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Pressure Measurement at Biomedical Interfaces

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1. Introduction

The science of pressure measurement is mature with many available pressure measurement technologies some of which have their origins dating back to steam power and the industrial revolution. Along with temperature, the use of pressure as a general physiological parameter is well established and it may be accurately and reliably determined for a wide range of medical applications and environments involving liquids and gases (Mootanah & Bader, 2006). However, pressure is also commonly used in biomedical engineering and medicine to quantify the mechanical interaction between biomedical interfaces such as those arising between tissue and support surfaces (Tissue Viability Society, 2010; Shelton et al, 1998), e.g. beds, seats, prosthesis, and between tissue and pressure applying devices such as tourniquets (Doyle & Taillac, 2008), bandages (Partsch et al, 2006), and surgical instruments. The term interface pressure may be used in such contexts. The materials constituting such interfaces are, in general, not fluidic but connective and so they may support shear and torsional forces in addition to the normal hydrostatic force quantified as pressure/interface-pressure (average normal force per unit area).

The non-fluid nature of many biomedical interfaces means that current mature quantitative pressure measurement technologies are not well matched to the biomedical interface environment, i.e. they are not media compatible. Pressure measurement devices may also contribute to erroneous and anomalous data since they need to be deployed at the interface site and so are necessarily intrusive (Casey et al, 2001; O'Brien & Casey, 2002). These problems have acted as a barrier to the more general use of pressure measurement to characterise mechanical interactions at biomedical interfaces. In cases where such measurements have been used, the resulting data is often degraded due to interface and sensor contact artifacts ultimately leading to difficulties with reproducibility and reliability (Buis & Convery, 1997; Fay & Brienza, 2000).

Significant benefits would arise if a reliable low cost interface pressure measurement technology were available for emergency, acute and home medical care environments as well as areas such as Intravenous Regional Anaesthesis (Casey et al, 2004), compression therapy and pre-hospital emergency care. Such a technology would allow improved diagnostics and treatments, early hazard warning and could save lives (Kragh et al, 2009; Noordin et al, 2009).

In this chapter, modifications to readily available MEMS devices are described which render the devices suitable for general purpose non-invasive pressure and pressure gradient measurement at biomedical interfaces. Sample data is presented for a range of biomedical application environments including pneumatic and non-pneumatic tourniquets, bandages

and support stockings. It is hoped that this will stimulate more extensive on-body testing and verification of the technology resulting in pressure measurement solutions which will contribute to a better understanding of the biomedical interface environment ultimately leading to improved efficacy of treatments and procedures and simple standardised measurement protocols.

2. Background

The forces that occur in many biomedical contexts are distributed over areas rather than being discrete, i.e. applied at a point with a well defined (arbitrary) direction. Consequently, pressure (defined as the average force per unit area acting normal to the surface) is the parameter of choice in quantifying many biomechanical interactions. In the case of static fluids acting on surfaces, pressure reliably quantifies the forces involved since we may neglect shear forces and the only forces acting will be perpendicular to the surface (i.e., hydrostatic). Common pressure measurement sensors and transducers will provide high integrity data for such applications, e.g. arterial blood pressure, urethral pressure and intra compartmental pressure.

In this work we are concerned with interfaces between two continuum phases one of which is living tissue and the other a biomedical device or support surface. Specifically we are interested in the non-invasive measurement of pressures at such biomedical interfaces. These arise, for instance, in compression therapy using bandages (Ferguson-Pell et al, 2000; Ghosh et al, 2008), which is the mainstay of venous leg ulcer treatment (Grace, 2003). In these situations specific pressures are required to achieve the desired clinical outcome. If the pressure is too low no clinical benefit will accrue while if the pressure is too high an adverse outcome may result. Pneumatic tourniquets are routinely used in intravenous regional anaesthesia to occlude arteries and to control anaesthetic and provide a bloodless operating field (McEwen, 1994). Tourniquets are also deployed in combat (Kragh et al, 2009; Noordin et al, 2009; Tien et al, 2008), in civilian emergency settings (Lee et al, 2007) and for remote emergency care (Fludger & Bell, 2009). Patient tissue is also compressed by support surfaces, e.g. beds, wheelchairs and prosthetics (Ferguson-Pell et al, 1993). With most of these applications one may expect shear (Wertheim et al, 1998) and torsional forces to coexist with normal forces, giving rise to resultant compound forces. Standard pressure measurement solutions that assume fluidic transmission of pressure will not yield reliable quantitative data in such applications.

Ideally, the physician would like to know the actual forces acting (e.g. normal and shear forces at a limb/organ surface; radial force near an arterial wall; tensile force on nerve tissue) at specific locations in order to inform decisions in relation to patient well-being and diagnosis. However, such detailed information is not routinely available (Shear Force Initiative, 2010) unless through expensive MRI (Oomens et al, 2010) and other imaging techniques. Standard pressure sensors and transducers are often used at biomedical interfaces because of their ready availability. However, the conditions under which verifiable data may be obtained for such use are rather restrictive. For instance, in the case where tissue of Poisson's ratio 0.5 is being perfectly hydrostatically compressed, i.e. the entire volume of the tissue is subject to a uniform normal force per unit area, then, the pressure within the tissue or at interfaces, will correlate exactly with the hydrostatically applied pressure. However, this is rarely the case in practice. Poisson's ratio for real tissue may vary over a range from about 0.2 to 0.4 (Cristalli et al, 1993). With departures of Poisson's ratio from 0.5 (incompressible), deformation of the tissue will occur and with connective tissue, shear forces, as well as compressive forces, will arise

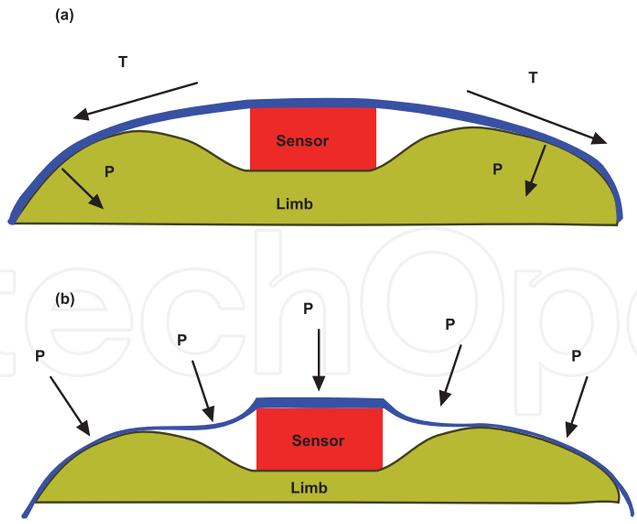

Fig. 1. Deformation of interface region due to intrusive rigid sensor. The 'Hammocking' effect for (a) Tension devices such as bandages and combat application tourniquets (CATs), (b) Pneumatic tourniquet type devices.

at the surface of the tissue and within its bulk. Additionally the pressure applicator rarely encompasses the entire body but is more usually applied to a localised section or sections of tissue. Time variant deformation of the compressed tissue is likely in such instances due to migration of tissue fluids from the compressed region. Inclusions such as bones and other hard tissue further complicate the situation. This makes it very difficult to compare values of interface pressure obtained by different research groups and has lead some groups to seek an alternative metric (e.g. statistical (Shelton et al, 1998), Limb Occlusion Pressure, LOP (Aziz, 2009; McEwen et al, 2002), deformation (Oomens et al, 2010), etc.) to reliably quantify the effects of compression on real tissue. Others question whether pressure or interface pressure should be used at all (Fay & Brienza, 2000; Oomens et al, 2010).

Further complications (Allen et al, 1993; Buis & Convery, 1997) arise with the measurement of pressure at biomedical interfaces using pressure sensors deployed at the interface since such sensors are in the main intrusive and may introduce measurement artifacts and errors. For instance, rigid sensors will prise the interface materials away from each other, Fig. 1 and give rise to the so called hammocking effect (O'Brien & Casey, 2002) which is pressure dependent and leads to anomalously high readings and difficulties with calibration. Low profile(van Hout et al, 2003), minimally intrusive, conformal sensors (Polliack et al, 2000) are available which allow localized (Picopress, 2009; van den Kerckhovea et al, 2010) and full body interface pressure mapping (distributed pressure) (Lai & Li-Tsang, 2009; Pliance, 2011). The latter systems can involve sophisticated signal processing and special calibration procedures tailored to the application environment. For such applications the most important technical performance parameters are repeatability, resolution, drift, creep and accuracy in that order (Skelton & Lott, 2007). Low cost, universal biomedical pressure and pressure gradient measurement systems with high spatial resolution which provide continuous, accurate, dynamic data flow for important biomedical applications are currently not readily available.

