TEMPERATURE AND PRESSURE FIBER-OPTIC SENSORS APPLIED TO MINIMALLY INVASIVE DIAGNOSTICS AND THERAPIES

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ABSTRACT

We present how fiber-optic temperature or pressure sensors could be applied to minimally invasive diagnostics and therapies. For instance a miniature pressure sensor based on micro-optical mechanical systems (MOMS) could solve most of the problems associated with fluidic pressure transduction presently used for triggering purposes. These include intra-aortic balloon pumping (IABP) therapy and other applications requiring detection of fast and/or subtle fluid pressure variations such as for intracranial pressure monitoring or for urology diagnostics. As well, miniature temperature sensors permit minimally invasive direct temperature measurement in diagnostics or therapies requiring energy transfer to living tissues. The extremely small size of fiber-optic sensors that we have developed allows quick and precise *in situ* measurements exactly where the physical parameters need to be known. Furthermore, their intrinsic immunity to electromagnetic interference (EMI) allows for the safe use of EMI-generating therapeutic or diagnostic equipments without compromising the signal quality. With the trend of ambulatory health care and the increasing EMI noise found in modern hospitals, the use of multi-parameter fiber-optic sensors will improve constant patient monitoring without any concern about the effects of EMI disturbances. The advantages of miniature fiber-optic sensors will offer clinicians new monitoring tools that open the way for improved diagnostic accuracy and new therapeutic technologies.

Keywords: point measurement, *in situ* monitoring, Fabry-Pérot sensor, high sensitivity, EMI immunity, miniaturization, catheter tip mounted medical transducer, intra-aortic balloon pumping (IABP) therapy, intracranial pressure monitoring, urodynamics, small animal testing, MRI monitoring.

INTRODUCTION

Rapid development over the last two decades of fiber-optic based telecommunication networks has led to the creation of high-sensitivity control and measurement systems based on fiber light-guides. Along with the advances in fiber-optics, micro-opto-electro-mechanical systems (MOEMS) produced by photolithography processes derived from the semiconductor industry increased production capacities thus decreasing prices and opening up a whole range of interests in the field of sensing technologies. For many years now, fiber-optic sensors (FOS) have been successfully used in various markets such as aerospace, the aircraft industry, nuclear plants, civil engineering and the oil and gas industry.

With better availability, packaging adaptation and democratization of fiber-optic technology, FOS have now expanded their medical applications which possess their own interest in these types of sensors. Price reduction and technology acceptance are among the main issues that have so far driven the development of reliable medical FOS. Affordable and disposable FOS have begun to be available on the market. Many interesting physiological parameters such as pH, dissolved oxygen, and even glucose levels are accessible using such sensors, but body temperature and fluid pressure measurements currently still remain the main fields of interests.

Because of the electrically insulating nature of fiber and sensor materials, these sensors are not disturbed by incident radio frequency (RF), electromagnetic (EM) or microwave (MW) fields. They also do not disturb surrounding EM field distribution and therefore do not interfere with equipments relying on EM fields as a means of producing images or producing effects on *in vivo* tissues. This makes the FOS technology a perfect candidate for physiologic sensing solutions in the presence of equipments generating or using RF, EM, or MW fields now often seen in clinical environments.

Miniaturization of such sensors as well as the use of inert components in making optical sensors permits *in vivo* and *in situ* measurements in restricted areas such as small vessels or in delicate tissues such as the brain. With the increased development of minimally invasive surgery using instrumented catheters there is now a growing demand for miniaturized, reliable, and accurate sensors mainly for temperature and pressure measurement in medical fields such as cardiology, neurology and urology. Responsiveness combined with high sensitivity is another FOS asset exploited in applications requiring detection of fast and/or subtle physiological changes. Typical applications would be in intra-aortic balloon pumping (IABP) therapy, intracranial pressure monitoring, or urology diagnostics which will be discussed in more detail in subsequent sections of this article along with other medical applications for temperature measurement.

MINIATURE MEDICAL FIBER-OPTIC SENSORS AND TECHNOLOGY

Most of the FOS developed by FISO Technology are constituted by a Fabry-Pérot cavity whose optical length changes with the physical parameter to be measured. The sensor is connected to a multimode optical fiber which acts as the light conveyor between the sensor and the signal conditioner. White light from a lamp is directed towards the Fabry-Pérot cavity which modulates the signal with a low coherence interference thus coding the sensor cavity length. The wavelength-modulated optical signal is then reflected back towards the signal conditioner which extracts the cavity length information using patented[1] white-light cross-correlation technology.

