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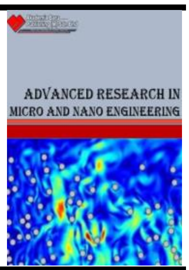

Design Considerations of Implantable MEMS Sensor for Bladder Pressure Monitoring: A Review

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ABSTRACT

One of the main urinary disorders that impair the bladder's capacity to regulate urine flow is urinary incontinence (UI). Currently, the new development of wireless implantable MEMS sensors has been studied to replace current clinical tests of bladder pressure. This new invention grants long-term pressure monitoring within the bladder, less painful with non-surgical, reducing the potential of infection and thus maintains the patient's quality of life. However, developing implanted BioMEMS sensors poses challenging tasks due to various constraint requirements that need to be fulfilled such as appropriate measurement ranges and bandwidth, suitable transduction mechanism, biocompatible materials, structural shape and size as well as wireless telemetry capability. This paper presents a review of implantable MEMS sensors and design challenges specifically for bladder pressure monitoring applications.

1. Introduction

Micro-Electro-Mechanical Systems (MEMS) represent a groundbreaking convergence of technology, bringing together miniature mechanical and electronic components on a single chip. These systems, often at the microscale or even nanoscale, have revolutionized various industries by enabling the creation of compact, efficient, and cost-effective devices. One notable application of MEMS technology is in pressure sensing, where MEMS-based pressure sensors, such as pressure bladder sensors, play a pivotal role. Implantable sensors that are equipped for giving a precise in vivo measurement of target examinations in animals and humans are developing significance in many fields, including clinical treatment, medical diagnostics and personal healthcare. In clinical and

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medical practice, the measurement of intracranial, intraocular, urinary bladder, and circulatory system pressure is quite common [1].

In recent decades, numerous researchers have investigated implantable pressure sensors as a promising alternative to conventional pressure monitoring in medical and therapeutic fields. Despite this interest, only a select few devices, such as Remon ImPressure/RemonbCHF for measuring pressures within an aneurysm sac during endovascular aneurysm repair (EVAR) and CardioMEMS for detecting heart failure, have undergone successful evaluation in clinical trials [2]. Research on other implantable pressure sensors for specific applications such as intraocular pressure (IOP) [3-7], intra-cranial pressure (ICP) [8-11], heart rate and cardiac output [12-15] and bladder monitoring pressure [16-20] are still ongoing research and in the development stage. Incontinence (UI) patients may benefit from an implanted bladder pressure sensor. The condition known as urinary incontinence (UI) impairs the bladder's capacity to regulate the flow of urine. In addition to making people feel uncomfortable and inconvenient, UI can cause major health issues for people, like renal failure [16]. This disease is common among the elderly and diabetic patients. The loss of bladder sensation is a result of afferent nerves impairment to the central nervous system from the bladder [21]. The MEMS pressure sensor that is implanted in the bladder is shown in Figure 1.