Our objective is to develop an inexpensive biomedical interface pressure transducer (BIPT) with minimum cross sensitivity to shear forces. We believe that such a sensor would provide

reproducible and reliable data for a wide range of application environments without overly complicated measurement procedures and protocols. This would increase the confidence of practitioners in the use of biomedical interface pressure data. The realisation of such a technology would represent a significant milestone in a much larger project seeking to develop an integrated sensor solution with capability to discriminate between, and quantify, both shear and pressure at biomedical interfaces.

3. Functional specification for an optimal biomedical interface pressure transducer

An optimal pressure transducer (Paris-Seeley et al, 1995) for non invasive pressure measurement at biomedical interfaces should combine the following features (Partsch et al, 2006):

- be capable of measuring the pressure applied by any one of a specified number of medical devices to a portion of a human body surface or tissue, in the pressure range 0-750 mmHg (0-100 kPa);
- permit fast, convenient and intuitive calibration checking in the target application environment;
- typical measurement errors should be less than 1% of full scale;
- hysteresis/creep over a one-hour time period should be less than $\pm$ 1% of full scale;
- allow easy compensation for ambient pressure and temperature variations;
- be digitally compatible, i.e. easily interfaced to modern microcontrollers;
- have similar compliance to that of the target tissue;
- conform to curved, compliant tissue surfaces with radii of curvature as low as 2 cm (pediatric cuffs);
- be low profile (<2 mm high) with very small footprint (<1 cm^2)
- must not significantly alter the tissue device interface;
- be made from biocompatible materials;

In addition to the above the transducer should be immune to electromagnetic interference, conform to relevant electrical standards, must not present electrosurgical or thermal hazards, must fail safe and, for reusable devices, must be sterilizable using at least one of the various conventional sterilization techniques. The performance to cost ratio must be high to allow deployment in disposable devices.

4. MEMS as a candidate technology for pressure measurement at biomedical interfaces

The specification for an optimal BIPT is very demanding. However, the combination of electrically excited solid state sensors/transducers with modern microcontroller/digital processors provides an excellent platform for the development of high performance, versatile, application specific measurement system solutions (Barker, 2000). Miniature general purpose pressure sensor devices are typically piezoresistive, i.e. transform a mechanical stress into a resistance signal. Of the various electrical device properties, electrical resistance is the easiest one to measure precisely over a wide range at moderate cost. The piezoresistive

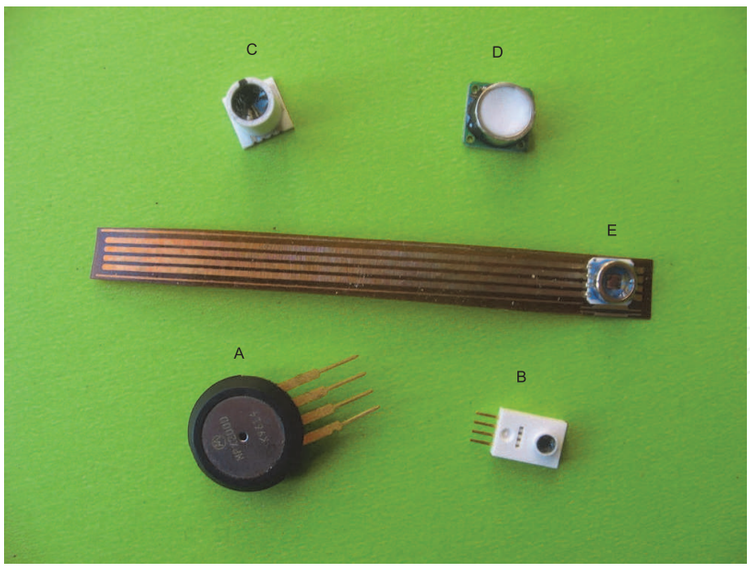

Fig. 2. Some MEMS pressure sensors: A, MPX2000 and B, MPX2300DTI, Freescale Semiconductor Inc.; C, MS Series, Merit Sensor Systems; D, HRPF0100, Hope RF (Rhopoint Components); E, MS5201, Measurement Specialties (Intersema) as mounted on flex-circuit.

silicon pressure transducer is a solid state, microelectromechanical system (MEMS) (Wise, 2007) sensor fabricated using silicon integrated circuit processing technology. Therefore, it benefits from the economies of scale and enhanced performance associated with silicon technology yielding devices with high performance to cost ratios. Devices may be voltage or current excited. They may be simple transducers, i.e. convert pressure/pressure changes to resistance or may be complete pressure transmitters converting the pressure signal to a standard electrical signal (analogue or digital) via integrated processing electronics. Consequently, MEMS technology deserves serious consideration as a candidate technology for the development of transducers for the measurement of pressure at biomedical interfaces. High performance, low cost MEMS pressure sensors, Fig. 2 are used extensively in fluidic pressure measurement for medical and biomedical applications (Wise, 2007). Their use in consumer goods such as wrist-watch altimeters and depth gauges is providing a technology push to lower profile, compact, devices (Epcos, 2009). Similarly, biomedical applications are providing an incentive to develop so called 'media compatible' devices (Lucas Novasensor, 2010). Recently, progress has been reported with the development of prototype MEMS devices having tissue compatibility, i.e. capable of measuring tissue contact pressures directly without gas/liquid lines/interfaces (Casey et al, 2010).

The delicate silicon micromachined membrane in MEMS pressure sensors must be mounted on a rigid chip carrier and it, plus any integrated circuitry and bond wires, must be protected from the ambient and target measurement environments to avoid drift in specifications and/or failure. Consequently, MEMS pressure sensors necessarily involve rigid packaging which may be many times the volume of the enclosed microchip, in order to confer the desired mechanical and chemical immunity. Typical package heights, without coupling nozzles, range from 3 to 5 mm for off-the-shelf devices. This, combined with package footprints of $1cm^2$ or more, and cumbersome electrical interconnects (Fig. 2) has made this attractive technology overly intrusive for general medical interface pressure measurement applications. Furthermore, these devices are optimized for measurement of pressure in fluid environments

where shear coupled forces either do not exist or are negligible compared to the normal hydrostatic forces which are easily coupled to the sensing diaphragm either directly or via a soft gel barrier layer. Therefore, the many benefits of MEMS pressure sensing technology could be exploited in pressure measurement at biomedical interfaces if the devices could be rendered media/tissue compatible while simultaneously making them minimally intrusive so as to reduce sensor related artifacts. Modifications to off-the-shelf MEMS pressure sensors are therefore necessary in order to address these coupled challenges.

5. MEMS sensor selection

Pressure sensors including MEMS pressure sensors, are generally categorised according to the reference pressure used with them. Three categories are common: absolute pressure sensors measure pressure relative to vacuum; differential pressure sensors measure pressure relative to a reference pressure line; gauge pressure sensors measure pressure relative to ambient atmospheric pressure. Differential devices tend to be formed with a fluid line connector to the reference port thereby adding to the overall size/profile of the device. Gauge devices, on the other hand, need only an in-plane hole to port to the atmosphere and so can have minimal overall size and profile. Absolute devices have a sealed reference chamber typically enclosing a high vacuum. MEMS biomedical pressure transducers normally operate at ambient atmospheric pressure and so either gauge or absolute devices may be used.

The sensor pressure range is dictated largely by the end application and the sensitivity required. Bandage and support surface environments, for instance, require relatively low range sensors, e.g. 0-100 mmHg or lower, while IVRA and general tourniquet measurements require a dynamic range stretching to in excess of 500 mmHg while prosthetic interface measurement devices require higher ranges again. Another important range related sensor parameter is the overpressure rating. Because of the necessary direct tissue-MEMS contact, devices must be able to withstand and recover from burst pressures and directly applied forces that are multiples of the maximum rated pressure. Overpressure ratings of two to three times the maximum rated pressure should be considered in order to ensure a reasonable engineering safety margin.

In addition to the normal electrical performance characteristics of high accuracy, high sensitivity, low drift, linearity, fast response time and low hysteresis desirable in all sensors, a full bridge configuration, energized by a unipolar supply is an additional desirable characteristic for biomedical pressure logging applications and digital product development.

Small size is critically important in order to reduce the intrusiveness of the device and thereby minimize artifacts due to hammocking (Casey et al, 2001) and shear and to increase spatial resolution. The device must also be mechanically robust and so a packaging technology which provides good mechanical and chemical protection combined with low profile package interconnects is essential. Surface mount (SM) packaging is a low cost assembly and interconnect technology which meets these needs and at the same time offers a route to low cost automated product assembly.