Some commercially available sensors suitable for medical applications are presented unpackaged in Figure 1. They are usually packaged into catheters or other medical devices according to the specific needs of the application.

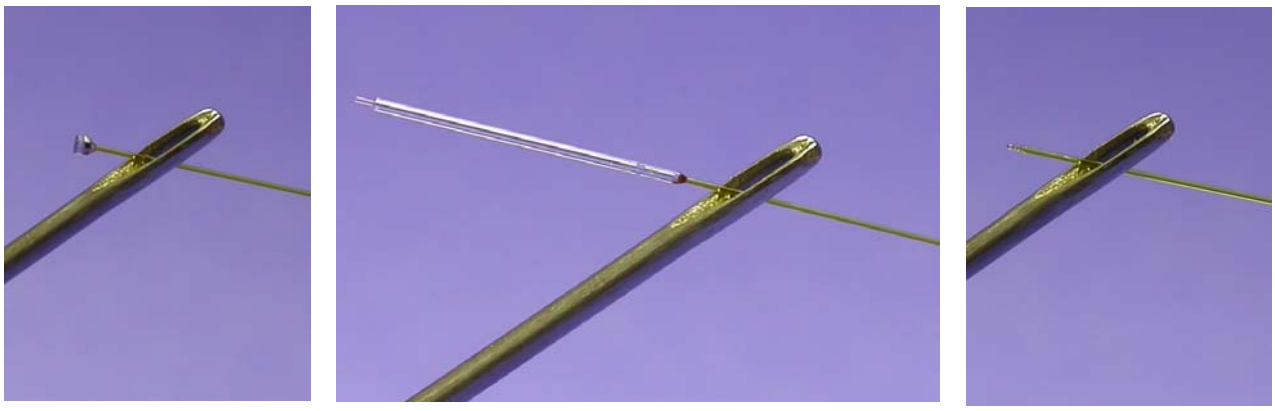

Figure 1: Bare fiber-optic sensors for medical applications commercially available from FISO Technologies. Left: pressure sensor FOP-MIV (∅ 550 µm), Center: temperature sensor FOT-L (∅ 800 µm), Right: temperature sensor FOT-MSP (∅ 210 µm)

The FOP-MIV pressure sensor (∅ 550 µm) is a miniaturized Fabry-Pérot cavity constituted by a micromachined silicon diaphragm membrane, acting as the pressure sensing element, bonded on a cup-shaped glass base. Using a vacuum inside the cavity prevents changes of internal pressure caused by gas thermal expansion that would otherwise distort the pressure measurements. Since a high vacuum is always maintained inside the cavity, the FOP-MIV is always measuring absolute external pressure. When pressure is increased on the sensor, the thin membrane is deflected and the Fabry-Pérot cavity length is reduced. Such small cavity length variations are translated to pressure variations by the signal conditioner thanks to factory calibration relating cavity lengths to pressure. The FOP-MIV pressure sensor is produced in large quantities using manufacturing technologies derived

from the semiconductor industry (photolithography processes and automated assembly). Being one of the smallest pressure sensors available commercially, it is well designed for many medical applications such as the one that will be described in this paper, but it could also be used in other non-medical applications where size is an important issue.

The FOT-L temperature sensor (∅ 800 µm) has a Fabry-Pérot cavity constituted by two optical fibers precisely assembled into a glass capillary. The cavity length changes with temperature variations due to differences in the coefficient of thermal expansion (CTE) between optical fibers and the glass capillary. Such variations are proportional to temperature changes which are thus evaluated using a calibration factor. Due to the length of the rigid glass capillary (10 mm), the very sensitive FOT-L is mostly used in medical applications where this does not cause any problems such as in applications of skin temperature monitoring. When more flexibility is required and lower sensitivity is acceptable such as for some instrumented catheters, the FOT-MSP temperature sensor (∅ 210 mm) should be preferred. For such sensors the Fabry-Pérot cavity, constituted by a transparent semiconductor material, also changes with temperature. Due to the incredibly small size of this sensor, the thermal inertia is virtually reduced to zero allowing ultra fast temperature monitoring.