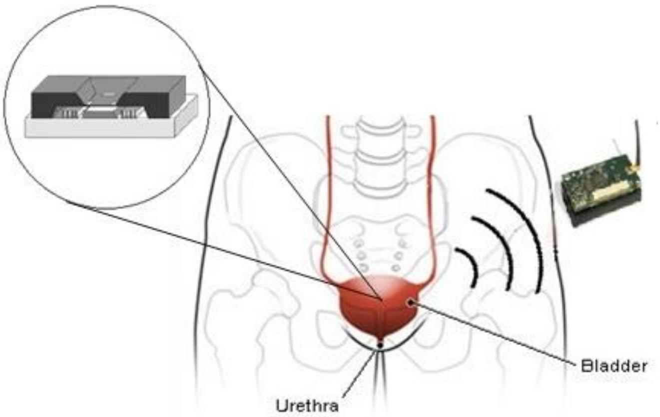

Fig. 1. MEMS pressure sensor implanted in the bladder

In the current clinical investigation, a catheter-based pressure sensor was implanted intraurethrally to measure fluctuations in bladder pressure as part of a cystometry study [2]. Nevertheless, despite being the most reliable approach, the catheterization procedure is painful and uncomfortable for the patients. Furthermore, the catheterization causes problems for the measurements, including breaks in the catheter line, variations in resonance, and response time delays due to fluid viscosity. These tube-like devices are impractical for long-term surveillance since they can lead to infection and bladder stones [17]. MEMS sensors offer appealing prospects for new opportunities of simpler and more practical ways of monitoring bladder pressure as science and technology advance. However, this new invention is a highly multidisciplinary and challenging task that are require minimally invasive, integrable with biocompatible materials, highly sensitive and selective, highly linear output response, and wireless telemetry capability.

Before an implanted MEMS sensor can be commercialized, a number of critical design considerations must be addressed. In this paper, we explore the design consideration behind MEMS pressure sensor development into the specific applications and features of pressure bladder sensing. Considering the design aspects before developing a sensor for a specific application, such as bladder sensing, is crucial for the successful development of the sensor. This thoughtful approach involves taking into account various factors to ensure that the sensor meets the requirements and functions effectively in its intended environment.

2. Design Considerations of MEMS Implantable Bladder Pressure Sensor

There were a number of research that has attempt at bladder urine volume sensing [22-27]. After that, researches on wireless implantable pressure sensor for bladder has been studied for many years [16-20,28-32]. Some researchers propose their own design of pressure sensors used for bladder monitoring and some use the existing sensor available commercially. In order to design an implantable MEMS sensor that meets desired specifications and performs effectively, a number of crucial considerations must be considered. The appropriate sensing technique, measurement accuracy and range, frequency response, size, material, and telemetry consideration are a few of the specifications that are taken into account throughout the design process. Surface acoustic wave (SAW), capacitive, piezoresistive, piezolectric (PZT), strain gauge, and pressure sensing resistor sensor (PSRS) all can be employed as the transduction mechanism or sensing method for pressure sensors. Each sensing method has advantages and disadvantages; thus, the right decision must be made before designing the sensors. Designers should know the relevant measurement range for the input sensor as it affects the precision of the sensor. Hence, in the case of bladder application, it is important to know a specific pressure range for human bladder. The shape of sensor gives enormous impact to the sensor's performance. Square and circle shapes are generally the safe choice for they are easier to be designed and fabricated. Likewise, in the case of implantable device, size is one of the critical factors in sensor's design. In order to avoid surgical procedure, the size of sensor should fit into the intra-urethral catheter to enable it to pass through the bladder without causing any pain and uneasiness to the patients. Additionally, the sensor's material selection is important to ensure the device is compatible with the host into which it is implanted; flexible and stretchable. One of the most important requirements for implantable devices is wireless connection since it eases patient discomfort and lowers the risk of infection or complications from cables. Wireless communications come in two different types: active telemetry and passive telemetry. Passive telemetry is a favourable choice among researchers for being battery-less, hence ensuring the miniaturisation on sensor and more user-friendly delivery system. When assessing a sensing device's performance, key parameters like sensitivity and linearity need to be analyzed [33]. One important component that determines the pressure sensor's accuracy and efficacy is its sensitivity. Sensitivity in pressure sensing is expressed as $S = dX/dP$, where S is the sensitivity and X and P stand for the applied pressure and quantitative output signal, respectively [34]. Linearity is another parameter that gives impact on the accuracy of sensor. In a linear operating range, the responses of a pressure sensor are more precise and consistent. Thus, the creation of highly sensitive pressure sensor with broad linear range for a specified application is highly recommended [34]. The next section discusses in details on design consideration by previous researchers in effort to design MEMS sensor for bladder monitoring application.

