Application of pressure sensors in monitoring pressure

David Tyler

Tyler, D. Application of Pressure Sensors in Monitoring Pressure, in Hayes, S.G. and Venkatraman, P (eds), *Materials and Technology for Sportswear and Performance Apparel*, Boca Raton, FL: CRC Press, December 2015, Chapter 12, pages 289–310.

Table of contents

12.1 Introduction

12.11 The challenge of measuring pressure

Monitoring pressure distribution using probes and sensors to ascertain the performance of a wide range of products in medical and clothing compression-wear is important for understanding the efficacy of products.

The technology challenge is substantial, because surfaces are 3D contoured and deformable. Textiles can stretch and recover according to their construction and fibre type, and human bodies are covered in skin, below which are various permutations of fat and bone.

Pressure is a term that describes the force applied per unit area. The equation that allows quantitative measurement of pressure is as follows:

$$P = F/A \quad \text{(Equation 1)}$$

Where P = pressure, F = applied force and A = area affected by the applied force.

When an object (like a part of the human body) is in contact with a stretch fabric (a bandage or a compression garment), it experiences a compressive force. According to Equation 1 above, the average interface pressure is the total force divided by the interface area. However, the average pressure is only part of the story. The human body is not a smooth cylinder, but a complex surface of extensible skin under which are soft tissues and rigid bones. Furthermore, stretch fabrics are not simple materials to understand, as they have different stretch properties in different directions and exhibit the phenomenon of relaxation after extension. Consequently, localized interface pressure measurement is necessary to assess the distribution of pressure and to find concentrations of peak pressure.

Pressure measurement technologies are designed to map the location and magnitude of peak pressures and to gain information about pressure gradients across interfaces. To handle exponential increases in information gathered, computerised systems have been developed to analyse the data and provide visual representations of the interface being studied.

For medical products, there are numerous tools used for the measurement of compression. For compression hosiery, the Hatra Mk2A+ Hose Pressure Tester and the Salzmann MST Professional have been developed. For other applications, the Kikuhime tester and the PicoPress® instruments are widely used. These are described in Section 2 (with brief mentions of other technologies).

At a research level, numerous additional sensors have been used for medical products as well as for clothing. The instruments are constantly changing, but emphasis is given in section 3 to the use of Tekscan pressure sensors, including the FlexiForce™ interface pressure sensors.

In medical contexts, where compression is applied frequently to limbs (which have cylindrical body forms), reference is often made to a variant of Equation 1, known as Laplace's Law:

$$P \propto T/R \qquad \text{(Equation 2)}$$

Where P (pressure) is directly proportional to T (tension) divided by R (radius).

This equation is the basis for data processing in the British Standard for compression hosiery (BS 6612, 1985). The medical background for compression bandages and stockings is summarised in Rotsch et al (2011).

Laplace's Law means that the smaller the radius (with constant tension), the higher is the compression pressure. Since the human leg is smaller in diameter nearer the ankle and larger nearer the knee, if bandages are wrapped at a constant tension, there will be a pressure gradient (known as graduated compression) with maximum pressure at the ankle and reduced pressure towards the knee. This graduated compression is considered to accelerate the venous flow rate, with medical benefits to the patient.

Equation 2 also suggests a potential problem when the radius is small. A pressure measuring device that has a thickness of a few millimetres has the potential of distorting the radius locally, thereby distorting the compression pressure locally. Questions have been raised about the accuracy of some instruments because of this effect.

12.12 Units of Pressure

Pressure is defined as Force divided by Area (with the Laplace Law being a special case of this). The international system (SI units) recognised the pascal as the unit of pressure. Physicists have defined one pascal (Pa) as the pressure exerted by a force of one newton applied over an area of one square metre. The SI unit of pressure honours Blaise Pascal as a pioneering 17th Century French scientist who made significant contributions relating to understanding pressure.

One pascal represents a low pressure, and there are many applications where other units are deemed more appropriate, sometimes for historical reasons. There are numerous metric and imperial units that were in common use before SI units were defined, and they continue to be employed. Examples of metric units are kilograms force per square metre (kgf/m^2) or grams force per square cm (gf/cm^2). An imperial unit of pressure is pounds per square inch (psi). Some important additional units of pressure in common use are: torr, mmHg and bar.

The torr is a unit honouring the 17th Century Italian physicist Evangelista Torricelli, who invented the mercury barometer and was the first to explain the concept of atmospheric pressure. He found that the column of mercury in a barometer positioned at sea level measured 760 mm. 1 torr is the pressure needed to sustain 1mm of mercury (Hg) in a barometer, so 1 torr is 1 mmHg. Most pressure-measuring medical instruments are calibrated in mmHg units. 1 torr is approximately 33 pascals.

One bar represents the mean atmospheric pressure at sea level. It is common to use this unit when referring to the pressure of water at depth (with reference to diving, for example). It is now defined as 100 kPa. Meteorological charts normally use hectopascals (hPa), where 1 hPa = 100 Pa and 1 bar = 1000 hPa.

12.2 Pressure sensors for medical applications

12.21 Compression Hosiery: the Hatra hose pressure tester

During the 1970s, a tool for measuring the properties of compression hosiery was developed by Derek Peat at the Hosiery & Allied Trades Research Association (Hatra, Nottingham, UK). The garment is stretched lengthways and widthways in a defined manner in a range of sizes. A measuring head utilises a strain gauge to record the compression provided at any position from the ankle upwards. The head has a rectangular plate (25 mm wide) that is pushed onto the stretched hose and the resistive forces are recorded. The equipment provides reproducible test data and was incorporated into British Standard 6612 in 1985. The Hatra tester was also adopted by two other British Standards: BS 7672:1993 and BS 7563:1999. The MK2 Hatra was available before 1990, after which the Mk2A was released, allowing tailored leg profiles to be easily added. The current model is the Mk2A+, illustrated in Figure 12.1.

