dressings. The testing carried out in their study differs from the testing in this investigation in that the entire dressings were tested as one sample, as opposed to many small samples from one dressing. This method gives a better representation of how the whole dressing would behave whilst applied on an orthopaedic wound. Dillon et al., 2007, also tested the Mepore adhesive dressing, with the results following a similar trend to those discussed earlier in this chapter. The load required to strain the entire Mepore dressing to approximately 5% was in excess of 60N, whereas to strain the other three dressings to >25% required loads as little as 6.5N, seen in Figure 2.5, Chapter 2.

Appendix B contains graphs presenting the stress-strain response for each individual dressing sample, along with their corresponding Young's Modulus values, seen in Table 4.1. The mean Young's modulus was calculated for each dressing and presented in Figures 4.11

and 4.12. AQUACEL is the least stiff dressing, with a mean Young's Modulus of 0.327 ± 0.04MPa and Mepore is the stiffest dressing with a mean Young's Modulus of 16.33 ± 3.15MPa. Applying these findings, along with the stress values reported earlier in this chapter, it is evident that the Mepore dressing is mechanically unsuitable for use in the orthopaedic wound care regime.

Mechanical unsuitability of a wound dressing plays a major role in the occurrence of blistering. Bhattacharyya et al., (no date), compared two different dressings (Mepore and Opsite PostOp) and the prevalence of blistering with the clinical use of these dressings. 6% of patients that were given the Mepore adhesive dressing developed blisters, whereas no blistering was observed with the use of the Opsite PostOp dressing. Abuzakuk et al., 2006 investigated the management of post-operative wounds and compared the use of AQUACEL and Mepore. They reported 13% of the AQUACEL patients experienced blistering, compared to 26% of the Mepore patients.

The majority of blistering reported with the use of the Mepore dressing is caused when the dressing is applied to patients having undergone joint replacement surgery. Since the dressing is applied over a joint, it must, therefore, be flexible to accommodate a wide range of joint movement. An inelastic dressing, such as Mepore, cannot accommodate the required joint movement, the results of which are undesirable. Using the findings from this investigation and knowledge from previous studies, it can be stated that the use of Mepore, and dressings with similar mechanical properties to Mepore should be avoided in the clinical setting to help reduce the incidence of post-operative blistering.

5.2 RELATIONSHIP BETWEEN DRESSING PROPERTIES AND SKIN

When studying the properties of a wound dressing, it is important to assess the implications of the dressing on the skin, as well as the mechanical behaviour of the dressing alone. To understand the relationship between dressing properties and skin, the mechanical properties of skin must be established.

Previous studies have discovered that the composition of skin creates a mechanically complex material. Ní Annaidh et al., 2012 state "Skin is a highly non-linear, anisotropic, viscoelastic and nearly incompressible material." When observing the stress-strain response for skin, it is evident that skin from varying locations on the body has differing mechanical properties. Figure 5.1 displays the stress response of healthy skin excised from the back region. It is observed that there is an initial slope at infinitesimal strains before the gradient of the curve then increases as the skin displays linear elastic behaviour, the gradient of which (B) represents the Young's Modulus (Ní Annaidh et al., 2012).

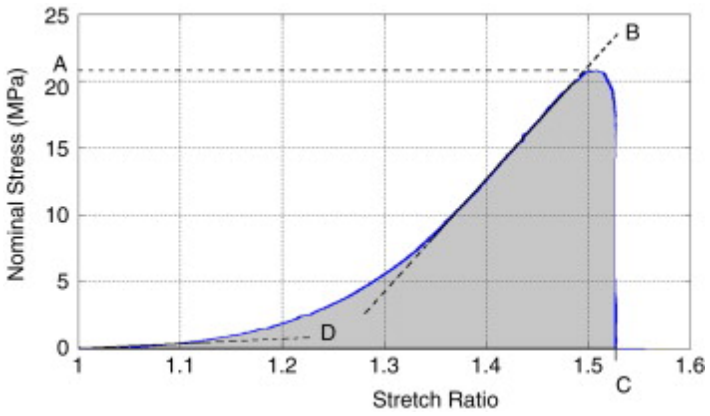

Figure 5.1: Nominal stress vs stretch ratio response of human skin to uniaxial tensile testing (Ní Annaidh et al., 2012)

On comparing the mean stress responses of the dressings in this investigation (Figures 4.9 and 4.10) to the stress response seen in Figure 5.1, it can be seen that the dressings do not experience the same initial slope that skin experiences. The dressings experience a uniform increase in stress, and the linear elastic phase begins from initiation of the strain. The difference with the stress response of skin is a result of its composition, where there is a delicate network of elastin fibres, thought to be responsible for the initial stiffness slope, and collagen fibres, the main provider of mechanical strength (Oxlund, Manschot and Viidik, 1988). The stress experienced by skin at 25% strain is about 3MPa (Figure 5.1), whereas the mean stress for the dressings (not including Mepore) at the same strain, ranges from 0.091MPa to 0.262MPa (Figure 4.9). Since the thickness of the dressing samples and the skin are similar, it can be concluded that the loads required to strain skin are larger than those required to strain the dressing samples. In relation to the effect the dressing has on the skin, if the load required to strain the dressing is less than the load

required to strain the skin, then the dressing will take strain before the skin does, so there should be no excess stress placed on the skin due to inelasticity of the dressing.