2.1 Transduction Mechanism

A transduction mechanism is a physical effect that transfers signals from one form of energy (e.g., electrical, magnetic, mechanical, thermal, chemical, or radiative) to another [35]. To convert pressure into an electrical signal, three main sensing principles are usually employed; piezoresistive, piezoelectric, and capacitive. Each sensing approach has advantages and disadvantages. The capacitive sensing measures the change of capacitance between two parallel plates; when the pressure is applied to one of the plates, caused is moved with respect to the fixed electrode thus causes the change in the capacitance. The term "piezoresistive effect" refers to the change in a material's electrical resistance carried on by an applied pressure in the context of piezoresistive sensing.

The piezoresistive effect changes the resistance of piezoresistors (gauge) coupled electrically in a wheatsone bridge circuit. Proper choice of transduction mechanism is very important in designing MEMS sensor to develop optimal configuration of a device. Selecting the transduction mechanism is much dependent to the application and a set of constraint guidelines. Amongst different types of transduction mechanism, capacitive pressure sensors are one of the most widely applied for implantable devices for bladder [16,17,19,29,31,36] because of their high sensitivity to pressure changes, consume less power, low noise, and low temperature sensitivity [37,38]. However, it suffers a nonlinear output signal [38]. Over the above mention, some researchers used piezoelectric (PZT) [28] piezoresistive [20], strain gauge [27] and pressure sensing resistor sensor (PSRS) [18] in monitoring pressure in the bladder. Selecting the right working principle would also affect the wireless capability of the sensors. Piezoelectric or capacitive sensing principle can be integrated easily with passive telemetry electronics [2] which requiring no incorporated battery.

2.2 Pressure Range and Precision

The bladder pressure sensor should be designed in a way it enables the measuring of pressure for clinically relevant normal and abnormal pressure range. Most of the previous studies in monitoring bladder pressure used pressure sensor for general application which commercially available in the industries [17,19,20,29-31,39]. Majerus *et al.,* used ready-made MEMS pressure sensor (SM 5102 and SM5112) which can detect pressure ranges from 5-300 psi [17,19,29]. These types of pressure sensor are suitable for wide applications include industrial controls, home appliances including medical instrumentation. Even it could be used for bladder pressure detector; however, it would not give a precise output for detecting pressure in the bladder. Only some researchers used specific range of pressure for specifically measure human bladder [18,27,28]. Cao, Hung *et al.,* (2013) estimates the pressure range in a rat bladder in unit of volume [16]. Even it can be used to initiate developing the system for bladder monitoring, but the pressure range is not accurate to detect the human bladder pressure variation. Kim, Albert *et al.,* referring the International Continence Society, (ICS) for typical minimal standard of bladder pressure range which are 0-50 cmH_20 and the maximum abnormal pressure can be up to 200 cmH_20 [28]. Kim *et al.,* [18] and Lee, Ho Young *et al.,* [36] presents the pressure sensor than can detect pressure between 0-10 kPa as they mentioned that is pressure range inside the human bladder. Christian A Gutierrez *et al.,* design strain sensor input pressure of 400-5000 Pa [27] while Giulio Fragiacomo *et al.,*used wide range of pressure; 0-330mmHg to detect internal bladder pressure. Lawrence Yu *et al.,* reviewed that the pressure ranges of bladder relies in between 10-70 mmHg and up to 150 mmHg in voiding condition [2]. The sensor's pressure resolution should achieve ±1 mmHg, an acceptable specification for bladder pressure sensor, relying between 5%–10% of the clinically normal range [2]. Converting the pressure range in the same unit of mmHg concludes that the bladder's pressure range lies between 0–150mmHg.

2.3 Structural and Geometrical Design

The structure design shape affects the sensor's performance vigorously. It is therefore important to identify the best structure design before considering other factors like geometrical size, material, and wireless telemetry. The majority of implantable pressure sensors have a simple design consisting of a membrane and a sealed area; the membrane element responds to and deflects under pressure. [2]. The implantable pressure sensor has been adapted in vivo environment and various applications due to its simplicity in design [9-10,40-43]. Some researchers do modification on the structure such as adding slotted [5,44-46] and spiral [31] to the diaphragm, including altering the design shape; square or circular. Some of them are design Interdigitated structures [16]. Easier way to identify, the best possible structure shape for the MEMS sensor design is by conducting simulative study and comparative analysis of various types of structural shapes by using finite element method (FEM) software such as Comsol Multiphysics CoventorWare, ANSYS, and Intellisuite.