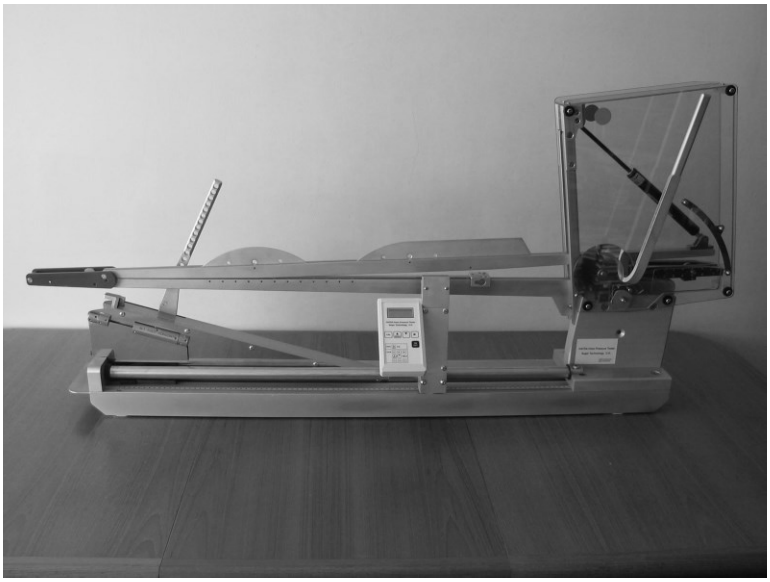

FIGURE 12.1
The Hatra Mk2A+ hose pressure tester. (Courtesy Segar Technology.)

12.22 Compression Hosiery: the Medical Stocking Tester

In 1977, the first Medical Stocking Tester (MST) was launched by Dr. A.A. Bolliger in Switzerland. The concept is similar to the Hatra tester. The main difference is that the compression stocking is placed on a leg-shaped former and a flat measuring device (40 mm wide, 0.5 mm thick, linked to an air pump and a pressure transducer) is used to quantify the compression forces. A separate former is needed for each size to be tested. This tool has also been developed over time, and the current model is Mk V. Alongside this, the MST Professional has a variable leg form that is claimed to cover 95% of known leg sizes. The measuring probe has the capability of measuring the compression exerted by stockings worn by live subjects, which means it can be used additionally as a research tool. The instrument is illustrated in Figure 12.2 and an example of its use in research for both *in vivo* and *in vitro* measurements is provided by Liu et al. (2013).

FIGURE 12.2
Salzmann MST Professional. (Courtesy Swisslastic AG.)

12.23 The Kikuhime tester

The sensor used in the Kikuhime instrument is an oval polyurethane balloon containing a 3-mm-thick foam sheet. This is connected to a syringe (for changing the air pressure within the balloon) and a measuring unit (with the pressure transducer). Testing starts by adjusting the syringe so that the balloon is at atmospheric pressure, and then it is placed in position (for example, between the leg and the compression garment). The transducer monitors the pressure experienced by the balloon and the output is a digital display (Figure 12.3). An assessment of the reproducibility and reliability of measurement was undertaken by Brophy-Williams et al. (2014), who concluded that the tester was suitable for use with sports compression garments.

FIGURE 12.3
The Kikuhime tester. (Courtesy TT Meditrade and mediGroup Australia Pty Ltd.)

12.24 Overview of other test instruments

Numerous other instruments can be found in the literature and there are many new ideas coming to the fore each year. A survey of the field was undertaken at an international consensus meeting of medical experts and representatives from industry held in January 2005 in Vienna, Austria. In vivo measurement of interface pressure was considered highly desirable, and methods for measuring the interface pressure were considered with that in mind. Table 12.1 presents the different technologies considered.

Table 12.1 Types of Interface Pressure Sensors

Pneumatic, Pneumatic-Electric or Pneumatic-Piezoelectric
Oxford pressure monitor (Talley Ltd, Ramsey, Hampshire, UK)
Talley pressure evaluator (Talley Ltd)
MST MKIII Salzmann (Salzmann Medico, St Gallen, Switzerland)
Digital interface pressure evaluator (Next Generation Co., Temecula, CA)
Scimedics pressure evaluator pad (Vista Medical, Winnipeg, MB, Canada)
Kikuhime (Meditrade, Soro, Denmark)
Juzo tester (Elcat, Wolfratshausen, Germany)
Sigat tester (Ganzoni-Sigvaris, St. Gallen, Switzerland)
Piezoelectric
MCDM-I (Mammendorfer Inst. Physik, Munich, Germany)
Fluid Filled, Fluid Filled Resistive
Strathclyde pressure monitor (University of Strathclyde, Scotland)
FlexiForce (Tekscan, South Boston, MA)
Skip air pack analyzer (AMI Co., Japan)
Resistive and Strain Gauge
FSR, FSA (Vista Medical, Winnipeg, MB, Canada)
Fscan, Iscan (Tekscan, South Boston, MA)
Rincoe SFS (Rincoe and Associates, Golden, CO)
MCDM (Mammendorfer Inst. Physik, Munich, Germany)
Fontanometer (Gaeltec Ltd, Dunvegan, Isle of Skye, Scotland)
Diastron (Diastron Ltd, Andover, Hampshire, UK)
Capacitive
Kulite XTM190 (Kulite Semiconductor Products, Leonia, NJ)
Precision (Precision Measurement Co., Ann Arbor, MI)
Xsensor (Crown Therapeutics, Belleville, IL)
Pliance (Novel, Munich, Germany)

Source: Partsch, H. et al. 2006. *Dermatologic Surgery* 32 (2): 227.

It is necessary to point out that many of these instruments are no longer available. There are many design concepts that have been explored, but finding commercial viability has not been easy. The first two entries listed in Table 12.1 are described below, as these have had a more significant impact in the literature. The Talley Group designed pressure sensors in the 1980s but production ceased around the year 2000. The company no longer offers instruments to measure pressure, but has concentrated on producing mattresses, cushions and other products providing pressure relief and compression therapy.

The Oxford Pressure Monitor MkII
The instrument was designed to monitor pressures between skin tissue and support media for chair or bed bound individuals. The name originates because of collaboration with the Oxford Orthopaedic Centre. Multiple sensors were used to enable the simultaneous monitoring of pressures below a reclining patient. Each sensor was constructed from two thin plastic sheets that could be inflated by a pulse of air, allowing a measure of the compression forces. The MkII had 12 sensors, and a later device, the MkIII, was equipped with 96 sensors and also marketed as the Talley IPM (Interface Pressure Monitor).