Dunn and Silver, 1983, found that skin from the abdomen and thorax had an average Elastic modulus of 18.8MPa, whereas Zahouani et al., 2009, found that skin from the arm has a mean Young's modulus of 8.3kPa. Unfortunately there is little, to no research been carried out on the mechanical properties of the skin around the knee joint, making it difficult to directly assess the effect mechanically different dressings have on the skin. An ideal dressing, however, will have a lower Young's Modulus than that of skin. A dressing with a greater stiffness than skin, would strain at a slower rate than skin, creating shear force between the dressing and the surface of the skin, and as mentioned earlier in this chapter, this is the main cause of skin blistering. The mean Young's Modulus values calculated in this study range from 0.327MPa for AQUACEL to 0.804MPa for DuoDERM (Figure 4.11) for the elastic dressings, and the inelastic Mepore dressing presenting with a much larger mean Young's Modulus of 16.33MPa (Figure 4.12). The reported level of blistering caused by the use of Mepore on orthopaedic wounds suggests that the dressing is too stiff for the use on joint wounds. It can therefore be concluded that the Young's Modulus of the skin around the knee is closer to that of the skin on the arm than the skin on the abdomen and thorax.

5.3 LIMITATIONS

On examination of the stress vs strain graphs presented in Chapter 4 and Appendix B, it is observed that the response of some of the dressing samples is very 'bouncy' and erratic. The increasing stress with strain trend is, however, characteristic with tensile testing, and the values presented are consistent with similar testing. Samples from two of the dressings (DuoDERM and Biatain) produce bouncier results than the other three dressings. The adhesive sections of these two dressings are thinner than AQUACEL and Contreet, and maybe experience some form of vibration while strained; a potential explanation for the 'bouncy' results. If the samples had all been the same thickness, there may have been less 'bouncing', but that was not a possible option for the testing of these dressings. Every sample was handled and tested in exactly the same way, eliminating human error during testing as justification for the erratic data. Other than the comments mentioned, it is not known what caused the bouncing seen in some of the results.

Samples containing both an adhesive and an absorbent part (AQUACEL, Mepore and Biatain) are non-uniform. One half is the thin adhesive, the other half is the thicker absorbent. When calculating the stress experienced in each sample, according to Equation 3.1, one cross-sectional area is required. To allow this equation to be used, an average cross-sectional area was calculated for the half adhesive and half absorbent samples. This instantly makes the stress response slightly inaccurate. More advanced research could be carried out for these samples to asses the stress experienced in each half of the samples, and the strain distribution for each half, as opposed to an average stress and strain for the whole sample.

For consistency and accuracy, the same 22N load cell was used for all testing procedures on the Bose test instrument. Due to the inelastic nature of the Mepore dressing, it was only possible to strain those samples by 5% before the limit of the load cell was reached. With a more appropriate testing instrument, further testing could be carried out to observe the behaviour of the Mepore up to 25% strain.

The original aim of this investigation was to carry out tensile testing on whole orthopaedic wound dressings to determine how the dressing would behave in a clinical application when applied over a large joint and stretched. An appropriate machine for carrying out these tests was not available for use, so the aims of the project were changed slightly to fall within the limits of, and allow the use of, the Instron ElectroPuls E10000 testing machine. Since small samples had to be taken from each dressing to test, it was instantly made harder to characterise the behaviour of the dressing as a whole.
On completion of data collection and analysis, it was made clear that the Instron ElectroPuls was not sensitive enough for measuring the low loads required to strain the samples (Figures 4.1-4.3). An alternative testing procedure was then to be developed to allow the use of the Bose ElectroForce 3200 test instrument. The maximum crosshead displacement in the Bose is ±6mm. This limited the amount of strain available to apply to the samples, a significant limitation. Due to the highly elastic nature of some of the dressings tested, it was evident that the samples were capable of strains much greater than 25%; an area that could be explored in the future.

6. CONCLUSION

6.1 FINDINGS

The aims of this project were to characterise the mechanical properties of dressings used in the protection of lower limb arthroplasty wounds. The objectives of this project were partially fulfilled; stress response to a prescribed strain was analysed and compared among different dressings, before calculating the mean Young's modulus for each dressing. If time had not been a limitation, the rate of stress relaxation within each dressing would have also been studied.