Research conducted by Yusof *et al.,* [47] analyzes in detail the four different diaphragm structures used in MEMS capacitive pressure sensors: clamped, slotted, corrugated, and bossed. The slotted diaphragm outperforms the corrugated, bossed, and clamped structures in descending order of mechanical and capacitive sensitivity, according to the results of the static study.

This comparative analysis may help the designers to choose the most appropriate structure shape meant for achieve their target application and performance. Apart on structure design, size of the structure would give a great effect on sensor's performance. The bladder sensor should be made as small as possible; it should not be larger than a human intraurethral catheter. This is to realize the non-surgical technique of insertion of the sensor into the bladder and to lessen the discomfort associated with the catheterization process. In term of size, Joshua N. *et al.,* suggested that in order to remain within the maximum acceptable comfort zone for urethral installation, the size pressure sensor should have a diameter of less than 10 mm. [32]. Whereas G. Fragiacomo *et al.,* (2010) design pressure sensor in 6 × 6 mm2 wide with thickness below 1 mm, and they claimed this size is small enough to avoid disturbing the bladder, and large enough to produce a measurable signal [31]. Size of the sensor designed by Kim, Jonghyun *et al.,* [18] is 2.6 x 2.6 mm2 with 1mm of thickness. Kim *et al.,* (2014) proposed their PZT circular shaped pressure sensor with radius membrane of 2.75 mm and 50µm thickness enclosed in a glass tube (8mm diameter,40mm long) [28]. Meanwhile, Lee, Ho Young *et al.,* [36] designed their circular capacitive pressure sensor with radius membrane of 2.5 mm and 120 µm thickness. By integrating sensors with application-specific integrated circuits (ASICs) rather than separate electrical components, the total system size can be further decreased. However, the comparatively high size of inductive powering coils and batteries limits the amount of downsizing that may occur [2].

2.4 Biocompatible Material

When considering implantable MEMS sensors, biocompatibility is an essential factor to consider. One of the most acclaimed definitions of biocompatibility is [48] as "the ability of a material to perform with an appropriate host response in a specific application" [49]. Many MEMS sensors using microfabrication technology use bio-incompatible materials such as silicon, poly-silicon, silicon dioxide (SiO_2), silicon nitride (Si_3N_4) and silicon carbide (SiC) [50]. For implantable MEMS sensors, a biocompatible encapsulation is desirable because these materials are crucial for the design of the MEMS sensor. Polydimethylsiloxane (PDMS) [16,18,28,31,51] is one of biocompatible materials that is often used to develop implanted MEMS sensor due to its high flexibility, ease of molding, chemical inertness and offers much higher sensitivity [52]. Ho Young Lee *et al.,* [36] used a natural rubber latex (NRL) for diaphragm's material, however, the future study is needed to confirm its compatibility for implanted sensors. Another compliant biocompatible material like parylene, polymide and polyethylene terephthalate are also currently the subject of in-depth research [52].

Lin *et al.,* [53] explores the development of implantable bladder pressure sensors utilizing a novel biocompatible silicone-oil modified polydimethylsiloxane (PDMS) for encapsulation. The silicone-oil modified PDMS offers heightened elasticity and a reduced Young’s modulus, positioning it as a promising material for sealing pressure sensing devices and forming pressure sensor capsules.