The Talley Skin Pressure Evaluator
The Talley SD.500 Skin Pressure Evaluator was designed as a portable device for checking tissue pressures in medical wards, wheelchair pressures and any scenario where tissue trauma is an issue. There were two main parts: a hand-held control unit with a digital display of pressure and a balloon type sensor that could be inflated manually. The sensor contained platinum wires on both sides of its inner surface. To obtain a measurement, the sensor was placed between the skin and clothing before inflating using the pump bulb. As the sensor inflated, the electrical contact between the two sets of platinum wire was broken. As air was allowed to flow out of the sensor, the platinum wires touched and the circuit was reconnected. At this point, the pressure was recorded and displayed on the control unit.

12.25 Evaluation of pressure sensing instruments

A detailed comparison of three instruments was undertaken by Flaud et al. (2010). They selected the Salzmann, Talley SD.500 and Kikuhime testers and compared their performance in terms of accuracy, repeatability and sensitivity to flexion on a curved surface. The first set of tests utilised a chamber that could be preset to defined pressures. The second set used a wooden leg model and inserted the sensors between the leg and compression stockings of known pressure. The results for the pressurised chamber tests are reproduced in Figure 12.4.

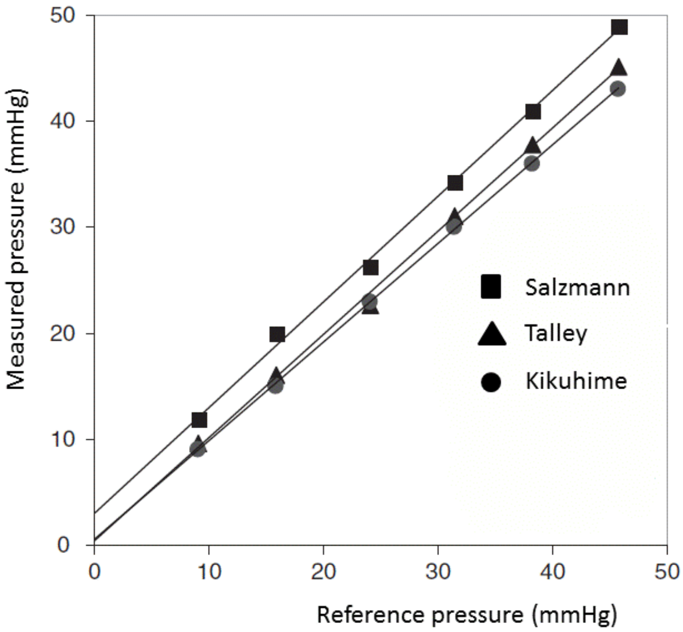

FIGURE 12.4
Measured pressures versus reference pressures for the three sensors in the pressurised chamber. (From Flaud, P. et al. 2010. *Dermatologic Surgery* 36 (12): 1930–1940.)

The equations of lines fitted to the data points are as follows:

Salzmann: $y = 0.99x + 3.32$ ($R^2 = 0.99$)
Talley: $y = 0.97x + 0.47$ ($R^2 = 0.99$)
Kikuhime: $y = 0.93x + 0.57$ ($R^2 = 0.99$)

The researchers summarised their results in this way:

> "In a pressurized chamber, the three systems gave linear responses and an overall error of 15.4%, 3.1%, and 4.3% for Salzmann, Talley, and Kikuhime, respectively. The repeatability error was less than 0.6mmHg. On the leg model, the overall errors differ between the systems. Repeatability was comparable between the sensors." (Flaud et al. 2010, p.1930)

This shows the sensors were capable of providing useful tools for medical practitioners. However, care must always be taken with stockings and bandages around legs and arms because material variability and user factors may introduce variability that is difficult to control.

Table 12.2 has some generalised comments on the advantages and limitations of different types of sensor.

Table 12.2 Some Advantages and Disadvantages of Sensors

	Advantages	Limitations
Pneumatic transducers	Thin and flexible probes; cheap, easy, and handy	Dynamic measurement is only possible with additional special equipment; sensitive for temperature and hysteresis
Fluid filled	Flexible; dynamic measurements	Thick when filled; problems during motion
Resistance	Thin sensors; dynamic measurement	Sensitive to curvature; stiff and thick; not useful for long-term measurements

Source: Partsch, H. et al. 2006. *Dermatologic Surgery* 32 (2): 227.

12.26 The PicoPress® instrument

With the passing of time, the instruments listed in Table 1 require extensive editing. There are both additions and deletions. In particular, a new product designed for medical applications is worthy of note: the PicoPress® produced by Microlab (Padua, Italy). The instrument has a manometer connected to a probe: a flexible circular plastic bladder (5 cm diameter). The bladder is placed in position in the deflated state and the bandage/compression garment is applied. To take measurements, the operator pushes on an embedded syringe to introduce 2 cubic centimetres of air to the bladder. The resultant expansion in thickness is constrained by the compression exerted by the bandage/garment and the manometer is used to record the pressure. Both static and dynamic measurements are possible, so the compression can be determined when resting, when walking and when standing. After collecting data, the probe is deflated and left in position until further readings are required. The PicoPress® is shown in Figure 12.5. Partsch & Mosti (2010) evaluated this instrument by comparing its performance with two other commercial systems. They concluded: "The results suggest that the Picopress® transducer, which also allows dynamic pressure tracing in connection with a software program and which may be left under a bandage for several days, is a reliable instrument for measuring the pressure under a compression device."

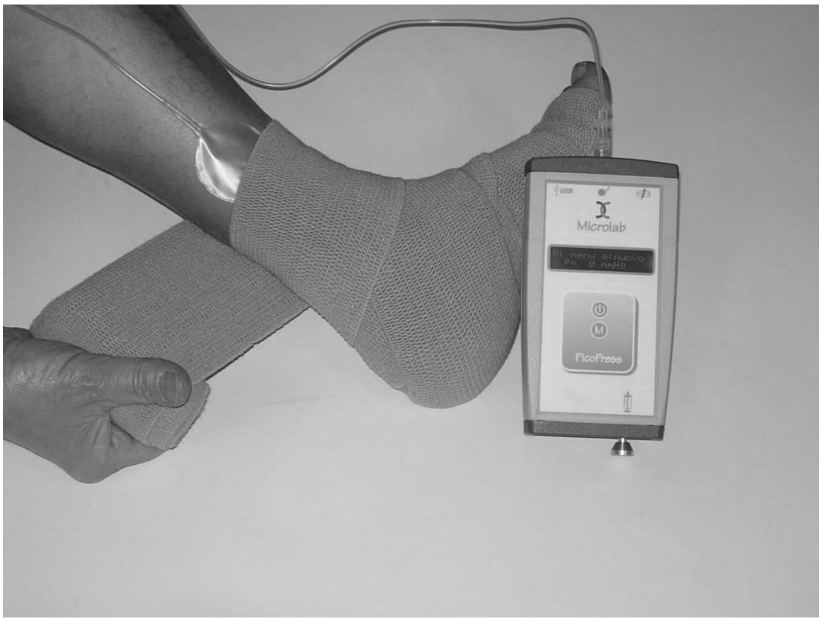

FIGURE 12.5
The PicoPress instrument. (Courtesy Microlab Elettronica.)