The results discussed in Chapter 5 indicate that AQUACEL, DuoDERM, Contreet and Biatain, AQUACEL in particular, are significantly less stiff than Mepore, correlating to clinical results, where the use of a dressing featuring a hydrocolloid adhesive is proven to reduce the occurrence of post-operative blistering.

Section 6.2 discusses areas of research that would be beneficial to explore, to better understand the mechanical behaviour of wound dressings.

6.2 FUTURE RESEARCH

This brief section covers future areas of research that, ideally, would have been carried out in this investigation if time and resources had not been limiting factors.

To allow for the mechanical properties of the whole dressings to be characterised, more extensive tensile testing would be carried out on the entire dressing instead of small samples from each dressing. Larger strains would be applied, up to 100% for some dressings, as well as potentially carrying out failure testing, to determine the ultimate tensile stress of the dressings. The results from those tests would be more accurate and applicable for the behaviour of the dressing whilst applied on a patients limb.

Assessing the rate of fatigue for a wound dressing is an important factor in the functionality of the dressing. The ability of the dressing to remain elastic after numerous strains is clinically relevant. Since orthopaedic wounds lie over major joints that have a wide range of motion, a dressing must be able to strain and recoil repetitively, for example, during

walking, where the knee is rhythmically flexing and extending. It would therefore be useful to carry out repetitive testing on dressings to assess the fatigue rate.

As mentioned in section 6.1, the stress relaxation is an important area of research when assessing the behaviour of an orthopaedic dressing. When a dressing is strained, it experiences a magnitude of stress, some of which is transferred to the skin as a shear stress. The rate at which the stress relaxes affects the shear stress applied to the skin. If a dressing relaxes very slowly, the magnitude of the shear stress on the skin will be prolonged, likely contributing to the formation of blisters. The ideal dressing will relax very quickly, keeping the time that the shear force is applied to the skin to a minimum. This research would help select the optimal dressing for clinical use, with the hope to reduce post-operative blister rates.

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APPENDIX A

Table A1: Types of dressing with corresponding sample numbers and cross-sectional area for each individual sample

DRESSING TYPE	SAMPLE NO	CROSS-SECTIONAL AREA (m^2)
AQUACEL (adhesive)	1	0.00002581
	2	0.00002716
	3	0.00002608
	4	0.00002776
AQUACEL (absorbent)	5	0.00008369
	6	0.00008450
	7	0.00008341
	8	0.00007496
	9	0.00007842
AQUACEL (half & half)	10	0.00005005
	11	0.00004993
	12	0.00004968
	13	0.00004693
DuoDERM Hydrocolloid	14	0.00001324
	15	0.00001343
	16	0.00001149
	17	0.00001421
	18	0.00001244
Contreet Hydrocolloid	19	0.00002851
	20	0.00002920
	21	0.00002773
	22	0.00002894
	23	0.00003352
Mepore (adhesive)	24	0.00000830
	25	0.00000848
	26	0.00000791
	27	0.00000778
	28	0.00000843
Mepore (absorbent)	29	0.00003859
	30	0.00003772
	31	0.00004117
	32	0.00003564
	33	0.00003745
Mepore (half & half)	34	0.00002395
	35	0.00002316
	36	0.00002224
	37	0.00002215
Biatain (adhesive)	38	0.00001077
	39	0.00000986
	40	0.00001015
	41	0.00001063
	42	0.00000993
Biatain (absorbent)	43	0.00007177
	44	0.00006869
	45	0.00006972
	46	0.00006856
	47	0.00007102
Biatain (half & half)	48	0.00004066
	49	0.00003918
	50	0.00004067
	51	0.00004041
	52	0.00003974