MEDICAL APPLICATIONS FOR FIBER-OPTIC SENSORS

Clinical use of FOS is one of the most demanding environments challenging the development of such sensors for real commercial successes, particularly for minimally invasive applications. Even submitted to medically harsh conditions, the sensor should always be reliable since sometimes a patient's life could depend on it. Besides specific requirements on material biocompatibility, sterilization resistance, long-term stability and accuracy, and on low degradation by the physiological fluids, high quality standards medical FOS are also highly sensitive to price concerns since in most applications they have to be disposable. Most of the time they are just one part of a complex system and their use should remain as friendly as possible to the end-user. In fact, since FOS technology is relatively new to the medical community, most medical technical staff still have only a minimum knowledge of fiber-optics and on how to properly handle such devices. If integrating FOS technology into medical diagnostics and therapies represents a great challenge, the advantages of such technology are often unique which allows for interesting new approaches as will be clearly demonstrated in the following examples.

Intra-aortic balloon pumping therapy

The intra-aortic balloon pumping (IABP) therapy consists of inserting, generally through the femoral artery, a catheter terminated by an inflatable balloon which is then positioned into the descending aorta just below the subclavian artery. Once in place, the balloon is timed to inflate at the onset of the diastole (increasing diastolic blood pressure and myocardial oxygen supply) and to deflate just prior to ventricular ejection (decreasing blood pressure in the aorta, left ventricular afterload and myocardial oxygen demand). Such counter-pulsation therapy improves mean arterial pressure and cardiac output, decreases left ventricular preload and afterload and as a result the cardiac work. It also allows for improvement in blood pressure quality in the superior vessels thus also improving the oxygen delivery to the brain.

Such therapy developed more than 30 years ago[2-4] now is one of the most popular forms of life-supporting mechanical assistance usually used when pharmacologic therapy fails or presents a high risk of mortality or morbidity due to high drug doses. This therapy is often used temporarily to help patients recover from critical heart diseases or to wait until a transplant is performed. As it is easily understood, synchronization of balloon inflation and deflation is critical for the success of such therapy. It is usually performed using the patient's electrocardiogram (ECG) or arterial pressure waveform. The use of ECG for accurate triggering is not always possible due to tachyarrhythmia, cardiac pacemaker functions, poor ECG signal, or the use of electrical surgical tools creating a burst of interference. In such situations arterial waveform monitoring is preferred and pressure is usually measured externally to the patient using fluidic pressure transduction through the catheter and interconnecting hydraulic tubing. However, due to pressure artefacts generated by fluidic transduction such approaches have intrinsic limitations. First, hydrostatic pressure generated by level differences between the catheter's end tip and the external measuring unit has to be subtracted each time the patient changes his position (generally the patient should stay lying down and the correction is performed at the beginning of the therapy). The greatest disadvantage of fluidic transduction is indeed the limitations concerning inaccurate dynamic response since fluid-transducted pressure is affected by the length and

diameter of the tubing (dissipation of energy due to fluid mass, friction on the walls and to tubing compliance, especially if the diameter is reduced), by the presence of tiny bubbles inside the fluid-filled system, and finally by possible tubing movements.

Another interesting approach avoiding all the fluidic transduction problems consists of integrating a pressure sensor directly at the tip of the catheter, exactly where the aortic pressure waveform has to be measured. With miniaturization and mass production available thanks to MOMS technology and to assembly automation, FOS such as FOP-MIV could advantageously be the solution for IABP therapy[5].

In order to illustrate the differences between *in situ* pressure monitoring and fluid-transducted external pressure monitoring, we simulated several clinical situations as illustrated in Figure 2. A Bio-Tek 601A pressure simulator was used to generate a pressure waveform inside a ~3 m 8 French (∅ 2.75 mm) catheter. Two FOP-MIV (300 mmHg) fiber-optic sensors were connected to the pressure line, one near the pressure generator (S1) and the other at the end of the catheter (S2). Pressure was monitored using two PM-250 from FISO Technologies operating at 250 Hz and connected to a digital scope. Two types of waveforms were selected in order to illustrate the dynamic response of the sensors depending on their location: a square waveform (80/140 mmHg, 0.5 Hz = 30 bpm, top graphs) and a normal aortic waveform (80/140 mmHg, 1.5 Hz = 90 bpm, bottom graphs) for physiological simulation. It could be seen on graphs A from Figure 2 that the FOP-MIV sensor S1 located near the pressure generator records the pressure with high fidelity since the square waveform does not have any dynamic artefact. The aortic waveform shows a characteristic shape[6] where the systolic (S) and diastolic (D) events could clearly be separated by the dicrotic notch event (DN arrow) indicating the aortic valve closure and thus the transition between the systole and the diastole. The detection of this important minute pressure event is critical for IABP therapy since it indicates the exact moment when the intra-aortic balloon should be inflated. The need for a sensitive sensor capable of clearly identifying the dicrotic notch is actually essential. Balloon deflation is completed at the end of the diastole which is more easily detected by the rapid pressure increase observed at the beginning of the systole.