12.3. Pressure sensors for clothing applications

12.31 Use of medical instruments

Several of the medical instruments discussed above have been used for measuring compression provided by clothing.

Chan and Fan (2002) used the Tally SD.500 skin pressure evaluator to clarify the relationship between subjective tightness sensation and the clothing pressure of girdles. This was followed by modelling work to predict the pressure of girdles on the human body (Fan and Chan, 2005).

Pressure garments are widely used in the treatment of skin damage caused by burns. Giele et al. (1997) expressed concern that devices like the Talley and Kikuhime testers were inadequate for three reasons:

1. Distortion of the garment, hence raising garment tension and increasing the pressure generated.
2. Poor conformity of the device to the skin.
3. Being unaware to what degree external pressure is transmitted to and through the skin.

Consequently, they adapted a method of probing the subdermal cutaneous pressure using hypodermic needles connected to a continuous low flow pressure transducer. They found that pressure garments have the effect of increasing pressure subdermally. They confirmed that the garment was responsible for controlling compression through and within the skin and were able to quantify the effects.

Another custom-built instrument to assess pressures obtained from garments worn for scar treatment was constructed by Teng and Chou (2006). This was based on an "air pack" sensor similar to those used in the Talley and Kikuhime testers.

However, despite this interest in customised instruments, research by Van den Kerckhove et al. (2007) into burned skin treatment went back to using the Kikuhime tester. They tracked reductions in pressure with time associated with different fabric constructions and concluded "that the Kikuhime pressure sensor provides valid and reliable information and can be used in comparative clinical trials to evaluate pressure garments used in burn scar treatment."

Another custom-made instrument has been called the "Textilpress" (Maklewska, et al. 2007). This has been designed to measure pressures exerted by compression bands, manufactured from knitted fabrics, on a cylinder surface of defined diameter. Its role is to test compression away from a human wearer, and it is not suitable for in vivo measurement. The device is based on tensometric sensors, to measure both the compression exerted by the fabric and the diameter of the cylinder.

12.32 Tekscan technologies (I-scan system)

The I-scan system provides ultra-thin (0.15 mm) sensors of varying sizes for measuring compression forces. These sensors are formed from two sheets of thin polyester, each coated with linear electrical conductors and enclosing a pressure-sensitive material. The electrical conductivity of the sandwiched interlayer material changes linearly in response to applied pressures. The array of linear conductors on the upper sheet are at 90 degrees to the array on the lower sheet. This creates a matrix of sensing locations (sensels) that is determined by the geometry of the sensor. Two examples of the many sensors available are shown in Figure 12.6.

FIGURE 12.6
Two sensors supplied by Tekscan (model 5250 and model 4201). (Source: David Tyler.)

The smaller of these two sensors is Model 4201, with a matrix width of 45.7 mm and a matrix height of 21.0 mm. The matrix itself is made up of 24 columns and 11 rows, making 264 sensels (see Figure 12.7). This gives a sensel density of 27.6 sensels/cm^2. The larger sensor is connected to a sensor handle (which carries the signals to a computer) and is Model 5250. This is a square sensor with both matrix width and matrix height being 245.9 mm. The matrix itself is made up of 44 columns and 44 rows, making 1,936 sensels.

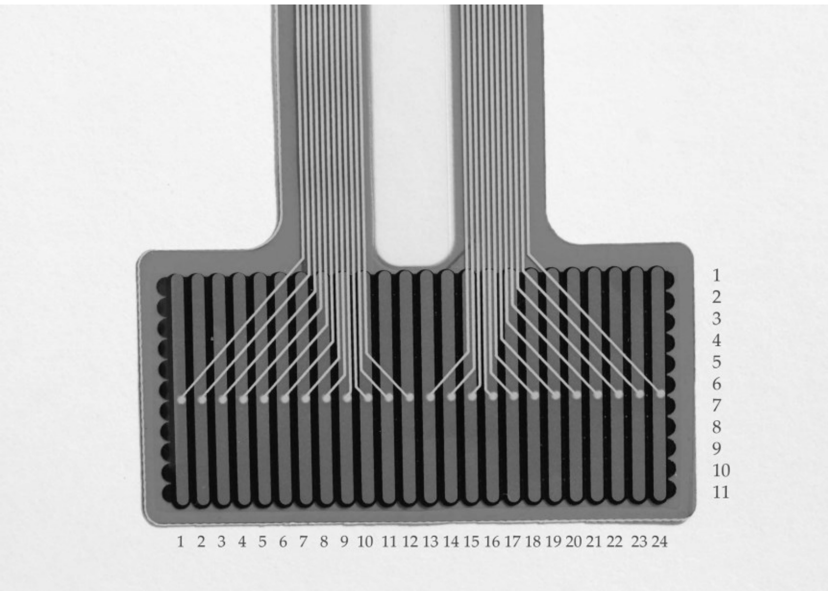

FIGURE 12.7
The matrix of model 4201. (Courtesy Sheena Tyler.)

An example of a screen plot of compression data using Model 5250 is in Figure 12.8.