The size of the sensor port exposed to the target media is also critically important in selecting a MEMS sensor for the measurement of pressure at biomedical interfaces. This port must be filled with a tissue compatible layer (interface layer) which couples the media/tissue pressure to the MEMS sensing element. The interface layer must be soft but durable and must not adhere to the target media. If the port opening is narrow, surface tension forces between the barrier medium/gel and the port walls can detrimentally affect sensor performance, particularly response time. It can also introduce hysteresis and significantly reduce device

Manufacturer	**Merit**	**Measurement Specialties Inc.**			**Hope RF**
Type	MS Series	MS5401	MS1451	MS5201	HRP0100
Range (mmHg)	0-774	0-750	0-258	0-750	0-750
(kPa)	0-103	0-100	0-34	0-100	0-100
Type	Diff.	Abs.	Gauge	Gauge	Gauge
Package	Cer.	Cer.+Met.	Cer.+Plas.	Cer.+Met.	HR4+Met.
Length (mm)	6.4	6.4	7.6	7.6	9.2
Width (mm)	6.4	6.2	7.6	7.6	9.2
Height (mm)	2.94	2.8	3.68 (3.2mm)	2.8	4.5
Port Dia. ϕ (mm)	4.4	4.4	4.0	6.0	7.4
AAR	0.31	0.38	0.22	0.34	0.51
Contacts	8 DIL	8 DIL	8 DIL	8 DIL	4 Pads

Table 1. Sample sensors used in modified MEMS Biomedical Interface Pressure Transducers (AAR - Active Area ratio).

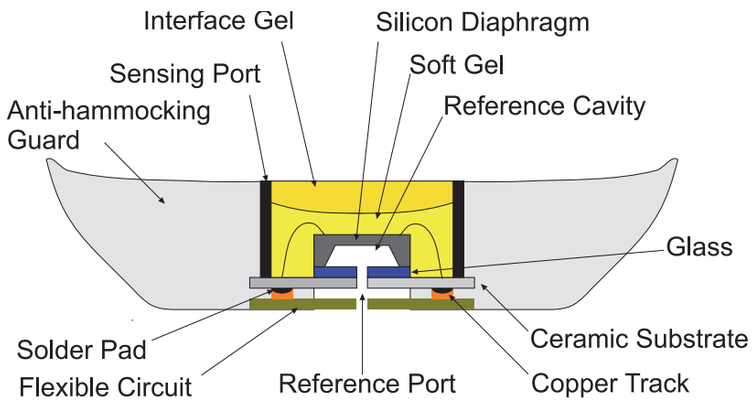

Fig. 3. Schematic representation of a gel modified MEMS pressure sensor with conformable anti-hammocking guard.

sensitivity (see Sec. 7 below). A useful index in selecting a suitable MEMS device for interface pressure measurement is the Active Area Ratio (AAR), i.e. device port size to footprint ratio. Ideally this should be as close to unity as possible but values of 0.3-0.5 are common with current technologies, see Table 1. Of course the overall footprint size will determine the spatial resolution in array configurations designed for gradient measurement and so should be as small as possible, e.g. less than one square centimetre. The package should preferably be made from biomedically compatible materials. Fortunately, many MEMS pressure sensor vendors provide SM package options designed for medical/biomedical applications. Many devices have on-chip processors which provide digital output. While this is desirable in many instances, it can add to package size, device cost, and impose constraints on follow on circuitry in the signal path.

The data in Table 1 summarises key specifications for a sample range of commercially available MEMS pressure sensors meeting the above design criteria or which may be easily modified to meet the criteria. The table is not a comprehensive survey of all vendors and their matching products but rather is a sample of devices that were tested by this group.

6. MEMS sensor modifications

Many MEMS devices come with barbed or plain molded plastic front port connection tubes (Meas. Specialties Inc., MS5201) but otherwise match the selection criterion outlined for biomedical interface sensors. For such devices the package tube and chamber top cap were removed using a cutting jig specially made for the purpose. The jig holds the sensor firmly in place while providing a reference surface along which a sharp blade is moved in order to slice off the cap and tube in a single clean movement, i.e. without sprinkling debris onto the isolation gel and sensitive chip/diaphragm.

Sensors were reflow soldered onto flexible circuit, see Fig. 3, to facilitate electrical interconnection while deployed on body. Polimide flexible circuit with copper cladding was patterned with pad landing areas matched to the sensor surface mount device to be used. Copper tracks extended from the bridge excitation and bridge output pads along a flying lead section where the track pitch was matched to that of standard zero insertion force (ZIF) connectors, i.e. pitch of 1.27 mm. Solder paste was spread onto the pads using a stencil cut from polimide tape. The sensor device was placed onto the substrate and manually aligned with the solder coated pads. The combination was placed in a solder reflow oven with a plateau temperature of 185°C.

Once the sensor was mounted on the flexible circuit, and in cases where a protective gel die coating was not already present, a soft ion free silicone gel (Dow Corning Sylgard 527) with viscosity less than 1000 cps and low hardness (Intersema, 2004), was added, after vacuum degassing, to protect the sensor die and bond wires and to provide a humidity barrier while leaving head room for the interface layer. The protective gel was allowed set for at least two hours before adding interface gel. The interface gel fills the cavity completely and provides a durable non-stick surface which may be brought directly into contact with body tissue or pressure applying materials/parts. Room temperature vulcanising silicon rubber with hardness in the range, Shore A, 23-30, was found to work reasonable well for this purpose. The two gel combination resulted in robust, durable, zero volume displacement devices.

A polydimethylsiloxane (PDMS) anti-hammock guard was moulded around the sensor assembly in order to provide a constant well defined sensing area/volume at the target interface. The rear end of the guard was flush with the rear surface of the flexible circuit and the guard front surface was flush with the top surface of the port interface gel surface. The objective of the symmetrical design is to ensure that all top and bottom guard-contact-forces, apart from those coming to bear on the sensing area, are mutually canceled. The overall geometry of the guard was diamond shaped with contoured side walls and a slight lip on the front surface perimeter which gives an air-tight cupping effect against tissue. The PDMS/flexible circuit in the region of the rear port of the MEMS sensor (gauge devices) was drilled to provide an opening to the rear port of the sensor thereby allowing equilibration of the device with ambient atmospheric air pressure. A composite teflon/PDMS anti-hammock guard was also tested but did not offer any benefits over a pdms guard on its own. A photograph of various biomedical interface pressure transducer (BIPT) configurations is shown in Fig. 4.

7. Results

Preliminary device testing and characterisation was carried out using an isolated nominal 6V supply. A teflon nozzle was machined to fit over the MEMS sensor front port to allow device testing and calibration under air. The other end of the nozzle was connected to an

Fig. 4. Discrete and array Biomedical Interface Pressure Transducers (BIPT) incorporating modified MEMS pressure sensors.

inflation bulb via a mercury manometer. A soft silicone washer provided an airtight seal between the nozzle and the sensor port. This allowed easy, reliable testing of the transducers in the 0-300 mmHg pressure range. The stiffness of the interface gel is critical to device performance. PDMS (Shore A, 50) was found to be too stiff introducing large zero offsets and drift plus hysteresis. Q-gel which is a very soft silicone gel strongly adhered to tissue and other media and divided when separated from the support tissue. Room temperature vulcanising silicones on the lower end of the hardness scale (Shore A 23-30) such as that supplied by Rubson for general domestic sealing applications were found to be very durable and non-stick while still being soft enough to allow in-specification MEMS performance. Mould makers' silicone rubber (T20 and T30, Alec Tiranti Ltd, www.tiranti.co.uk) was also used successfully as MEMS media interface gel as was room temperature vulcanising silastic rubber from Dow Corning. The hardness of both the mold makers' rubber and the silastic rubber could be lowered by dilution with silicone fluid. In addition, it was possible to vacuum degas these rubbers to reduce air bubble entrapment which can lead to dimensional instability, over time, in the cured rubber. Device specifications (zero offset, sensitivity and response time), after gel modifications, remained within the manufacturer's specifications for all devices tested, see calibration curve, Fig. 5.