For the sensor located at the extremity of the catheter (S2) the recorded pressure is in fact significantly different from the pressure generated by the Bio-Tek 601A pressure simulator. It can be seen on the square waveform (left part of graphs B, before 2 s) that pressure overshoots and undershoots are present indicating that fluidic pressure transduction is in fact a secondary dynamic system. Such pressure bouncing impacts the aortic waveform by artificially increasing the visibility of the dicrotic notch but also by slightly shifting its time position. Such an effect is not recommended for precise triggering purposes such as required in the IABP therapy.

On the right part of the graphs B in Figure 2 (after 2 s) the fluid-filled catheter has been manually vibrated in order to simulate vibrations that could occur in real IABP therapy such as the one encountered during patient transportation. It could be seen on both graphs (square and aortic waveforms) that the vibrations of the catheter are creating erratic pressure fluctuations which make it virtually impossible to locate the dicrotic notch on the aortic pressure waveform. Interestingly, during the disturbance, the sensor S1 located near the pressure generator did not see the erratic variations in pressure (pressure waveforms before and after 2 s are identical as seen in graphs A).

Another effect that could change the pressure waveform in fluid-filled catheters is the presence of tiny bubbles that are sometimes difficult to localize and to remove, requiring additional time for settling the IABP therapy. An illustration of the effect of the presence of bubbles in the catheter is presented in graphs C of Figure 2 as recorded by S2 sensor. Bubbles are in fact compressible air that creates a damping effect on pressure fluidic transduction; thus the highest frequencies are attenuated as seen on the square waveform that now has more of a trapezoidal shape. The true pressure is also attenuated since part of the pressure energy is absorbed by bubbles compression. The resulting aortic pressure waveform is dramatically modified in such non-optimal conditions: the dicrotic notch is no longer visible making intra-aortic balloon triggering impossible. Besides the presence of bubbles, such effects could also be observed when elastic or compliant material are part of the fluidic transduction system which usually are reduced by design but which can not be completely removed since the catheters have to be flexible.