Recent study investigated Buchwalder *et al.,* [54] notice that parylene C's superiority over parylene VT4 grade coating in terms of preventing encrustation on a MEMS sensor system for measuring urine pressure. Further improvement is also investigated by the use of silicon oxide (SiOx) as a final coating. Apart from being costly and complex, biomaterials with mechanical properties are more similar like soft tissue, like bladder is preferable for in vivo application [27]. Graphene and carbon nanotubes (CNTs) have garnered significant interest in a range of applications because of their distinct chemical and physical characteristics. Graphene has also been extensively researched as inactive materials, such as electrode and conductive filler for piezoelectric and capacitive pressure sensor [55]. Nonetheless, medicinal application of graphene is still at the early stage, despite some advancements in diagnosis and drug delivery. Thus, there is a serious need to recognize environmentally friendly and uncomplicated methods in preparing biocompatible graphene materials for biomedical applications [56].

2.5 Frequency Response and Telemetry Capability

The existence of pressure profile is caused by hydrodynamics fluid which is continuously being produced and depleted in the human body. Hence, the pressure sensor design should have the ability to recognize this ordinary and irregular state of physiological hydrodynamics. In other word, the disease status should be displayed through the frequency response of the sensor, which should be chosen appropriately. In the case of bladder pressure monitoring, the typical frequency response is between 3–5 Hz [2,57]. A few studies reported about the wireless implantable in replacing conventional wired pressure monitoring system [17,19-20,28-29]. Damaged tissue problem inside the bladder and infection risk can be reduced with wireless sensing system [31]. In order to create wireless monitoring, the system should not heat tissue or emit more electromagnetic radiation than should be avoided. If the implanted system uses electromagnetic field emission or radio frequency (RF) communication, it must follow recommended human exposure limits for these fields [58]. Wireless sensing can actively or passively be achieved. Passive devices can be measured remotely without a power source and have an unlimited lifespan, but active devices require an implanted microchip that is powered by an inductively connected power supply or battery [31]. This passive telemetry is usually preferred due to simplicity of its circuitry and smaller-size design [2]. Surface Acoustic Waves (SAW) or LC resonators can be used to create a passive wireless sensor, but because of low sensitivity of SAW operation [59], many past studies chose wireless passive LC pressure sensor for bladder pressure monitoring [16,18,28,31,36]. Previous research showed that operating frequencies within the range of 10–100 MHz are considered safe for wireless in vivo measurements when it comes to wireless monitoring sensors [60].

3. Discussion

The comparison of design specifications for bladder pressure sensor from previous studies is summarized in Table 1. Most previous studies used capacitive sensing due to the advantages of this sensing: high sensitivity to pressure change, low noise, consume less power, and low temperature sensitivity, yet it is capable to be easily integrated to passive telemetry. Previous researchers used various ranges of input pressure with many units from mmHg, kPa, cmH_{2O}, and psi for expressing input pressure. By converting it to similar unit in mmHg as shown in Table 1, it can be concluded that the pressure in the bladder relies between 0–150 mmHg. Most of researchers suggest the typical normal pressure in bladder within range 0-75mmHg [2,18,28,36] and for abnormal pressure can be up to 150mmHg [2,28]. Also, from this review, it was found that most researchers used square shape with maximum size 6mm x 6mm and circular shape with diameter less than 10 mm. In the case of capacitive sensing, even smaller size is needed to fit the human catheter, but in general, reducing the size decreases the sensor's sensitivity. The capacitive pressure sensor's high-pressure sensitivity is attained by reducing the sensing gap, widening the diaphragm, and decreasing its thickness [5]. Hence, it is advisable to do size optimization to achieve optimal configuration of the sensor's design and improve the performance particularly sensor's sensitivity and linearity. Likewise, the consideration of bladder's frequency response is not given serious attention in the design process of the sensors even though it is one of the important requirements in designing an implanted MEMS sensor. The typical frequency response for bladder pressure monitoring is between 10–100 MHz, considering the wireless sensing transduction method. From previous studies, most researchers were concerned on the sensitivity measurement to express their sensor's performance. However, not many of them analyzed the linearity performance from the sensors design. High linearity is required in sensors design as it increases the sensor's accuracy. One method that can be applied to increase the linearity is by reducing the diaphragm thickness [5]. Many researchers in the past proposed various types of biocompatible materials and most of them used PDMS due to its advantages: biocompatible, high flexibility, ease of molding, chemical inertness, and offers highly sensitive. However, great potential materials like graphene and carbon nanotubes (CNTs) are good options due to their excellent mechanical chemical and electronics properties. Nonetheless, detailed studies should be conducted to synthesis graphene or CNTs for its suitability usage and fulfill its potential for implantable device. Next, passive telemetry is a favorable choice for the type of wireless sensing for being remotely battery-less and for functioning without any power supply. This reduces the overall size and complexity of the sensor design. Table 2 shows the typical design specifications for bladder pressure sensor summarized from this review. This discussion is hoped to help researchers to consider the critical design specifications before specifically developing sensor to monitor bladder pressure.