7.1 Pressure measurement under a surgical pneumatic tourniquet

On body data logging was carried out using a National Instruments USB data acquisition card (USB-6008) to interface the transducers to a laptop computer running NI Labview 8.5. This device can accommodate 4 differential analogue input channels and also has an analogue output voltage channel which may be programmatically controlled within the range 0-5V. This was used to provide a reference voltage (3V) to a unity gain operational amplifier powered by the USB-6008 5V supply. The output of the buffer amplifier was used to

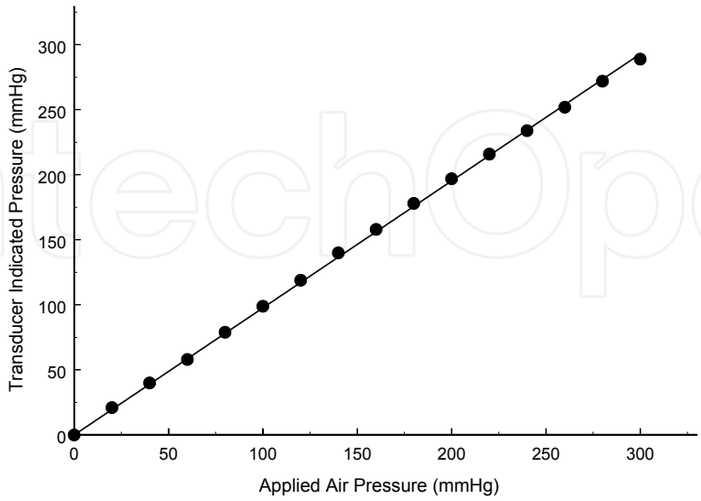

Fig. 5. BIPT calibration curve using air applied pressure (inflate-deflate cycle).

energise the MEMS sensor(s). An interconnect card with 4 Berg, 4-way sockets allowed easy make-break connections between the transducers and the USB-6008. A Labview program (Virtual Instrument) was developed to facilitate transducer calibration, data display and data storage. This configuration allowed data logging of up to four transducers simultaneously. One channel was generally reserved as a reference channel to monitor tourniquet inflation pressure or ambient atmospheric pressure.

The following protocol was adopted for on-body measurements using a 100 mm wide Zimmer pneumatic tourniquet cuff. Subjects were seated with their arm (normally left arm) resting on the bench and extended. Measurements were made at two locations on each subject's arm: the inner lower arm and the inner upper arm (biceps). The BIPT was zeroed and calibrated at 150 mmHg set-point (air applied pressure using the same bulb/manometer as used for cuff inflation). The BIPT was placed directly onto the tissue with the sensing volume facing the tissue and the interconnect lead taped in place. One round of cotton wool was loosely wrapped around the arm and sensor to cushion the tourniquet. The tourniquet was centered over the sensor and fitted so that it gave a snug fit to the arm. The measurement sequence comprised initial zeroing of the transducer on the bench; application of the sensor and tourniquet; inflate/deflate cycle (0, 20, 50, 100, 150. . . .300, 250,. . . 100, 50, 20, 0); zeroing of the BIPT in situ under cuff if required; repeat inflate/deflate cycle.

The results for two complete log sequences are shown in Fig. 6. The BIPT indicated pressures track the tourniquet applied pressure but at slightly elevated values particularly at the higher inflation pressures. A slight negative BIPT pressure is indicated after completion of the first inflate-deflate cycle. This is largely due to the temperature sensitivity of the device's zero pressure offset voltage which drops as the sensor temperature rises to body temperature. Re-zeroing the transducer at zero tourniquet pressure while still under the cuff (see second 'Lower Arm' cycle) after body-temperature equilibration increases the degree of elevation in indicated pressures, i.e. + 25 mmHg at 300 mmHg tourniquet inflation pressure. The

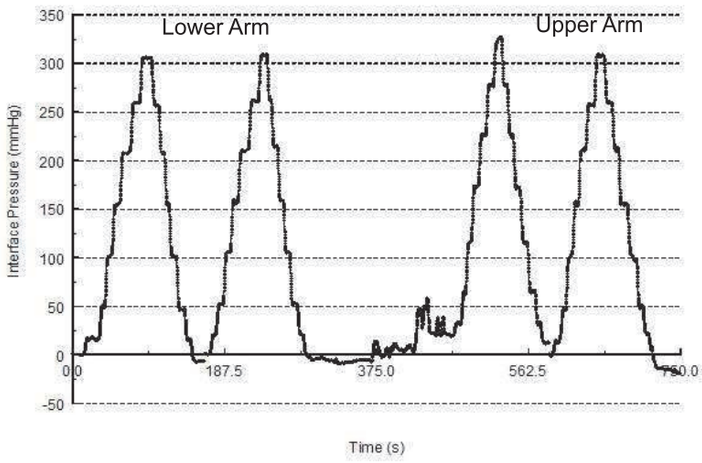

Fig. 6. Interface pressure measured under 10 cm Zimmer Cuff on Lower and Upper Arm

sharp pressure transitions after 375 s in Fig. 6 occur as the tourniquet cuff is tightly re-fitted onto the arm over the transducer prior to inflation. The BIPT therefore allows one judge the tightness of fit of the cuff (not detectable using the cuff inflation pressure gauge). This initial cuff applied pressure provides a further elevation in the indicated pressures which may be compensated/nulled by re-zeroing the transducer once it has settled onto the arm, i.e. after a complete inflate-deflate cycle. The elevated BIPT indicated pressures are probably due to residual sensitivity of the transducer to shear forces and tissue contact artifacts.

Data-log sequences were carried out on 11 healthy volunteers in order to gauge the spread in BIPT indicated pressures for a sample of limb sizes and tissues. Subjects comprised 4 females (age range 22-52) and 7 males (age range 18-54). The mean and standard deviation of the sub-tourniquet pressure readings obtained for all 11 subjects at the inflate/deflate set-points (second cycle) for both lower and upper arm locations is presented in Table 2. The error is the ratio of the standard deviation to set-point value at each set-point expressed as a percentage, i.e. relative set-point error. The relative errors are highest at low pressures where offset is most pronounced. In the critical range for intravenous regional anaesthesia (100-300 mmHg) errors can be as large as 8%. The composite data for all 11 subjects plotted in Fig. 7, also displays a positive skew in the error particularly at higher pressures, as noted earlier with the single inflate-deflate plots and attributed to residual shear sensitivity.

In tests where large positive offset errors were indicated, (upper arm Fig. 8) flexing of the arm muscle, i.e. tensioning and relaxing it a number of times, while the cuff was inflated to pressures greater than 200 mmHg reduced the offset error. This supports the view that errors in BIPT indicated pressures, where they arise, are largely due to the coupling of shear forces to the BIPT sensing area.

7.2 Emergency and Military Tourniquet (EMT) and Combat Application Tourniquet (CAT®)

Body extremities - arms and legs - bear the brunt of traumatic injury in both civilian and military settings. Quickly addressing life-threatening hemorrhage from an extremity with the

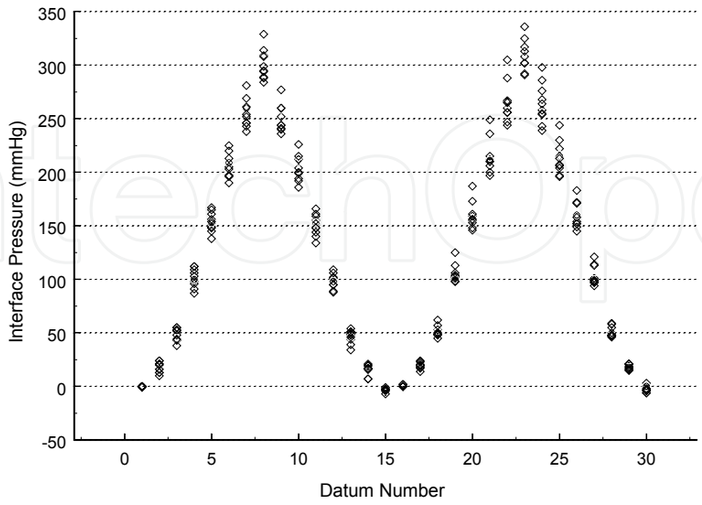

Fig. 7. Measured interface pressure data from the upper and lower arms (second inflate/deflate cycle) of 11 subjects. Inflate-deflate sequence, 0, 20, 50, 100 ..300, 250, ..50, 20,0 mmHg.