APPENDIX B

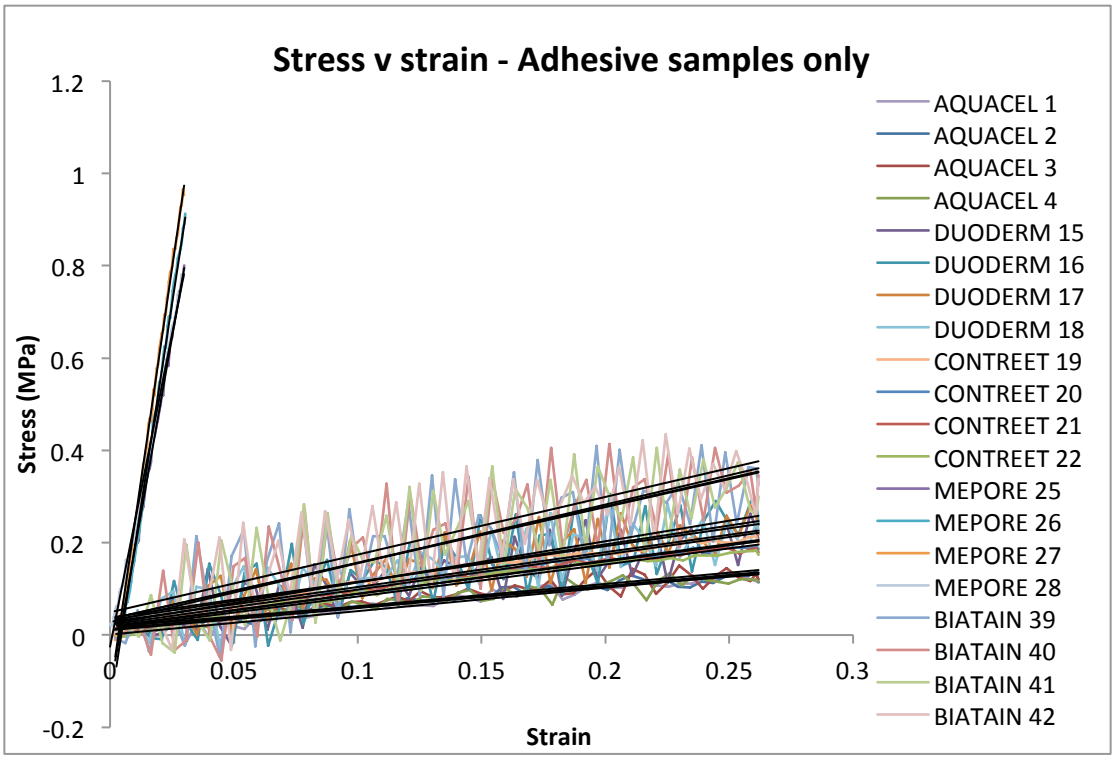

Figure B1: Stress-strain response of adhesive only samples from each of the five dressings and average trendlines to obtain Young's Moduli

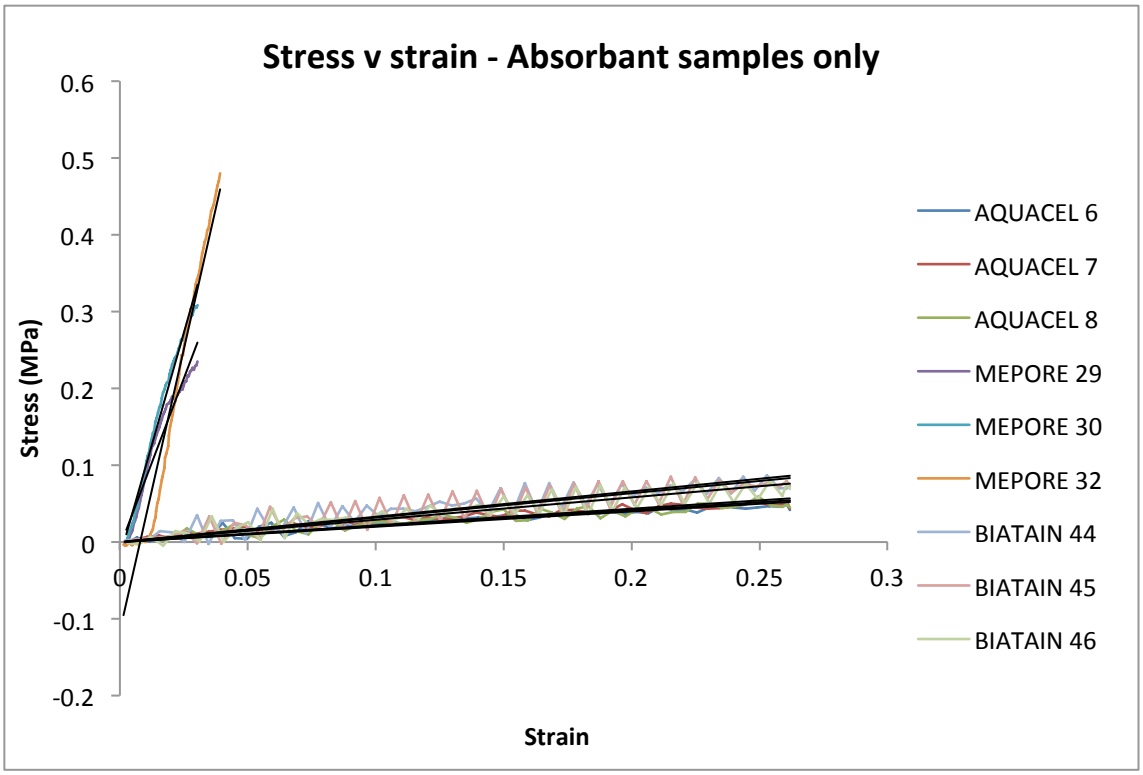

Figure B2: Stress-strain response of purely absorbent samples from Mepore, AQUACEL and Biatain dressings. Average trendlines are present to obtain Young's moduli