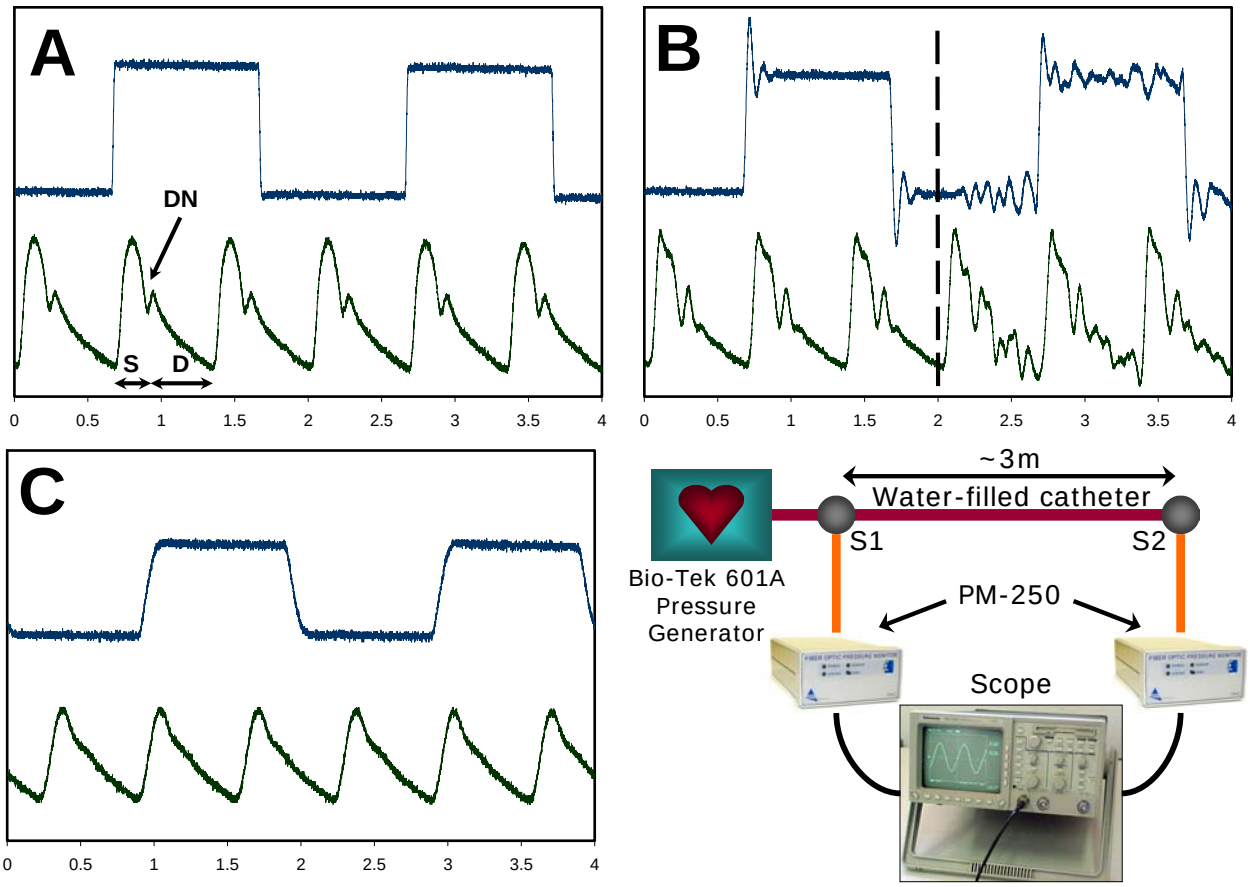

Figure 2 : Pressure waveforms (fluctuating with time expressed in seconds between 80 mmHg and 140 mmHg; Top graphs: square pattern 0.5 Hz = 30 bpm; Bottom graphs: normal intra-aorta pattern 1.5 Hz = 90 bpm) generated by Bio-Tek 601A pressure simulator as recorded by FOP-MIV sensors. A: Sensor S1 located near the pressure generator. B: Sensor S2 located at the end of a 3 m fluid-filled catheter (agitated after 2 s). C: Sensor S2 located at the end of a 3 m fluid-filled catheter having bubbles. Bottom right: Experimental setup.

These examples simulating real life situations clearly demonstrate that *in situ* pressure monitoring is definitely more accurate and safer than external pressure monitoring via fluid-filled catheters. The use of FOS for IABP therapy should therefore provide a better control of triggering at the right time. Furthermore, it should be noted that the current intra-aortic catheter size of 7.5 to 8 French (∅ 2.5 mm to 2.75 mm) is probably the present physical limit for fluid pressure transduction[7]. As a matter of fact, diameter reduction of the catheter is highly desirable in order to reduce the incidence of IABP vascular complications such as ischemia which represent the highest risk of the therapy. Integration of miniature pressure FOS into IABP catheters obviously offers such an opportunity without compromising the accuracy.

Intracranial pressure monitoring

The brain is contained in the skull, a rigid container, and any liquid accumulation such as blood or cephalospinal fluid (CSF) or mass lesions such as tumors, pus or hematoma may increase intracranial pressure (ICP). High ICP is a common cause of death in neurological patients and sustained high ICP suggests poor prognosis[8]. Forty percent of patients admitted unconscious have high ICP. In this group, high ICP will be the leading cause of death in half of cases[9] and effective treatment of high ICP was proven to reduce mortality[10].

There are generally two types of measurements: the punctures (such as lumbar or ventricular punctures) and continuous pressure measurements with implantable catheters. With lumbar and ventricular punctures, the pressure is defined as the pressure necessary to prevent CSF escaping from the introduced needle. The lumbar puncture gives only an indirect measurement that does not always correlate with the ICP[11] and involves a risk of herniation: the slight vacuum created by the loss of CSF in the spinal cord may provoke a displacement of the brain in different skull compartments such as transtentorial herniation (midbrain movement through the tentorial notch) or outside the skull through the Foramen magnum (the skull opening at the neck junction). Repeated pressure punctures measurements also obviously increase the risks of infection[8] and therefore do not allow a continuous ICP monitoring.