	Lower Arm			Upper Arm		
P (mmHg)	Mean (mmHg)	SD	Error	Mean (mmHg)	SD	Error
0	0	0		1	1	
20	18	4	22%	20	3	14%
50	48	5	11%	51	4	9%
100	101	8	8%	104	8	8%
150	153	8	5%	159	11	7%
200	206	10	5%	214	14	7%
250	255	12	5%	265	17	7%
300	301	13	4%	314	18	6%
250	252	12	5%	264	17	7%
200	205	11	6%	214	14	7%
150	151	9	6%	160	11	7%
100	98	6	6%	104	8	8%
50	47	6	11%	51	4	9%
20	15	4	22%	19	3	15%
0	-4	1		-2	3	

Table 2. Mean and standard deviation of readings for all 11 subjects at the inflate/deflate set points for both lower and upper arm locations.

use of a relatively simple maneuver such as applying a tourniquet can reduce morbidity since limb-injury exsanguination is a leading cause of preventable trauma deaths on the modern

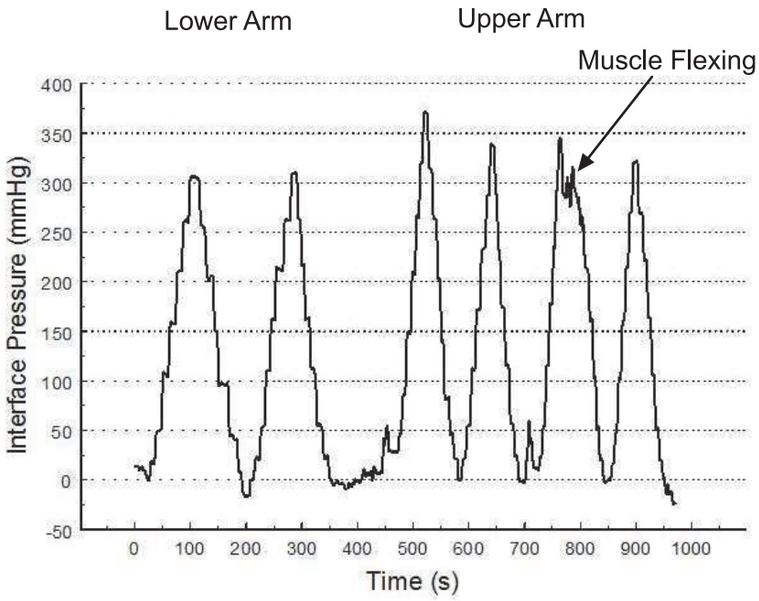

Fig. 8. Interface pressure on arm of subject with large shear artifacts relieved by muscle flexing.

battlefield (Rush et al, 2009). The primary function of the tourniquet is to occlude major arteries in order to save life (Walters et al, 2005). However, in cases where the limb may be salvageable, it is important that the tourniquet pressure should not be excessive and should be evenly and uniformly applied around the limb (Glinz & Jameson, 2010). In these and many other biomedical settings, interface pressure gradients are as important as peak or local pressure values (Oomens et al, 2010). It is a relatively simple matter to arrange the BIPT described here into multisensor arrays, see Fig. 4. The planar three sensor array configuration is useful for determining the local pressure and pressure gradients under cuffs, bandages and other extensive biomedical pressure applying devices. Clearly the spacing may be varied to suit the particular pressure applicator, and more sensors may be added as desired since the flexible carrier circuit is easily customised. Gradient data for a pneumatic emergency and military tourniquet (EMT, Delfi Medical Innovations Inc.) (Lee et al, 2007), is presented in Fig. 9. In this case the gradient transducer comprised three gel modified Measurement Specialties Inc., MS5201-AD MEMS sensors mounted linearly onto a single flexible circuit substrate on 1 cm centres. A PDMS anti-hammock guard was moulded around the combination with similar profile to that described earlier for individual devices. First inflate-deflate cycle data indicate anomalously high readings for set-points in excess of 100 mmHg. However, second cycle data shows much better correlation between set-point data and sub-EMT pressure measurements. This is attributed to cuff settling and consequent reduction in shear forces acting on the overall transducer structure. The second cycle pressure data indicates a drop in pressure of about 20 mmHg/cm either side of the centre of the cuff at pressures greater than 200 mmHg, i.e. a relatively low pressure gradient.

Corresponding data obtained for different transducer positions under a combat application tourniquet (CAT, Composite Resources, USA) applied to the upper leg is shown in Fig.

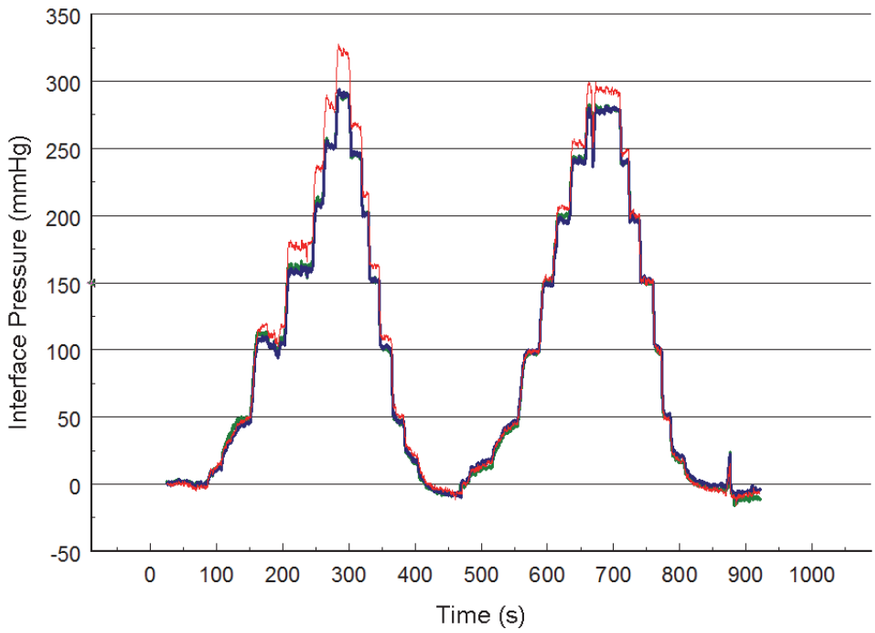

Fig. 9. Concurrent interface pressure from a BIPT three sensor array placed transversely under an EMT (Delfi Medical Innovations Inc.) on the upper leg for two inflate deflate cycles. Sensor positions: central, black; distal, red; proximal, blue.

10. The transducer was placed transversely under the CAT and tightened by turning the windlass handle through four full revolutions before reversing the procedure to release the applied pressure. The first cycle shows data for the transducer placed centrally under the CAT. The subsequent two cycles correspond to transducer positions progressively closer to the CAT distal edge. The last two cycles correspond to CAT positions off-centre and progressively closer to the proximal edge of the CAT. The CAT position was not changed during these measurements. The first spike plus plateau with pressures in the range 20-40 mmHg is due to tightening of the CAT velcro strap. The subsequent spikes followed by plateaus in pressure correspond to full turns of the windlass handle. The spikes are largely due to the requirement to twist the windlass rod/handle past the securing structure before allowing it to untwist slightly back into the securing hook (Casey & Little, 2010). The pressure gradients indicated for the CAT (250-300 mmHg/cm for peak applied pressures of 300-400 mmHg) are considerably greater than those for the EMT consistent with previously published results (Noordin et al, 2009) with particularly high gradients close to the CAT edges. Data for the gradient transducer placed longitudinally under the CAT is presented in Fig.11 indicating relatively small pressure variations circumferentially under the CAT. While shear forces are also likely to contribute to indicated BIPT pressure values for the CAT, a fully independent pressure measurement is needed in order to access the degree of shear contribution. One possibility is to incorporate a dynamometer into the tensioning strap of the CAT (Casey & Little, 2010) from which pressure might be inferred using the LaPlace rule and the limb dimensions.