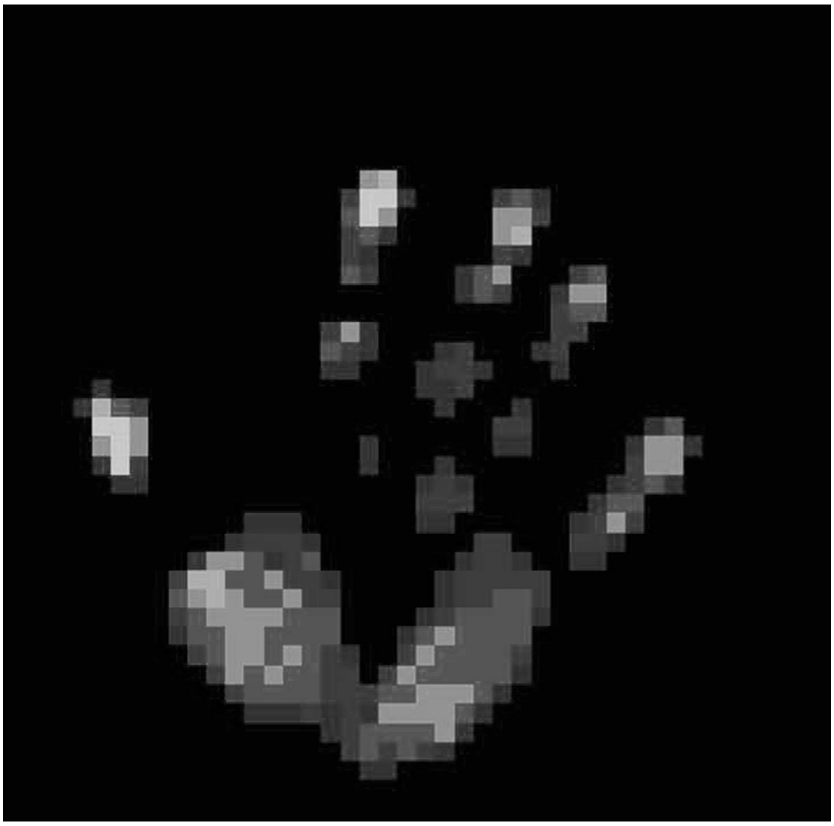

FIGURE 12.8
Screen display showing sensels recording different pressures of a human hand against a flat surface. The sensels are colour coded, with black/deep blue representing the lowest pressures and, as higher pressures are recorded, the colours move through the spectrum to red (maximum pressure). In this black-and-white illustration, sensels with lighter shades of grey are recording higher pressures.

A review of technologies available in 2005 and suitable for measuring interface pressures relevant to the formation of pressure sores in patients was produced by Swain (2005). Six commercial systems were considered, including Tekscan and Tally. Tests with Tekscan found evidence for significant hysteresis and creep in the data output, but the author noted that clinicians preferred this system because of its real-time display capabilities, resolution and display options.

A detailed investigation of the calibration issues affecting one particular I-scan sensor was undertaken by Macintyre (2011). The paper contains an overview of the measurement inaccuracies affecting many instruments, but there is a recognition that Tekscan sensors and the supporting software has become “widely used in recent years”. The range of interface pressure of interest is between 6 and 50 mmHg, which are “often at the lower end of the measurement range of commercial pressure sensors”. Using the Tekscan 9801 sensor, a very detailed evaluation was undertaken of standard calibration procedures to quantify the accuracy of test results. These are referred to as the “2-Point power law calibration” and the “Linear calibration” procedures. A difference of 3% between measured and applied loads was considered unsatisfactory.

> "Despite considerable effort and many attempts to calibrate these sensors the results were disappointing and unsatisfactory. The 2-point power law calibration was most accurate in the middle of the calibrated range, while the linear calibration was most accurate towards the top of the calibrated range (and was completely inaccurate at low applied loads). [. . .] This level of variability was unacceptably high for precise product development work so another method of calibration was sought." (pp.1176-7)

The rest of this paper is concerned with a revised calibration procedure and additional probing of sources of error. The findings are presented in a table of the mean differences between measured and applied pressures, where the largest difference is 2.1 mmHg. The results were shown not to be time or use dependent and the new method was adopted for "the accurate measurement of pressures delivered by pressure garments (and compression bandages)". This calibration method was used to evaluate design solutions for pressure garments used in the treatment of hypertrophic burn scars (Macintyre, 2007).

Brorson et al. (2012) used Tekscan technology (the I-scan® system) for in vitro measurements on compression garments for treating lymphedema. Their aim was to define a protocol for gaining quantified data relating to compression hosiery. Garments from three manufacturers were selected; wear and tear was simulated by washing the garments before putting them on plastic legs every day for 4 weeks. Whilst there were differences between garments from different manufacturers, no difference was found between garments from the same manufacturer. During the trial period, decreases of subgarment pressure were not observed. They concluded that "Tekscan pressure measuring equipment could measure subgarment pressure in vitro."

An example of Tekscan sensors being used in sports science is provided by Pain et al. (2008). The authors set out to measure in vivo impact intensities during enacted front-on tackling in order to assess the effectiveness of rugby shoulder padding for reducing peak forces experienced by players. The work reported limited benefits from using shoulder pads, but raised a number of issues about the selection and use of sensors:

> "These potentially inaccurate force measurements exposed three issues with the Tekscan sensor. Firstly, the sensor area was too small as large forces were generated up to and beyond the sensor boundary. Secondly, despite performing a dynamic calibration, the sensor's method of multiplexing data within a long sampling window (4 ms) may have caused the total impact peak force to be missed. Thirdly, the sensor has a low dynamic response time. [. . .] These issues can be alleviated with the use of a larger sensor that employs a higher sampling frequency. A further limitation is the inability to measure shear force and the fact that the sensors will produce a response if creased or curved too acutely." (p.862)

These reflective comments are noted here as they show that experiment design considerations have to be addressed carefully, so that the instruments used are capable of delivering useful results. Sometimes it is necessary to analyse activities in terms of several elements, and then focus attention on those elements separately. This has been a way forward for the analysis of forces on rugby players. Usman et al. (2011) looked specifically at the forces in tackling, using a tackle bag equipped with four Tekscan sensors. Participants were asked to tackle the bag in four different ways: (1) dominant side, (2) non-dominant side, (3) dominant side with shoulder pads, and (4) non-dominant side with shoulder pads. With repeated tackling, an assessment of the variability of the forces experienced by participants was gained.