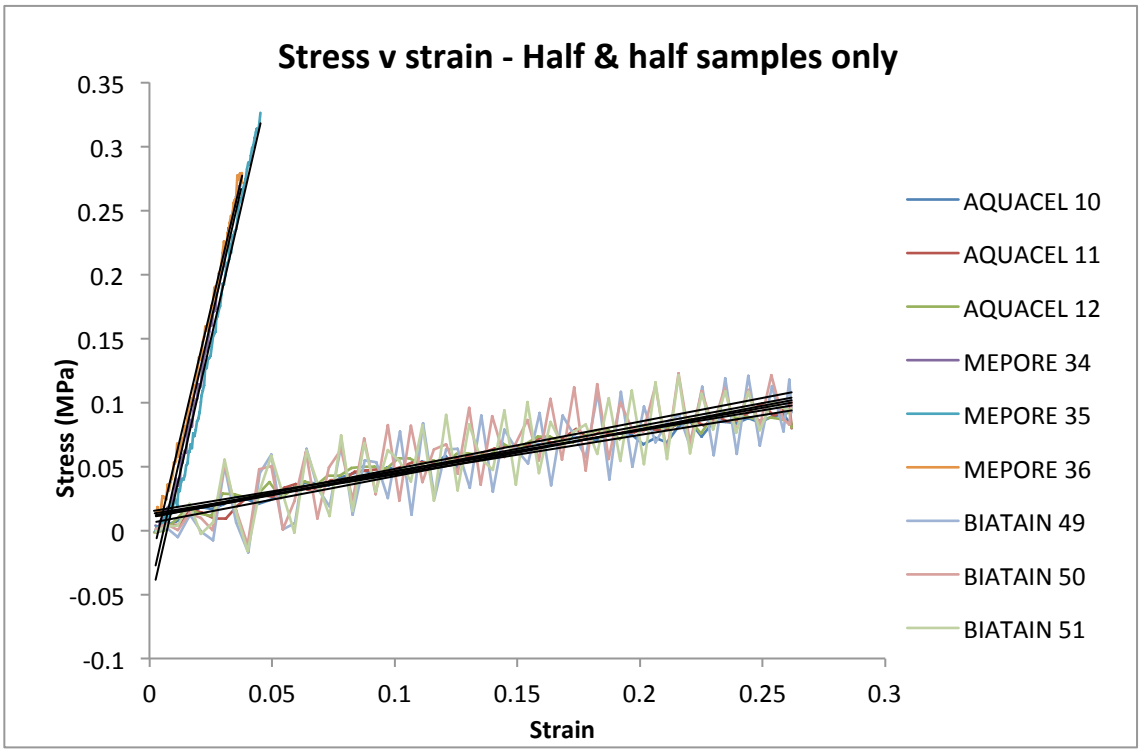

Figure B3: Stress-strain response of samples containing both adhesive and absorbent sections and corresponding trendlines for obtaining Young's Moduli