Table 1
Commonly used relevant design specifications in the development of bladder pressure sensor
*t= thickness

Author [Ref. No.]	Mechanism	Pressure	Shape and Size	Material	Telemetry	Sensitivity
Cao *et al.,* [16]	Capacitive	1 – 1.6 (vol.) ml	Interdigitated 500 µm x 200 µm	PDMS	Passive	na
Ghoni *et al.,* [28]	PZT	36.78 ~147.11 mmHg	Circular t=50 µm Ø5.5mm	PDMS	Passive	1 kHz/cm$H_2$0
J.Kim and Choi [18]	Resistor Sensing	0-75 mmHg	Square t=500 µm 2.6 mm x 2.6 mm	PDMS	na	0.0360
Fragiacomo *et al.,* [31]	Capacitive	0 -330 mmHg	Square t<1mm 6 mm x 6 mm	PDMS	Passive	650 kHz/mmHg
Lee *et al.,* [36]	Capacitive	7.5-75 mmHg	Circular t=0.12mm Ø5.0mm	NRL	Passive	6385 ppm/kPa

Table 2
Typical design specifications for bladder pressure sensor

Specifications	Value
Pressure range (mmHg)	0-75mmHg(normal) [28,18,36,2] ~150mmHg(abnormal) [2,28]
Sensitivity (kHz/mmHg)	650 kHz/mmHg [31]
Pressure resolution (mmHg)	±1 mmHg [2]
Frequency response (MHz)	10–100 MHz [60]
Maximum Size	Square size : 6mm x 6mm [18] Circular:8mm [28] (encapsulated)

3.2 Current Advancements in Implantable MEMS Sensors for Bladder Pressure Monitoring

The advances in materials science, sensor integration, and microfabrication technologies have contributed to the improvement of the implanted MEMS sensors for monitoring bladder pressure throughout recent years. This section of the paper will discuss the critical advances as well as emerging trends for sensor design that may influence the field of clinical applicability and offer some potential directions for future investigations.

3.2.1 Miniaturization and Integration

Miniaturization is critical in sensor design as it lowers suffering for the patient and enables even less invasive implantation techniques. Microfabrication technologies, such as MEMS and photolithography made it simple to build sensors with compact dimensions as well as improved spatial resolution. Miniaturization reduces surgical complications and tissue damage while also allowing for accurate implantation of the sensor within the bladder. Furthermore, the transformation of bladder-health monitoring has been accentuated by adding yet another step, which involves our abilities to combine multiple sensing modalities in a single device and assess urinary function and pathology simultaneously. For example, the aforesaid combined pressure and temperature sensing, allow the simultaneous monitoring of bladder physiology and the condition of bladder inflammation [61]. It has been acknowledged that this development helped doctors in their treatment decisions and make the diagnosis more precise and simplified the urine data collection.

Different modern approaches have been recently applied including novel materials and fabrication techniques to facilitate the improvement of sensor integration and miniaturization. The utilization of 3D printing and flexible substrates has facilitated the development of tailored and conformal sensor designs due to the high anatomical diversity of patients [61-63]. In addition, nanomaterial-based sensors have been integrated to develop ultra-compact, high-sensitivity, and highly biocompatible devices [64]. Such a focus on sensor performance, its miniaturization, and integration with other parts contribute to the reduction of implantable device, as well as associated hardware, footprint. It simplifies the processes of implementation and makes them less challenging for both healthcare providers and patients. Additionally, compact sensor designs provide opportunities for minimally invasive implantation methods like endoscopic or percutaneous methods, which reduce recovery time and healthcare expenses [65,66].