12.33 Tekscan technologies (FlexiForce®)

Another type of Tekscan sensor is a single element force sensor with the brand name FlexiForce®. There are similarities between the construction of force sensors and pressure sensors but, instead of a matrix of sensels, the resistive layer uniformly covers the whole area of the sensor. Force sensors do not map pressure distributions but provide feedback about the aggregated force experienced. Inevitably, data acquisition and analysis are simplified. Applications for these sensors have been found in sportswear research (Lin et al. 2011, Lin et al. 2012), modelling the compression effects of high-performance sportswear. The FlexiForce® sensors provided empirical data to compare with the simulation:

> "Seven FlexiForce® A201 force sensors (Tekscan, Inc., USA) were placed at seven important muscles that flex or extend when running, namely: (i) vastus lateralis (VL); (ii) vastus medialis (VM); (iii) rectus femoris (RF); (iv) tibialis anterior (TA); (v) semimembranosus (SE); (vi) gastrocnemius lateralis (GL); (vii) gastrocnemius medialis (GM)." (p.1472)

Undoubtedly, new interface pressure measurement systems will emerge, and there are numerous others that have been reported in the literature that do not appear in this review. There is a problem with custom-made instruments and with commercial products that have a short market life: there is simply not the time to assess the reliability and reproducibility of these instruments and to build a knowledge base to achieve "best practice". Consequently, the literature considered in this chapter has been selected to stimulate thoughts on experimental design and helping research aims to be achieved by an appropriate choice of sensors.

12.4 Discussion of Laplace's Law

Rotsch et al. (2011) point out that the compression pressure exerted by a bandage is dependent on four factors:

- The type of bandage, particularly its elasticity
- The pre-stretching applied during application
- The number of layers of bandage
- The state of the bandage as it is in use

Laplace's Law means that the applied pressure is directly proportional to the tension in a bandage but inversely proportional to the radius of curvature of the limb to which it is applied. This has immediate relevance to the selection of the material to bandage the limb. When an inelastic bandage is applied, the tension in the material tends to be low and the contact pressure when resting tends to be relatively low. However, when moving about, the limb expands as muscles contract, and the tension in the bandage tends to increase rapidly so that the resultant applied pressure is high. By contrast, the use of elastic bandages will result in more even compression whether resting or exercising. This leads Rotsch et al. (2011) to refine the conceptual model by distinguishing between static pressure and operating pressure. The static pressure is effectively the pressure when the bandage is applied to relaxed tissue. The operating pressure results from changes in the volume of muscles during movement.

Clearly, Laplace's Law does not provide a comprehensive mathematical model of compression pressure.

It is also necessary to point out that the human leg has a complex shape, and is not well represented either as a cylinder or a cone. There are solid bones covered by various types of soft tissue, and there are many permutations depending on the individuals being bandaged. Deformation of the skin may vary significantly when considering different parts of the leg. This raises further questions about the application of Laplace's Law and its relevance to compression garments, whether for medical purposes or for sport.

Thomas (2003) referred to the widespread recognition of the Laplace equation, but pointed out that it has not been well understood. He made reference to a book he published in 1990 that set out a version of the equation that would be more useful to practitioners. He wrote: "it is also necessary to consider two further factors: the width of the bandage and the number of layers applied. Although these variables may not appear initially to form part of the original Laplace formula, they are essential to obtain an accurate value of tension." (p.22). The modified equation used units selected because they are familiar to practitioners and incorporated bandage parameters.

$$\textit{Pressure (mmHg)} = \frac{\textit{Tension (kgf)} \times n \times 4620}{\textit{Circumference (cm)} \times \textit{Bandage Width (cm)}} \qquad \text{(Equation 3)}$$

Where n = the number of layers applied.

The goal of bandaging is to achieve graduated compression, with the highest pressures close to the ankle and the lowest pressures closest to the knee. However, despite great care being taken to apply bandages correctly, there has been an ongoing problem of demonstrating graduated compression. Schuren & Mohr (2008) drew attention to the various explanations that have been proposed to explain the problem: poor operator technique, poor measurement technique, and the difficulty of maintaining constant tension during application.

With growing scepticism, Schuren & Mohr (2008) reviewed 3 detailed studies of graduated compression by standardising on the leg shape. Study 1 involved 32 experts, 4 commercial compression bandage systems and an artificial leg fitted with three Kikuhime pressure sensors. The participants were asked to repeat the bandaging exercise 3 times for each commercial system. Study 2 was an evaluation of a commercial prototype compression bandage, using 3 experienced orthopaedic technicians, and an artificial leg with 6 strain-gauge force transducers. Each participant applied 40 bandages to the artificial leg. Study 3 used the same leg but selected a different compression bandage, and 8 nurses comprised the expert practitioners. Altogether, these studies yielded a database of 744 sets of data relating to graduated compression.

Using the Laplace formula, theoretical compression pressures were calculated. These all showed the desired graduated compression, mostly in the range 30-60 mmHg (Figure 12.9).

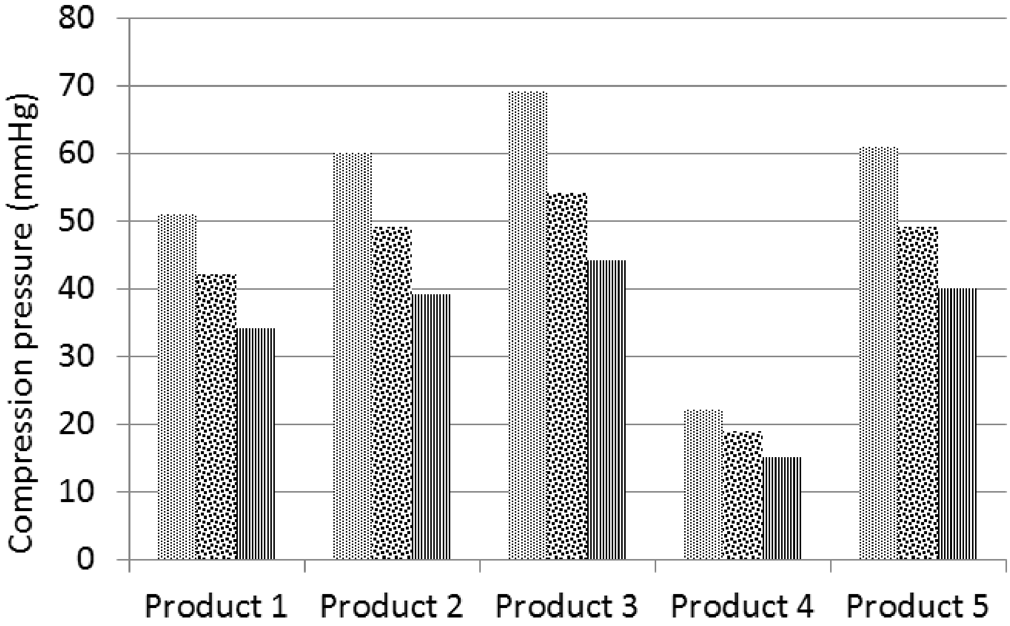

FIGURE 12.9
Theoretical compression according to Laplace's law. (Based on Schuren, J. and Mohr, K. 2008. *Wounds UK* 4 (2): 38–47.)