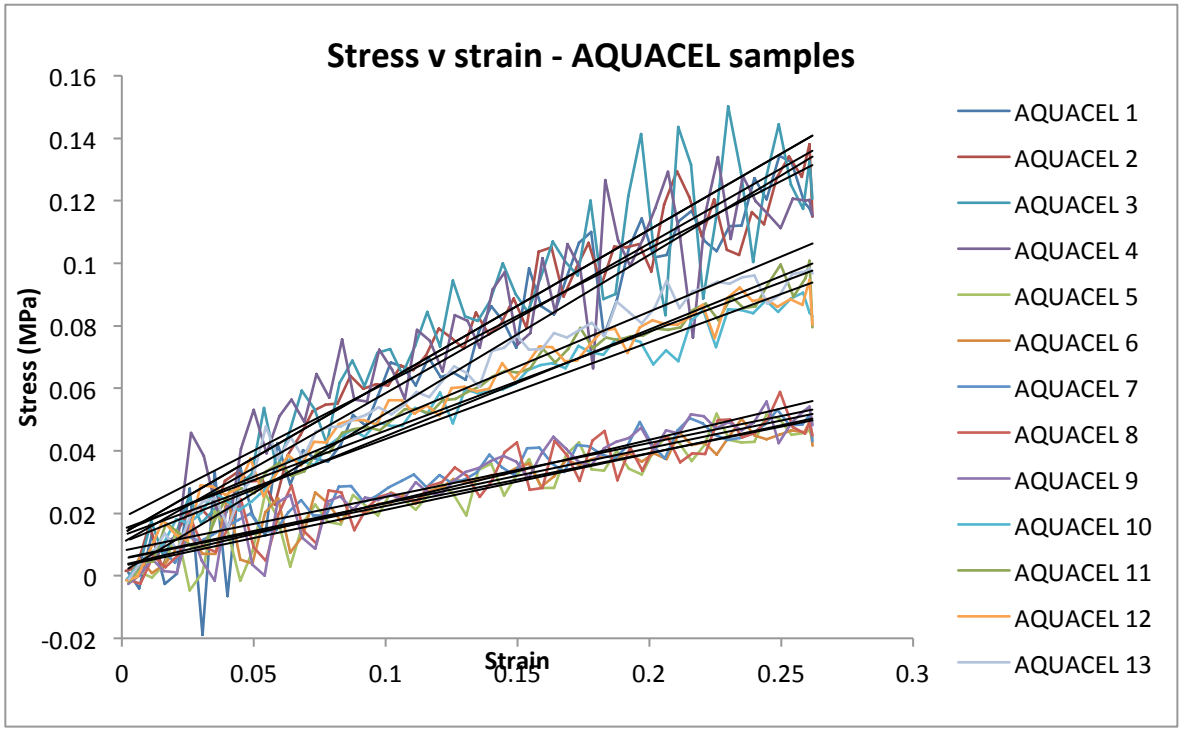

Figure B4: Stress-strain response of the 3 types of sample from AQUACEL dressing and average trendlines to obtain Young's Moduli

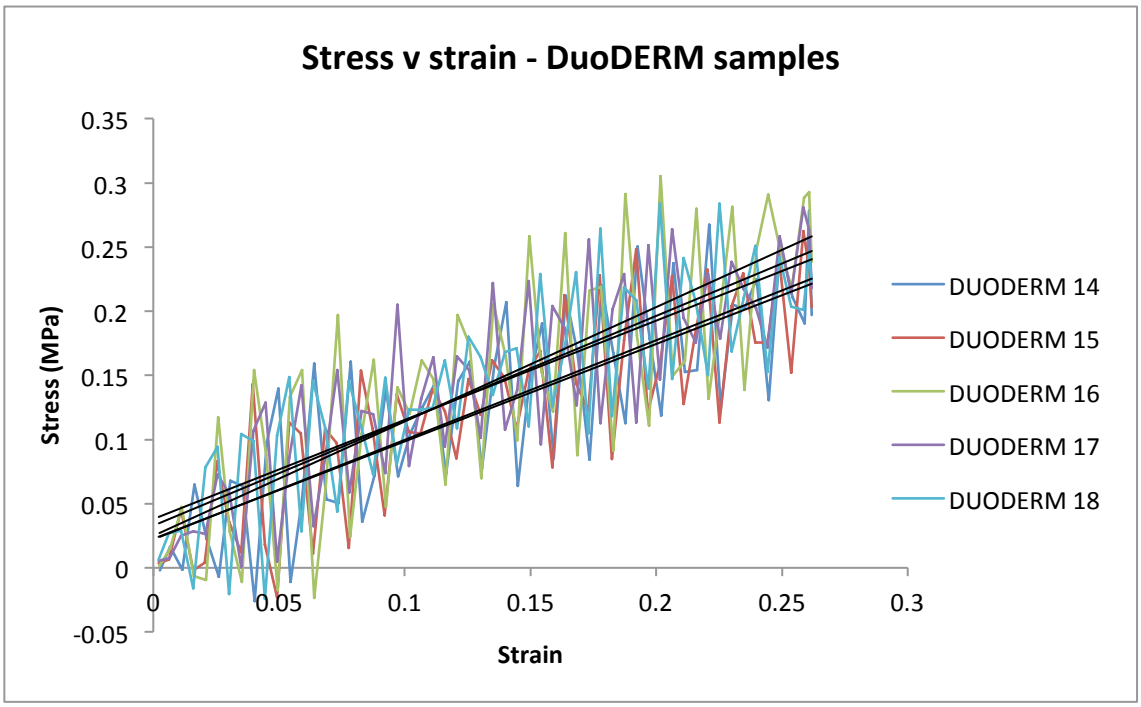

Figure B5: Stress-strain response of samples from DuoDERM dressing and average trendlines to obtain Young's Moduli