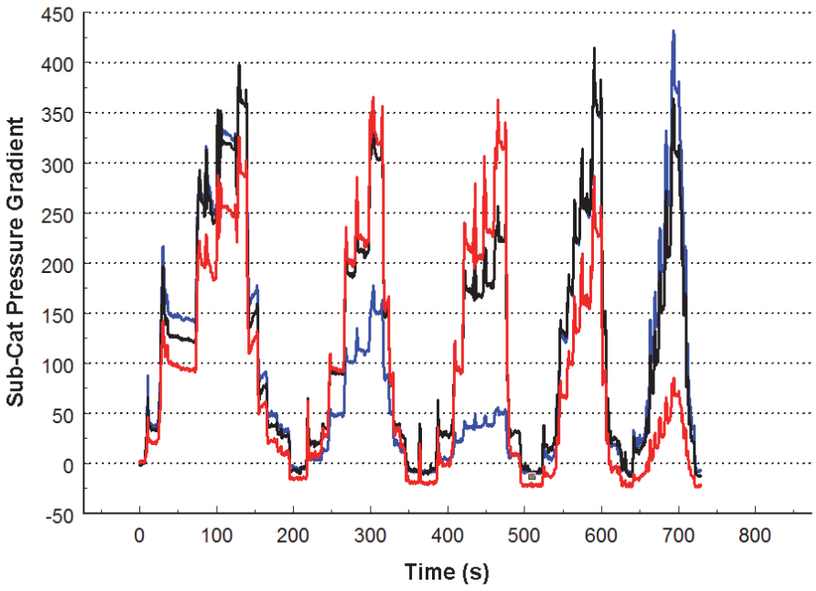

Fig. 10. Concurrent interface pressure for a Combat Application Tourniquet (CAT) on the upper leg with BIPT three sensor array placed centrally (A) and at two positions distal to (B) and proximal to (C) the central position. (Array sensor positions: central, black; distal, red; proximal, blue)

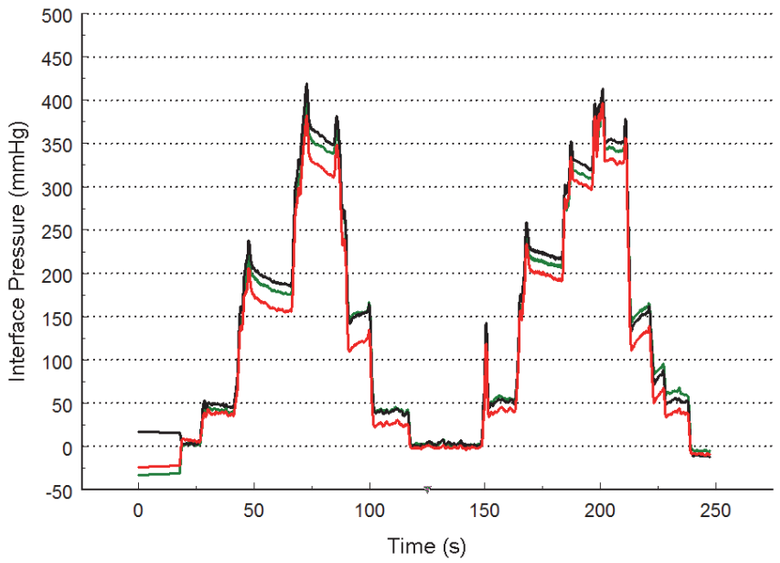

Fig. 11. Concurrent interface pressure for a CAT on the upper leg with the BIPT three sensor array placed longitudinally under the CAT. First cycle - outer leg position; second cycle - inner leg position.

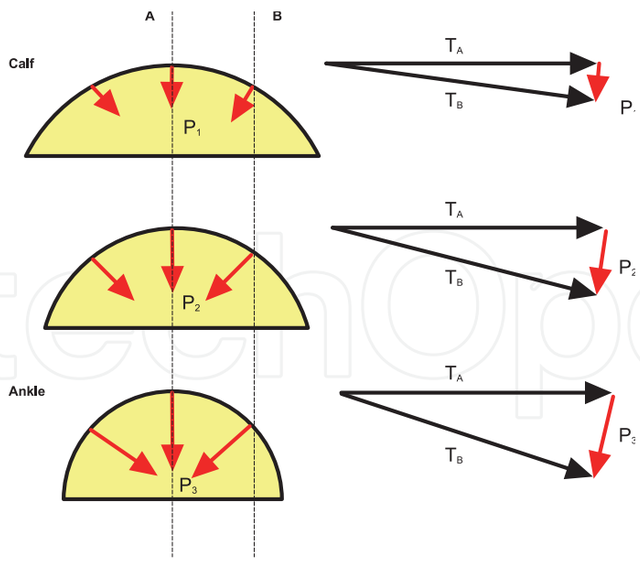

Fig. 12. According to the law of LaPlace, a constant tension membrane produces a pressure which is proportional to the local curvature.

7.3 Bandages and support stockings

In compression therapy, bandages or support stockings are used to aid venous return and reduce venous hypertension which results from chronic venous insufficiency. The pressures involved are much lower than those encountered in tourniquet applications (artery occlusion) and are typically in the 20-50 mmHg range. The tension in the bandage or stocking membrane produces a compression or normal pressure on the supporting limb according to the so called 'Law of LaPlace'. The pressure is proportional to the membrane tension, the number of turns (membrane layers) and the curvature of the limb, $P \propto TN\kappa$ where the curvature $\kappa = 1/R$ for a cylindrical geometry of radius R. For a bandage applied with constant tension and extending from foot/ankle up to and including the calf, regions of high curvature such as the region around the ankle, see Fig. 12, will be subject to high interface pressure compared to a region of low curvature such as the calf. A properly applied bandage should, therefore, produce a pressure profile which decreases from ankle to calf. Such a pressure profile is believed to aid venous return of blood from the ankle region and produce a favourable effect on subcutaneous interstitial pressures (Giswold & Moneta, 2005)

Bandages are designed to generate a pressure in the ankle region of about 40 mmHg dropping to about 20 mmHg just distal to the knee. A sub-bandage interface pressure transducer based on micromoulded (soft-lithography) elastomer springs and flexible circuit technology has been used to measure sub-bandage pressures (Casey et al, 2010). The modified MEMS devices described here were also tested under bandages and support stockings. The interface pressure transducer was placed 5 cm above the medial malleolus facing the limb and held in place using adhesive tape. The bandages were applied to the leg by a trained practitioner and the interface pressure data was logged. The results obtained for a Smith & Nephew *ProGuide* bandage are shown in Fig. 13. The bandage generates a steady pressure of around 40 mmHg in the subject while seated. The fast dynamic response of the sensor also shows the effect of muscle flexing, i.e. the muscle pump action. Previous studies have shown that muscle contraction in the presence of a compression bandage or stocking results in a significant increase in venous blood flow (Lyons et al, 2002). Standing and elevation produce

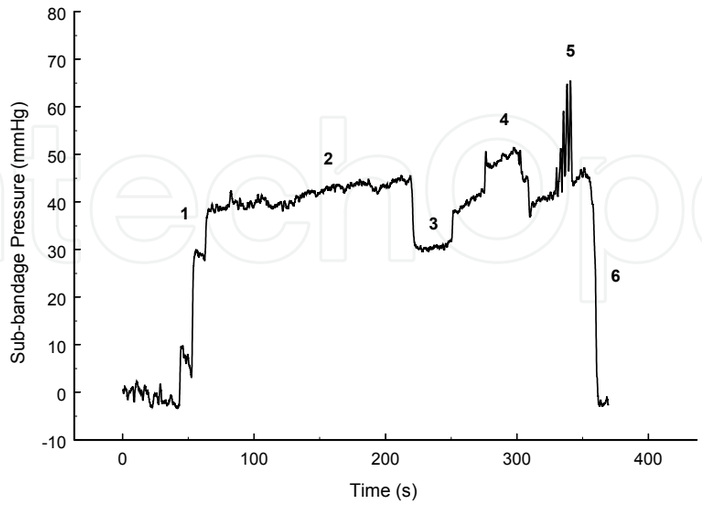

Fig. 13. Interface pressure measured with BIPT located above the medial malleolus under a Smith & Nephew ProGuide bandage: 1 Bandage Application, 2 Sitting with foot on floor, 3 Foot elevated, 4 Standing, 5 Flexing calf muscle, 6 Bandage removal.

pressure fluctuations above and below the value for a seated subject respectively, as expected. Similar results were obtained using Smith & Nephew, *Profore* bandages. The interface pressure measured above the medial malleolus under a Medivan CCL1 support bandage is shown in Fig. 14. According to the manufacturers, the CCL1 should generate pressures in the region of 30-40 mmHg at the malleolus.

8. Discussion and conclusions

The measurement of pressure at biomedical interfaces is complicated on the one hand by the continuum nature of the interface media and on the other by the intrusiveness of the pressure transducer. The latter problem may be ameliorated to some extent using low profile, small foot print transducers with flexible contoured packaging designed to conform to the biomedical interface shape. Biomedical media compatibility is a more difficult problem to solve. However, the results obtained with modified MEMS BIPT devices presented here for a range of biomedical interface environments are encouraging. These transducers have been rendered bio-media compatible by filling the sensor port with two gels, a MEMS protection gel and an interface gel. The first gel protects the delicate silicon diaphragm, chip and bond wires while the second interface gel couples the biomedical tissue or pressure applying element to the sensor. A contoured anti-hammocking guard ensures there is no void or lift-off zone created by the sensor at the interface. Since MEMS pressure sensors are manufactured to operate across a wide range of pressures, devices may be matched to specific applications, i.e. 0-100 mmHg for sub-bandage pressure measurement; 0-500 mmHg for sub-tourniquet measurements and 0-750 mmHg or higher for prosthetics.