When comparing theoretical with measured values, there was a marked disparity. First, the experimental results typically showed that less than 10% of the bandages applied achieved graduated compression. Furthermore, the measured values were consistently lower than theoretical values. The aggregated test results are plotted in Figure 12.10.

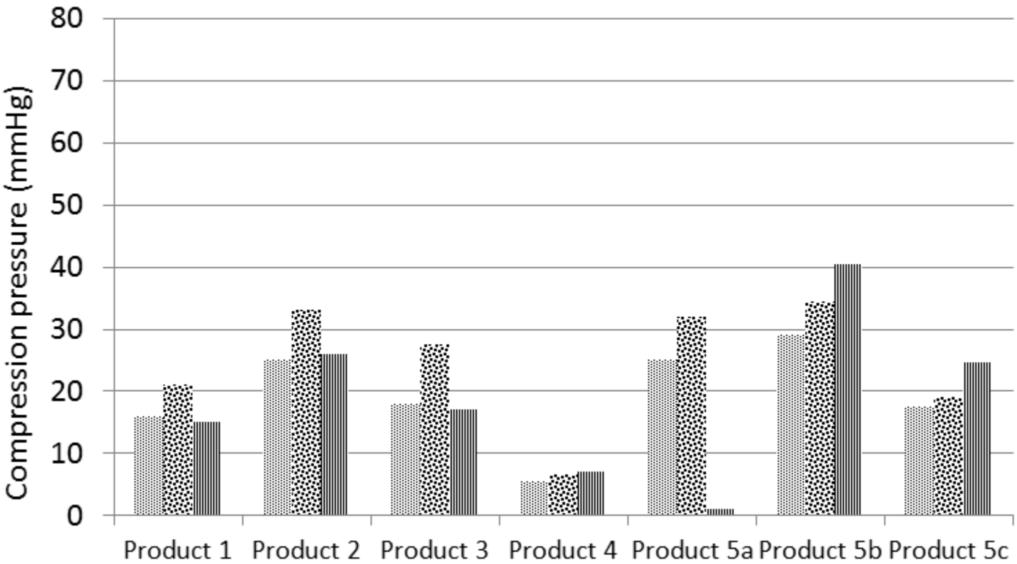

FIGURE 12.10
Measured mean pressure values from all studies. (Based on Schuren, J. and Mohr, K. 2008. *Wounds UK* 4 (2): 38–47.)

Schuren & Mohr (2008) discuss their findings critically, acknowledging the problems of working with artificial legs and in a laboratory (rather than in a clinical) environment.

However, they question the widespread belief in the usefulness of the Laplace equation and also the assumption that graduated compression is the norm.

> "It is true to say that these studies should have produced data that presented Laplace's law in its best light as the environment, subject, and bandagers were well controlled. However, the pressure calculations made using the modified Laplace's law equation did not accurately predict the pressure values found in these three studies. In fact, true graduated compression was observed in only 53 of the 744 (7.1%) applications." (Schuren & Mohr, 2008, p.46)

Of course, these findings do not mean that the Laplace equation should be abandoned, but merely that its limitations should be recognised along with the complexities of human anthropometrics. No one should assume graduated compression, but procedures are needed to check this experimentally. Even the goal of graduated compression should be questioned, as do Schuren & Mohr (2010). With sportswear, merely wearing a compression garment is no guide as to what effect it is having on the wearer. The very varied reports of no benefit/some benefit/measurable benefit coming from the sportswear research literature may be simply a pointer to the uncontrolled (and unmeasured) compression that these garments exert. For an example of research that has sought to model and measure compression forces in a more rigorous way, see Dias et al. (2003).

12.5 Summary and Conclusions

1. There are many different measurement systems for monitoring the compression pressures of garments. These use sensors based on a variety of technologies (see Table 12.1). Many of these have had a short lifetime and are of historic interest only.

2. To develop a standard test, many variables have to be excluded. This has been achieved with compression hosiery by eliminating the variability of the human leg and by establishing protocols for loading the garment to the instrument. The Hatra Hose Pressure tester is the test instrument for British Standards 6612:1985, 7672:1993 and 7563:1999.

3. Medical applications need simple test instruments that can be transported easily and where setting up and measurement times are short. Instruments need to be capable of making in vivo measurements. The most widely used test systems are the MST Professional (for compression hosiery), the Kikuhime tester, and the PicoPress®. These have all withstood the test of time and have been developed over the years to incorporate enhancements.

4. Measurement of sportswear compression has made use of medical equipment, but there has been much interest in custom-made systems. There is a recent tendency to use Tekscan technologies. Researchers appreciate the paper-thin sensors, the variety of off-the-shelf sensors available, the sophisticated data-processing software and the visualisation tools. The main problem reported has been drift, and various approaches have been used to obtain reproducible outputs. With a combination of calibration and standardised measurement protocols, acceptable accuracies have been reported.

5. Medical practitioners and researchers appear to have under-estimated the problems of getting a controlled and predictable compression. This is particularly apparent in the

difficulties in producing graduated compression with leg bandages, but it is symptomatic of the variability associated with compression garments. There is an urgent need for sportswear compression research to be accompanied by detailed measurements of compression pressures. Without this, informed assessments of the value of compression garments cannot be made.