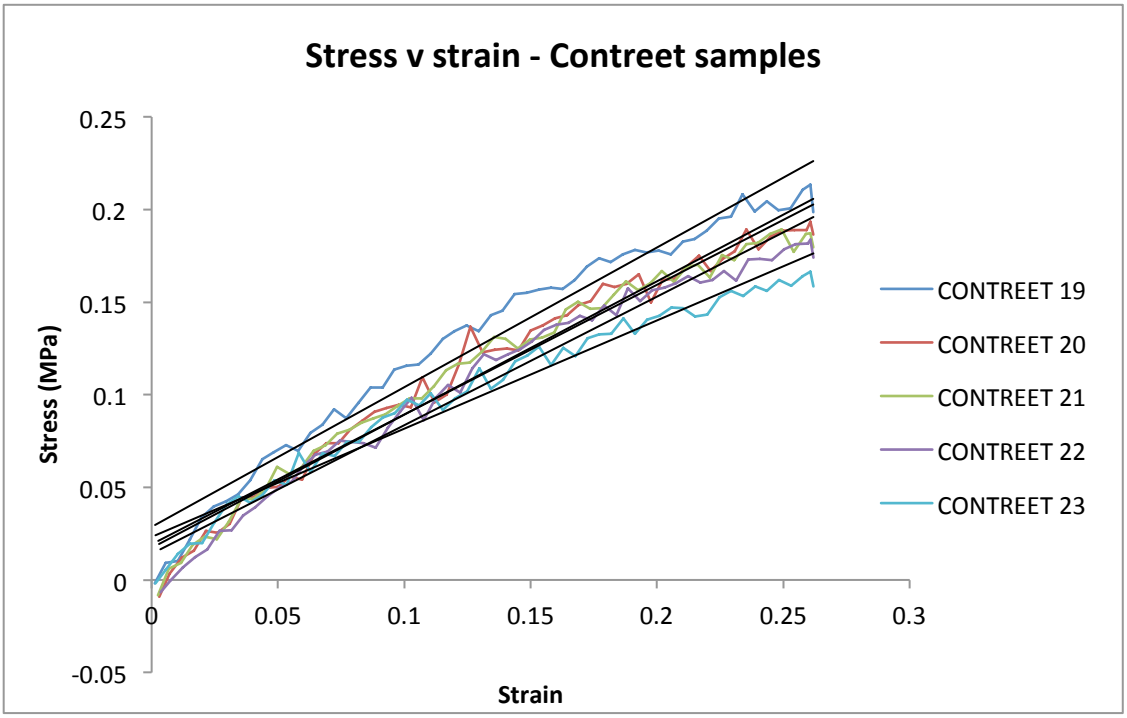

Figure B6: Stress-strain response of samples from Contreet dressing and average trendlines to obtain Young's Moduli

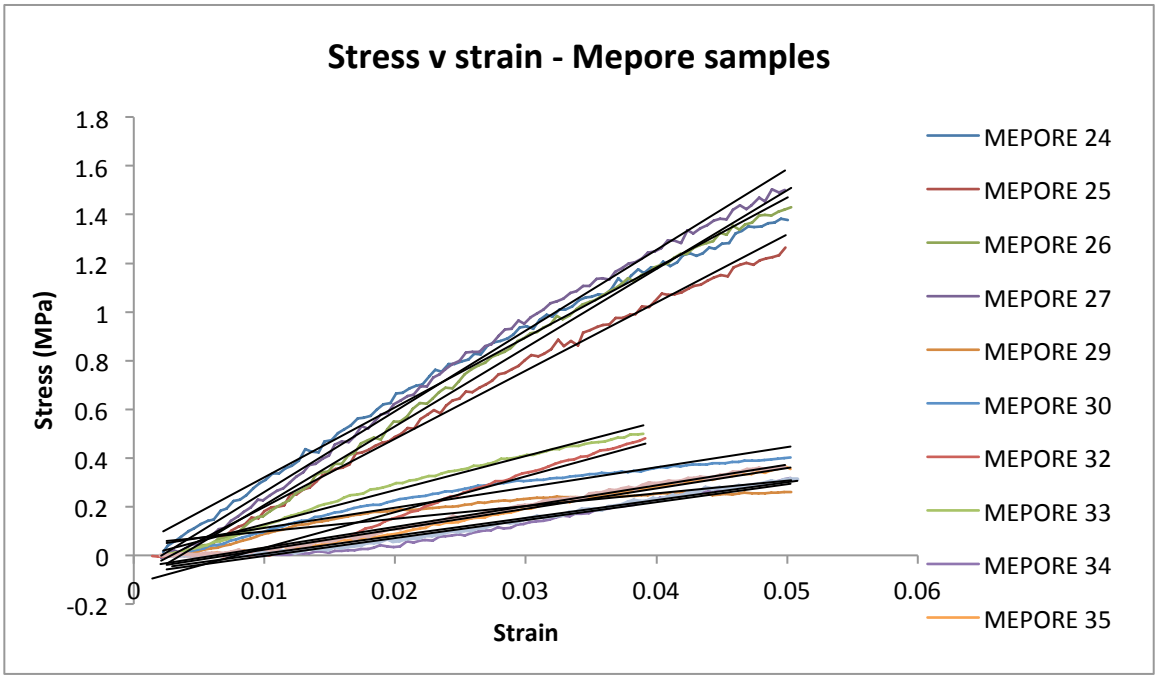

Figure B7: Stress-strain response of samples from Mepore dressing and average trendlines to obtain Young's Moduli