As ICP varies continuously and especially at high ICP, a single measurement may be misleading. Therefore, a continuous record of the ICP wave is necessary to avoid missing a sudden rise in ICP[8]. The intraventricular fluid-filled catheter externally connected to a strain gauge is considered to be the gold standard method of ICP continuous monitoring. It offers the most reliable measurement because it can be externally calibrated against a reference. However, fluid-filled catheters have bigger diameters which, besides the other problems already described in the IABP section, complicate the placement in the ventricle when it is displaced or squeezed. It also involves potential injury to other parts of the brain and a serious increase in infection risk after three days of implantation[12].

Another method, called hollow skull (also known as Richmond screw), measures pressure at the dura level. Dura is the first of the tree layers of meninges which envelope the brain. It is composed of two fibrous layers and the external one adheres to the skull. This method does not necessitate a deep catheter insertion into the brain. This method is therefore easier to achieve but studies show that, particularly at high ICP, it underestimates the true ICP[13,14] which could have dramatic impacts.

An implantable catheter-tip instrumented with our MOMS based FOP-MIV optical sensor introduced into the catheter offers the possibility of using significantly reduced diameters (∅ 1.2 mm or smaller). It allows easy insertion up to 1-2 cm in the brain. Such system measures absolute pressure and therefore do not require correction for differences in hydrostatic levels such as is the case with fluid-filled catheters. Even though this system cannot be calibrated *in situ*, studies have shown close correlation between the intraventricular method and the catheter-tip transducer method[15-17].

Also, evaluation of brain tissue injuries often rely on computed tomography (CT) scans and magnetic resonance imaging (MRI). MRI is widely used to evaluate tissue injuries, lesion types and their extent. Therefore, MRI testing is often required in diagnosing the cause of high ICP. As MRI technology uses high-intensity magnetic fields, any material that may be influenced by EM fields or create imaging artefact should be avoided during MRI testing.

As an essential part of high ICP management, ICP monitoring cannot be stopped during MRI procedures. Being intrinsically inert to EM fields, optical sensors do not interfere with MRI and offer the most appropriate solution for continuous and secure ICP monitoring even during MRI procedures. In conclusion, the miniature size of pressure optical sensors, such as FOP-MIV, widens access to intraventricular pressure measurements, considered to be the most accurate method in the determination of the patient's ICP status. MRI compatibility of FOS permits continuous monitoring of critical patients during all diagnostic imaging procedures.

Urology diagnostics

The urinary system is physiologically divided into lower and higher systems. The lower urinary system includes the urethra and the bladder. They form a functional unit whose function is to store and evacuate urine and both can become dysfunctional. The most common health problem related to the lower urinary system is incontinency. It is a major cause of disabilities and discomfort among patients particularly for elderly people. Urine storage capability relies on the equilibrium of pressure within the bladder and the capacity of the urethra to counteract this pressure and seal the orifice. Many parameters may contribute to increasing intravesical pressure such as lower bladder compliance or bladder muscle (detrusor) overactivity. Para-urethral factors determine urethra competency to retain urine against bladder pressure. Incontinency may represent a synergic contribution of different dysfunctions. Although they cannot be completely differentiated, they can be classified into general types which respond to similar treatments.

Urodynamics testing is considered the gold standard method in defining incontinency types[18]. The most important urodynamic parameter is pressure which can be measured in the urethra, the bladder and the abdomen. Pressure may be measured with lumen catheters which are composed of tubing filled out with liquid and connected to an external pressure transducer. The main disadvantage of lumen catheters is the flow obstruction due to their large diameter which may interfere with clinical observations even though relatively small diameter lumen catheters are available. As already mentioned in the section of this paper concerning IABP therapy, there is a limitation of minimal diameter for accurate pressure fluid transduction.

The other means of measuring the pressure is by micro-transducer catheters. Being much smaller in diameter than their lumen counterparts, micro-transducer catheters do not produce significant flow obstruction and permit urodynamic measurements with limited invasiveness and consequently with minimal discomfort for the patient. On the other hand, it may be prone to pressure artefact due to contact with tissues. An example is the side force of the urethra wall on a side-mounted micro-transducer catheter. When properly packaged into a catheter, the miniature size of an optical sensor, such as FOP-MIV, permits a tip front looking mounting on a catheter diameter of 1.2 mm or less. The optical sensor offers the best compromise since it then avoids side tissues contact pressure artefact while still providing minimal flow resistance.