6. Further Information

Hatra Mk2A+ hose pressure tester
Segar Technology
email: segar@segartechnology.com
Web: www.segartechnology.com

I-Scan and FlexiForce®
Tekscan, Inc.
email: marketing@tekscan.com
Web: www.tekscan.com/

Kikuhime®
mediGroup Australia Pty Ltd.
email: sales@medigroup.com.au
Web: www.medigroup.com.au
Manufacturer: TT Meditrade, Soledet 15, DK 4180 Soro, Denmark

MST Professional
Swisslastic Ag St. Gallen. (formerly Salzmann AG)
email: info@swisslastic.ch
Web: www.swisslastic.ch/en/

PicoPress®
Microlab Elettronica
e-mail: info@microlabitalia.it
Web: www.microlabitalia.it

7. Acknowledgements

The author wishes to thank the following for supplying images, providing feedback on developments and granting permission to publish: Segar Technology (Hatra Mk2A+ hose pressure tester), TT Meditrade and mediGroup Australia Pty Ltd. (Kikuhime®), Swisslastic Ag St. Gallen. (MST Professional), Microlab Elettronica (PicoPress®), Adrian Smith of the Talley Group, and Sheena Tyler for the image used in Figure 12.7.

8. References

British Standards Institution (BS 6612), 1985. Graduated compression hosiery

British Standards Institution, (BS 7672), 1993. Specification for compression, stiffness and labelling of anti-embolism hosiery.

British Standards Institution (BS 7563), 1999. Specification for non-prescriptive graduated support hosiery.

Brophy-Williams, N., Driller, M.W., Halson, S.L., Fell, J.W. & Shing. C.M. 2014. Evaluating the Kikuhime pressure monitor for use with sports compression clothing, *Sports Engineering*, 17:55–60.

Brorson H, Hansson E, Jense E, Freccero C. 2012. Development of a pressure-measuring device to optimize compression treatment of lymphedema and evaluation of change in garment pressure with simulated wear and tear. *Lymphatic Research and Biology*, June 2012, 10(2), 74-80

Chan, A.P. and Fan, J. 2002. Effect of clothing pressure on the tightness sensation of girdles, *International Journal of Clothing Science and Technology*, 14(2), 100-110.

Dias, T., Yahathugoda, D. Fernando, A. and Mukhopadhyay, S.K. 2003. Modelling the interface pressure applied by knitted structures designed for medical-textile applications, *The Journal of The Textile Institute*, 94(3-4), 77-86.

Fan, J. and Chan, A.P. 2005. Prediction of girdle's pressure on human body from the pressure measurement on a dummy, *International Journal of Clothing Science and Technology*, 17(1), 6–12.

Flaud, P., Bassez, S. and Counord, J.-L. 2010. Comparative in vitro study of three interface pressure sensors used to evaluate medical compression hosiery, *Dermatologic Surgery*, 36(12), 1930–1940.

Giele, H.P., Liddiard, K., Currie, K., Wood, F.M. 1997. Direct measurement of cutaneous pressures generated by pressure garments, *Burns*, 23(2), 137–141.

Lin, Y. Choi, K-f. Luximon, A. Yao L. and Hu, J. and LI. Y. 2011. Finite element modeling of male leg and sportswear: contact pressure and clothing deformation, *Textile Research Journal,* 81(14), 1470-1476.

Lin, Y. Choi, K-f. Zhang, M. Li, Y. Luximon, A. Yao L. and Hu, J. 2012. An optimized design of compression sportswear fabric using numerical simulation and the response surface method, *Textile Research Journal*, 82(2), 108-116.

Liu, R., Lao, T.T., and Wang, S-x. 2013. Technical knitting and ergonomical design of 3D seamless compression hosiery and pressure performances in vivo and in vitro, *Fibers and Polymers*, 14(8), 1391-1399.

Maklewska, E., Nawrocki, A. and Kowalski, K. 2007. New measuring device for estimating the pressure under compression garments. International Journal of Clothing Science and Technology, 19(3/4), 215-221.

Macintyre, L. 2007. Designing pressure garments capable of exerting specific pressures on limbs. *Burns*, 33(5), 579–586.

Macintyre, L. 2011. New calibration method for I-scan sensors to enable the precise measurement of pressures delivered by 'pressure garments', *Burns*, 37(7), 1174–1181

Pain, M.T.G. Tsui, F. and Cove, S., 2008. In vivo determination of the effect of shoulder pads on tackling forces in rugby. *Journal of Sports Sciences*, 26(8), 855–862

Partsch, H., Clark, M., Bassez, S., Benigni, J-P., Becker, F., Blazek, V., Caprini, J., Cornu-Thénard, A., Hafner, J., Flour, M., Jünger, M., Moffatt, C. and Neumann, M. 2006. Measurement of lower leg compression in vivo: recommendations for the performance of measurements of interface pressure and stiffness: consensus statement, *Dermatologic Surgery*, 32(2), 224-32.

Partsch, H. and Mosti, G. 2010. Comparison of three portable instruments to measure compression pressure, *International Angiology*, 29(5), 426-430.

Rotsch, C. Oschatz, H. Schwabe, D. Weiser, M. and Mohring, U. 2011. Medical bandages and stockings with enhanced patient acceptance, In: Bartels, V. (ed), *Handbook of Medical Textiles*, 2011, Cambridge: Woodhead Publishing Ltd., 481-504.

Schuren, J. and Mohr, K. 2008. The efficacy of Laplace's equation in calculating bandage pressure in venous leg ulcers, *Wounds UK*, 2008, 4(2), 38-47.

Schuren, J. and Mohr, K. 2010. Pascal's law and the dynamics of compression therapy – a study of healthy volunteers, *International Angiology*, 29(5), 431-435.

Swain, I. 2005. The measurement of interface pressure, In: Bader, D., Bouten, C., Colin, D. and Oomens, C. (eds). *Pressure Ulcer Research, current and future perspectives*. 2005, Berlin: Springer-Verlag, 51-71.

Teng, T. and Chou, K. 2006. The measurement and analysis of the pressure generated by burn garments, *Journal of Medical and Biological Engineering*, 26(4), 155-159.

Thomas, S. 2003. The use of the Laplace equation in the calculation of sub-bandage pressure, *EWMA journal*, 3(1), 21-23.

Usman, J., McIntosh, A.S. and Fréchède, B. 2011. An investigation of shoulder forces in active shoulder tackles in rugby union football, *Journal of Science and Medicine in Sport*, 14(6) 547–552.

Van den Kerckhove, E., Fieuws, S., Massagé, P., Hierner, R., Boeckx, W., Deleuze, J., Laperre, J., Anthonissen, M. (2007. Reproducibility of repeated measurements with the Kikuhime pressure sensor under pressure garments in burn scar treatment, *Burns*, 33(5), 572–578.