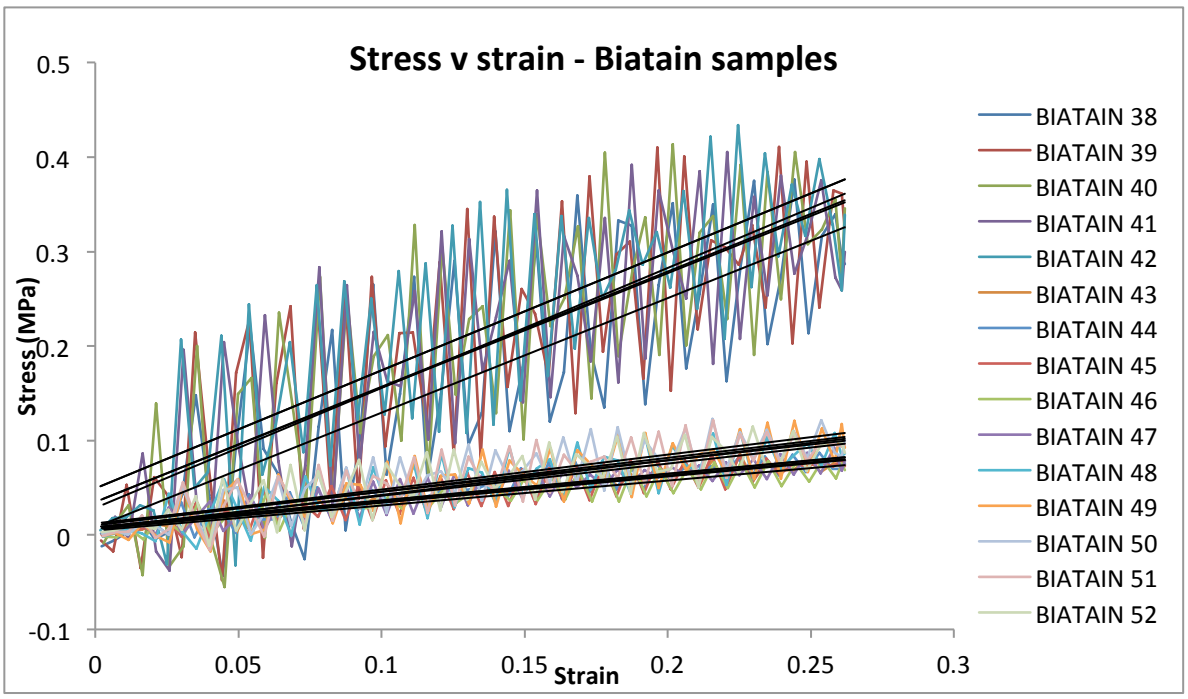

Figure B8: Stress-strain response of samples from Biatain dressing and average trendlines to obtain Young's Moduli

APPENDIX C

Table C1: Equations of trendlines applied to stress-strain graphs

Sample No.	Equation of trendline	Sample No.	Equation of trendline
1	y = 0.508x + 0.0011	27	y = 33.147x - 0.0709
2	y = 0.4792x + 0.0105	28	-
3	y = 0.4868x + 0.0134	29	y = 5.2154x + 0.0459
4	y = 0.4313x + 0.0185	30	y = 8.2982x + 0.0299
5	y = 0.1803x + 0.0031	31	-
6	y = 0.1694x + 0.0054	32	y = 14.684x - 0.1159
7	y = 0.1726x + 0.008	33	y = 14.009x - 0.0122
8	y = 0.1766x + 0.0055	34	y = 7.3689x - 0.0762
9	y = 0.2006x + 0.0034	35	y = 8.3743x - 0.0601
10	y = 0.3095x + 0.0128	36	y = 7.3366x - 0.0663
11	y = 0.3412x + 0.0106	37	y = 8.5169x - 0.0533
12	y = 0.3164x + 0.0148	38	y = 1.2132x + 0.0081
13	y = 0.3523x + 0.0141	39	y = 1.2203x + 0.0349
14	y = 0.7762x + 0.0221	40	y = 1.2705x + 0.0286
15	y = 0.7594x + 0.0226	41	y = 1.2117x + 0.0349
16	y = 0.8914x + 0.0249	42	y = 1.2496x + 0.0493
17	y = 0.8174x + 0.0329	43	y = 0.2785x + 0.0063
18	y = 0.7733x + 0.038	44	y = 0.2827x + 0.0084
19	y = 0.7544x + 0.0287	45	y = 0.2811x + 0.0067
20	y = 0.7199x + 0.0173	46	y = 0.2644x + 0.0047
21	y = 0.7008x + 0.0193	47	y = 0.2833x + 0.0079
22	y = 0.6943x + 0.0141	48	y = 0.3407x + 0.0074
23	y = 0.5846x + 0.0232	49	y = 0.3668x + 0.0057
24	y = 28.711x + 0.0339	50	y = 0.37x + 0.0111
25	y = 27.975x - 0.0804	51	y = 0.3574x + 0.0102
26	y = 32.31x - 0.116	52	y = 0.334x + 0.0123