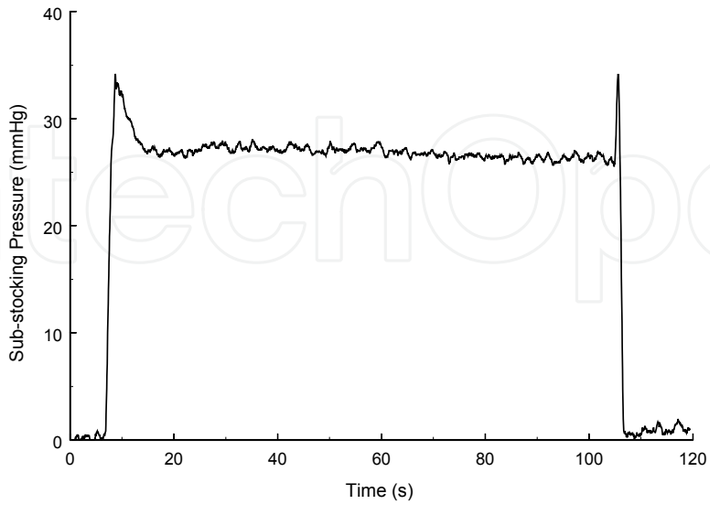

Fig. 14. Interface pressure measured under a Medivan CCL1 support stocking with the sensor located above the MM.

While the primary operating mechanism of the BIPT is the deflection of an elastic diaphragm, the actual deflections involved are microscopic and therefore negligible on the scale of the overall BIPT size. The transducer is effectively a constant volume device. In particular, the sensing volume comprising the two-gel filled sensor port does not change under typical biomedical pressures and so pressure dependent hammocking artifacts are avoided. Transducers may be reliably calibrated using inflated bladders, set-point loads or, as done routinely in this work, by using a pressure air-line coupled to a suitable sensor shroud, i.e. plastic chamber which provides an air-tight seal around the sensing area. The latter approach has the advantage of being independent of the actual interface/application environment. The sensor performance characteristics, such as calibration stability, sensitivity, response time and hysteresis of these devices stays within or very close to the MEMS pressure sensor manufacturer's specification.

MEMS modified BIPTs may also be configured as multisensor arrays. With footprints less than 0.25 cm^2 linear arrays with sensor density of 2/cm are feasible. Two sided flexible circuit interconnect with vias/through-holes would allow two dimensional arrays with sensor density of $4/cm^2$. Lower density two dimensional arrays could be implemented using single sided flexible circuit provided there is space to route the interconnect tracks between the devices. While it is useful to know absolute or peak pressures in many medical and biomedical settings, there are many instances when knowledge of the local pressure gradients is equally important. For example, in the management of venous leg ulcer disease, establishing a pressure gradient from ankle to knee using compression is an essential part of the treatment. A wide-spacing extensive linear array of BIPTs would facilitate the application of the correct pressure gradient and provide objective evidence of good bandaging technique which is 'operator dependent' thereby improving safety and aiding training. Equally, the simple close-spaced three sensor configuration described here can provide useful information

on pressure gradients under pneumatic and non pneumatic tourniquets indicating hazard conditions and cuff tightness.

In close-spaced array form, the PDMS guard/flexible-circuit combination acts as a carrier for the individual sensors. The overall structure can conform to limb curvature while maintaining a stable interface contact zone. For instance a linear BIPT paediatric array could accommodate in excess of 20 pressure sensors on a limb with radius of curvature of 2 cm using currently available devices (array thickness less than 3 mm). However, while PDMS is compliant, it is significantly stiffer than the target soft tissue for some biomedical applications. In the measurement of biomedical interface pressures there is, in general, a trade off between compliancy and accuracy. Compliant, totally conformable sensors present very significant calibration and stability challenges in real applications. Rigid structures provide the highest performance specifications but are intrusive and may be subject to significant measurement artifacts. The design used here minimizes the compromise on performance while addressing the compliance/conformability need to a useful extent for applications involving sub-bandage and sub-tourniquet settings.

The temperature change from room to body temperature produces changes in the span and zero offset voltage in simple voltage driven piezoresistive MEMS devices. There are many standard temperature compensation techniques available for full-bridge piezoresistive elements ranging in sophistication from simple passive component circuits to microcontroller implemented correction algorithms using integrated temperature measurement. As the offset voltage temperature-sensitivity is significantly larger than that of the span sensitivity, it is possible to compensate for offset, in many instances, by simply re-zeroing the sensor when on-body after allowing sufficient time for the device to reach body temperature. Gauge and differential sensor configurations should not require ambient pressure compensation provided the reference port is vented to the ambient atmosphere. However, this may not always be possible or advisable in biomedical interface pressure measurement since the open port can present a contamination risk and compromise the integrity of the sensor as well as presenting sterilisation problems for reusable devices. If the reference port is sealed or if absolute devices are used, then ambient pressure variations may be monitored using a reference sensor, and appropriate corrections applied to measured interface pressures.

The BIPT devices show residual sensitivity to shear and frictional forces. These can vary with body tissue properties as seen in the sample population data presented. Such shear forces can be reduced using protocols specific to the particular application. For instance, shear contributions with pneumatic tourniquet cuffs may be reduced by using an initial inflate-deflate cycle to 'settle' the cuff onto the limb. However, even with such protocols, standardised procedures and controls, there is still significant scope for variability in results, since 'no two humans, even of the same weight and stature, are anatomically identical' (Shelton et al, 1998). This situation is likely to continue unless a completely shear independent biomedical interface pressure transducer emerges. Clearly, a lubrication gel may be added to the sensor-tissue interface to decouple such shear forces. However, the use of such gels could pose a contamination problem in surgical applications and has therefore been avoided in this work which targets a fully solid state biomedical interface pressure measurement solution.

On-going developments in the MEMS pressure sensor industry in the area of miniature altimeters and depth gauges as well as navigation technology is resulting in ever smaller devices. The availability of such small MEMS devices will allow for further reduction in the intrusiveness of BIPTs, and, increases in the spatial resolution of BIPT arrays. Clearly, for volume market applications, custom MEMS package designs optimised for the measurement

of pressures at biomedical interfaces could be justified using bare chip MEMS available from many foundries. Bond wire strain-relief-loops to the top-side of MEMS silicon die adds to overall device. Custom devices with under-side bonding of the MEMS die to the substrate carrier would significantly reduce overall transducer height. Combined with flexible surface mount technology, integrated 16 bit analogue to digital converters and digital interfaces, it should be possible to develop biomedical interface pressure measurement products and applications which are minimally intrusive, compliant, temperature and ambient pressure compensated, and which can be reliably calibrated, all at relatively low cost. Further reduction or complete elimination of the residual shear sensitivity of these devices will be the focus of future work aimed at developing a general purpose biomedical interface pressure transducer which does not call for any special 'standardised' procedures or application specific protocols, in order to yield reliable biomedical interface pressure data.

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10. References

Allen, V., Ryan, D. W., Lomax, N. & Murray, A., (1993) Accuracy of interface pressure measurement systems, *J. Biomed. Eng.*, Vol. 15:344-348

Aziz, E. S., (2009) Tourniquet use in orthopaedic anaesthesia, *Current Anaesthesia & Critical Care*, Vol. 20:55-59

Tissue Viability Society (2010) Laboratory measurement of the interface pressures applied by active therapy support surfaces: A consensus document, *J. Tissue Viability* Vol. 19:2-6

Barker, B., (2000) Interfacing Pressure Sensors to Microchip's Analog Peripherals, Microchip Technology Inc., Data Sheet AN695.

Buis, A. W. P. & Convery, P., (1997) Calibration problems encountered while monitoring stump/socket interface pressure with force sensing resistors: Techniques adopted to minimise inaccuracies, *Prosthetics and Orthotics International*, Vol. 21:179-182

Casey V., Griffin, S. & O'Brien, S. (2001) An investigation of the hammocking effect associated with interface pressure measurements using pneumatic tourniquet cuffs. *Medical Engineering and Physics*, Vol. 23:511-517

Casey, V., O'Sullivan, S. & McEwen, J. (2004) Interface pressure sensor for IVRA and other biomedical applications, *Medical Engineering & Physics* Vol. 26(No.2):177-182

Casey, V. & Little, E., (2010) Novel Interface Pressure Transducers for Tourniquets and Other Medical Devices, *CMBEC 33*, Vancouver, Canada.