The higher urinary system, composed of the kidney and the ureter that transports urine produced in the kidney to the bladder, is much less accessible than the lower system thus requiring smaller sensors for minimally invasive diagnostics. Pressure measurement is complicated by the fact that the ureter is also not a passive vessel. Peristaltic wave in the ureter helps urine flow down to the bladder. High bladder pressure, anatomical obstruction such as calculi, or physiological dysfunctions may disrupt urinary flow and increase pressure in the kidney. Being relatively soft tissue, the kidney may expand under the increased pressure which is called hydronephrosis. Severe hydronephrosis may lead to renal dysfunction and it is thus very important to treat hydronephrosis quickly and to find its cause. In his diagnosis search, the urologist has access to many different diagnostic means such as X-rays, ultrasound imaging, and even intravenous urography using dyes. Even with these modern diagnostic technologies, the most subtle defects are sometimes not visible and the information is mostly a static imaging of organs under investigation and not a precise evaluation of real pressures which could be a valuable parameter for diagnostics.

Profilography which is a collection of pressure data taken at different levels in the ureter, offers interesting additional clinical information. It produces the pressure profile in the ureter giving an evaluation of the ureter peristaltism. This method is being evaluated as a means of determining ureter peristaltism patterns[19]. It could demonstrate normal and two pathologic ureter pressure profiles: ureteroarrythmic (unsynchronized contraction) and silent profile (dilated ureter without contraction). This technique is still new and further studies should prove its efficiency but the initial findings are interesting. Along these lines, miniature FOS, which allows better accessibility to smaller vessels (such as the ureter), facilitates studies of local pressure profiles and ureter peristaltism giving new avenues for better investigation and treatments of more complex causes of various kidney dysfunctions such as hydronephrosis. The use of miniature FOS applied to urology thus opens up new avenues for better diagnostics with direct *in situ* pressure measurements.

Temperature monitoring

One part of a patient's vital signs is body temperature. It is an important parameter indicating a patient's condition and it is usually continuously monitored for critical care patients. When diagnostics on such patients requires magnetic resonance imaging (MRI) or X-rays, clinicians are faced with the problem of having to stop temperature monitoring during this procedure since it is usually done with conventional electrical sensors. Including preparation and exam, this may mean losing several hours of temperature monitoring that in some cases may be critical. Therefore, using an optical temperature sensor immune to electromagnetic interference (EMI), such as FOT-L, avoids interruption of temperature monitoring. When using fiber-optic temperature sensors for critical patient care, the task of medical technical staff will be greatly simplified since a single monitoring instrumentation will be operational in all diagnostic environments the patient could encounter.

In the field of minimally invasive surgery, radio frequency (RF) waves are used locally via a probe or over a region of the body introduced in a tunnel. This method permits resection of masses or tumor treatment with minimal bleeding and minimal damage to surrounding tissues. However, it also involves close temperature monitoring. The small size and intrinsic immunity of optical sensors to electromagnetic fields permit continuous *in situ* temperature measurements without adding harm to tissues. Integration of ultra-miniature temperature FOS (such as FOT-MSP) into minimally invasive surgery involving the use of RF waves could definitely help the optimal control of these types of therapies.

Animal testing

With advances in genetics, single genotype small animal production offers advantageous models for human body physiology and even for behavior studies. Thus, laboratories often use animals (usually small animals such as rats, mice, rabbits, Guinea pigs or hamsters) to test medicines and responses to different events a human could encounter. This implies the monitoring of diverse parameters (mainly temperature and fluid pressure) in different animal organs and under varying testing modalities such as CT scans or MRI which are not environments commonly accessible with conventional electrical sensors. Being smaller, animals have faster metabolisms which imply a faster detection of varying conditions such as blood pressure. Small animals have therefore faster heat beat rates than humans, typically 328 bpm for rats, 520 bpm for mice, 205 bmp for rabbits, 280 bpm for Guinea pigs and 450 bpm for hamsters. They could also have lower mean pressure than humans. Having virtually no inertia, fast reacting miniature pressure FOS such as FOP-MIV, could obviously be used advantageously for small animal studies.