Casey, V., McAree, B., Moloney & M.C., Grace, P. (2010) Wearable sub-bandage pressure measurement system, *IEEE Sensors Applications Symposium (SAS)*, Limerick, pp.41-45.

Cristalli, C. Ursino, M. Palagi, F. & Bedini, R., (1993) FEM simulation and experimental evaluation of the 'squeezing' phenomenon in Riva-Rocci blood pressure measurement, *Computers in Cardiology*, 655-658

Doyle, G. & Taillac, P. (2007) Tourniquets: A review of current use with proposals for expanded prehospital Use, *Prehospital Emergency Care*, Vol. 12(No.2):241-256

See 'World's smallest barometric sensor'; URL:*http://www.epcos.com* Also ABS1200E data sheet.
Fay, B. & Brienza, D., (2000) What is Interface Pressure?, *Proc. 22nd Annual EMBS Conf.*, Chicago, pp.2254-2255
Ferguson-Pell, M. & Cardi, M. D., (1993) Prototype development and comparative evaluation of wheelchair pressure mapping system, *Assist. Technol.*, 78-91
Ferguson-Pell, M., Hagisawa, S. & Bain, D.(2000) Evaluation of a sensor for low interface pressure applications, *Medical Engineering & Physics*, Vol.22:657-663
Fludger, S. & Bell, A., (2009) Tourniquet Application in a Rural Queensland HEMS Environment, *Air Medical Journal* Vol.28:291-293
Giswold, M.E. & Moneta, G.L., (2005). Non operative treatment of chronic venous insufficiency, *in* Rutherford R.B. (ed.), *Vascular Surgery* 6th Ed., Vol.2, Philadelphia, pp.2241-2250
Glinz, K.L. & Jameson, M., (2010) Development and technical evaluation of military tourniquets and emergency tourniquets for pre-hospital settings, *CMBEC 33*, Vancouver, Canada.
Ghosh, S., Mukhopadhyay, A., Sikka, M. & Nagla, K.S., (2008) Pressure mapping and performance of the compression bandage/garment for venous leg ulcer treatment, *J Tissue Viability*, Vol.17:82-94
Grace, P. A. (2003) The Management of Leg Ulcers, Editorial, *Irish Medical Journal*, Vol. 96:37
Intersema Application Note, AN508, "Mounting of Pressure Sensor Dice", Intersema Sensoric SA, URL:www.intersema.ch
Kragh Jr., J.F. , Walters, T.J. , Baer, D.G. , Fox, C.J. , Wade, C.E. , Salinas, J. & Holcomb, J.B. (2009) Survival with emergency tourniquet use to stop bleeding in major limb trauma, *Ann Surg* , Vol.249:1Ű7
Lai, C. H. Y. & Li-Tsang, C. W. P., (2009) Validation of the Pliance X System in measuring interface pressure generated by pressure garment, *Burns*, Vol. 35:845-851
Lee, C., Porter, K. M, & Hodgetts, T. J., (2007) Tourniquet use in the civilian prehospital setting., *Emerg Med J.* Vol.24:584-587
See 'NPI-12 Stainless Steel Media Isolated Sensor' Lucas NovaSensor, URL:http://www.ge-mcs.com/en/pressure-mems/mems-sensors/npi-12-series-novasensor.html
Lyons, G.M., Leane G.E. & Grace P.A., (2002) The effect of electrical stimulation of the calf muscle and compression stocking on venous blood flow velocity *Eur J Vasc Endovasc Surg* Vol.23:564-566
McEwen, J. A. (1994) Apparatus for Intravenous Regional Anesthesia, *US Patent Number 5352195*
McEwen, J.A., Inkpen K. B. & Younger A., (2002) Thigh tourniquet safety. Limb occlusion pressure measurement and a wide contoured cuff allow lower cuff pressure. *Surgical Technologist*, Vol.21:55-62)
Mootanah, R. & Bader, D. (2006) *Pressure Sensors*, Wiley Encyclopedia of Biomedical Engineering, Online Publication, Wiley
Noordin, S., McEwen, J., Kragh, J., Eisen, A. & Masri, B., (2009) Surgical Tourniquets in Orthopaedics, *J Bone Joint Surg Am* Vol.91:2958-2967
O'Brien, S. & Casey, V., (2002) Numerical and asymptotic solutions for hammocking. *Quart J Mech App Math* Vol.55:409-420

Oomens, C., Loerakker, S. & Bader, D., (2010) The importance of internal strain as opposed to interface pressure in the prevention of pressure related deep tissue injury, *Journal of Tissue Viability*, Vol.19:35-42

Paris-Seeley N.L., Romilly D.P. & McEwen J.A. (1995) A compliance-independent pressure transducer for biomedical device/tissue interfaces. Proceedings of RESNA (Rehabilitation Engineering Society of North America Conference), Montreal

Partsch, H., Clark, M., Bassez, S., Benigni, J.P., Becker, F, Blazek, V., et al. (2006) Measurement of lower leg compression in vivo: recommendations for the performance of measurements of interface pressure and stiffness: consensus statement. *Dermatol Surg*, Vol.32(No.2):224-232

PicoPress Compression Measurement System, Microlab Elettronica, Italy, URL:*www.microlabitalia.it*

Pliance System, novel Germany, URL:*http://novel.de/novelcontent/pliance*

Polliack, A. A., Sieh, R. C., Craig, D. D., Landsberger, S., McNeil, D. R. & Ayyappa, E, (2000) Scientific validation of two commercial pressure sensor systems for prosthetic socket fit, Prosthetics and Orthotics International, Vol.24:63-73

Rush, R.M., Beekley, A.C., Puttler, E.G. and Kjorstad, R.J., (2009) The mangled extremity, *Curr Probl Surg*, Vol.46:851-926

Shelton, F., Barnett, R & Meyer, E. (1998) Full-body interface pressure testing as a method for performance evaluation of clinical support surfaces, *Applied Ergonomics*, Vol.29(No.6):491-497

Shelton, F. & Lott, J. W., (2007) Conducting and interpreting interface pressure evaluations of clinical support surfaces, *Geriatric Nursing*, Vol.24:222-227

Shear Force Initiative, National Pressure Ulcer Advisory Panel (US).

Tien, H.C., Jung, V., Rizoli, S.B., Acharya, S.V. & MacDonald, J.C. (2008) An evaluation of tactical combat casualty care interventions in a combat environment, *J Am Coll Surg*, Vol.207:174-178

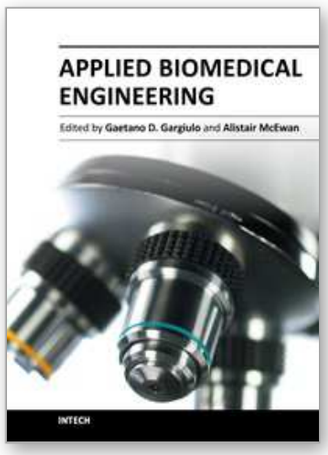
Van den Kerckhovea, E., Fieuwsc, S., Massagéd, P., Hierneri, R., Boeckxe, W., Deleuzef, J.P., Laperreg, J & Anthonissenh, M., (2010) Reproducibility of repeated measurements with the Kikuhime pressure sensor under pressure garments in burn scar treatment, *Burns*, Vol.33:572-578

van Hout, J., Scheurer, J. & Casey, V. (2003) Elastomer microspring arrays for biomedical sensors fabricated using micromachined silicon molds, *J. Micromech. Microeng.*, Vol.13:885-891

Walters, T.J., Wenke, J.C., Kauvar, D.S., McManus, J.G., Holcomb, J.B. & Baer, D.G., (2005) Effectiveness of self-applied tourniquets in human volunteers *Prehosp Emerg Care* Vol.9(No.4):416-422

Wertheim, D., Melhuish, J., Llewellyn, M., Hoppe, A., Williams, R. & Harding, K., (1998) An integrated instrumentation approach to the study of wound healing, Proceedings 20^{th} Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Vol. 20:1760-1761

Wise, K. D., (2007) Integrated sensors, MEMS, and microsystems: Reflections on a fantastic voyage, *Sensors and Actuators A: Physical*, Vol.136:39-60



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This book presents a collection of recent and extended academic works in selected topics of biomedical technology, biomedical instrumentations, biomedical signal processing and bio-imaging. This wide range of topics provide a valuable update to researchers in the multidisciplinary area of biomedical engineering and an interesting introduction for engineers new to the area. The techniques covered include modelling, experimentation and discussion with the application areas ranging from bio-sensors development to neurophysiology, telemedicine and biomedical signal classification.

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