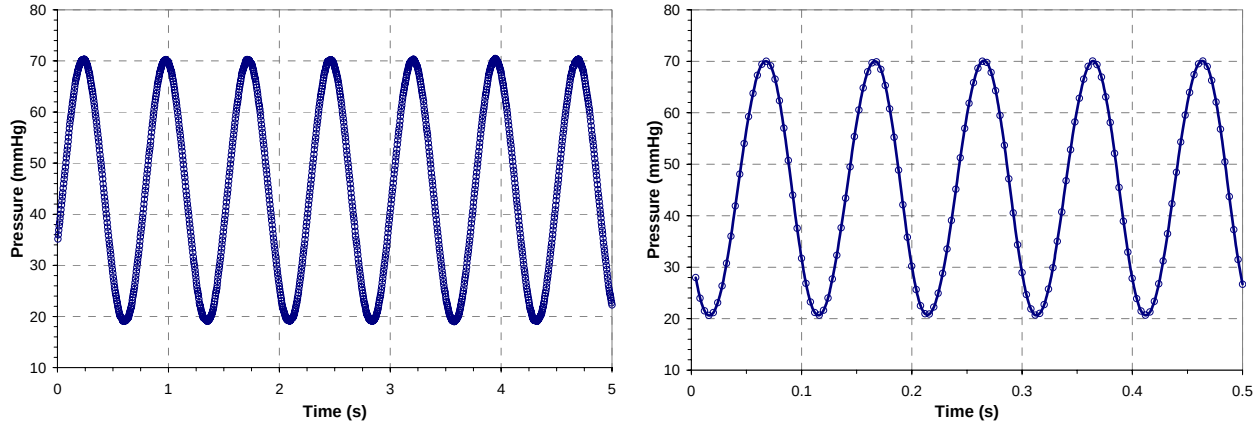

Figure 3 : Sinusoidal pressure waveforms fluctuating between 20 mmHg and 70 mmHg and generated by Bio-Tek 601A pressure simulator and measured with a FOP-MIV sensor (300 mmHg) connected to a PM-250 signal conditioner (left: 1.5 Hz = 90 bpm; right: 10 Hz = 600 bpm)

Figure 3 shows sinusoidal pressure waveforms (at 1.5 Hz = 90 bpm and at 10 Hz = 600 bpm simulating respectively normal human heart beat and small animal tachycardia) recorded with our FOP-MIV sensor connected to a PM-250

signal conditioner operating at 250 Hz. In these graphs, each circle represents an experimental data point (recorded every 4 ms). Pressure variations were generated by a Bio-Tek 601A pressure simulator triggered by an external electrical function generator (sinusoidal function). The graphs illustrate just how accurately the pressure could be monitored for normal human as well as for fast small animal heart beat rates. Actually the experimental frequency response of the FOP-MIV is limited here by the signal conditioner. Faster signal conditioners such as BUS (1 kHz) or Veloce (200 kHz) systems commercially available from FISO Technologies could also be used if needed but for most medical applications the PM-250 (specially developed for pressure monitoring) is usually sufficient and better adapted since it has an internal atmospheric pressure compensation which is valuable particularly for long-term animal studies.

CONCLUSIONS

We have seen that FOS have many characteristics that make them suitable for various medical applications such as those discussed in the selected examples presented in this paper. The miniature size of such sensors is probably the most interesting feature for this field of application since it allows their integration into minimally invasive medical devices and permits direct *in situ* measurement of essential parameters such as temperature and pressure. This opportunity could dramatically change some well established procedures such as avoiding fluidic pressure transduction and associated illustrated drawbacks in favor of a point measurement exactly where it is needed. This concept will certainly revolutionize several diagnostics and therapies in the near future.

Thanks to their small size, FOS have also virtually no inertia and thus respond quickly to rapid physiological changes such as in the case of blood pressure. Being also accurate, they are perfectly suitable for the measurement of subtle changes that could be critical for some diagnostics or therapies. Finally the intrinsic immunity of FOS to RF, EM, or MW interference is a great advantage that not only increases the reliability of such devices in environments with potential interference sources, but also allows using such sensors in environments with high fields, where conventional electrical sensors usually fail. Specific diagnostics or therapies using such fields could therefore greatly benefit from the monitoring opportunities offered by FOS technologies.

Beside the fact that challenging technical and commercial issues still have to be overcome to guarantee the real success of FOS applied to medical applications, such sensors have obviously many unique advantages which open doors for the development of new diagnostics and therapies. We will certainly see in the near future an increased interest in FOS technologies applied to medicine as they will be more clearly understood and become more accessible to clinicians.

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