3.2.2 Wireless Connectivity and Remote Monitoring

Integration of implantable sensors equipped with wireless connectivity allows for remote monitoring and instant data transfer, increasing patient comfort and accelerating therapeutic measures. Development in wireless sensor networks and protocols, such as Zigbee and Bluetooth Low Energy (BLE), ensures smooth connectivity with external monitoring systems. Thus, the continuous monitoring of bladder pressure dynamics in an ambulatory setting becomes an achievable reality [67,68]. In addition to further improvement in clinical observation and patient comfort, wirelessly enabled sensors can significantly contribute to telemedicine and patient remote management efforts. The integration of wireless sensors in healthcare facilitates prompt intervention as well as personalized adjustments to therapy by transmitting real-time sensor data to healthcare providers remotely. This innovation enhances patient outcomes as well as diminishes healthcare disparities [69,70]. Particularly in monitoring bladder pressure, the incorporation of wireless communication capabilities into implantable sensors signifies a notable advancement, heightened patient involvement, remote monitoring, as well as enabling real-time data transmission [71]. Through the utilization of innovative communication protocols and hybrid solutions, researchers are leading efforts to enhance the dependability, effectiveness, as well as clinical applicability of wirelessly enabled sensors, ultimately raising the standard of care for individuals with bladder-related conditions.

3.3.3 Smart Materials and Functional Coatings.

Advances in material science have led to the exploration of smart materials and functional coatings to improve sensor biocompatibility, longevity, and performance. Nanomaterial-based coatings, such as graphene oxide and titanium dioxide, offer enhanced surface properties, including reduced inflammation response and improved tissue integration, thereby prolonging sensor lifespan within the bladder environment [72-74]. Furthermore, the development of stimuli-responsive materials holds potential for dynamic adjustment of sensor properties in response to changing physiological conditions [75,76]. This is due to the fact that these materials are mobile and can respond to internal or external stimuli including fluidic, electrical, thermal, magnetic, light and chemical.

One notable avenue of research involves the exploration of novel stimuli-responsive materials with inherent biocompatible properties or the surface modification of existing materials to enhance their compatibility with biological systems. For instance, Yu *et al.,* and Ngiejungbwen *et al.,* demonstrated a dual-layer architecture of the GO/SU-8 actuator, revealing that as ambient humidity levels rise, the GO/SU-8 driver bends towards the SU-8 side [65,66]. The exploration of solvent responsive materials such as hydrogels [68,77,78] and semi-crystalline polymers [79,80], as well as pH responsive materials such as polyacrylic acid (PAAc) in the other hand has also advanced and paved more complex deformation and actuation capability of bladder MEMS pressure sensors [81]. To address the intricate requirements of engineering applications, integrating various stimuli-responsive materials could be the logical next progression. To achieve desired effects using this method, synchronization of multiple designed and environmental stimuli is essential. Ensuring consistency in actuation throughout the design process is crucial to avoid potential inconsistencies. This necessitates meticulous mechanical modeling to ensure that each actuator stimulus exhibits similar mechanical properties.

3.3.4. Implantable Power Sources and Energy Harvesting

The exploration of novel power sources and energy harvesting strategies has been driven by the demand for implantable sensors capable of self-sustainability. Several energies harvesting technologies, including piezoelectric and triboelectric generators as well as battery-free solutions

[75,82,83] that are biocompatible, are being developed. These devices are aimed at supporting sensor's functioning for a long time without the need for on purpose power input from outside or changing batteries periodically. That could help make the device more independent and decrease malfunction risks. In future, such achievements might completely revolutionize the industry of implantable medical devices.

As can be seen from the references in the previous section in [84,85] and many other authors, battery-free implantable sensors are hopeful attempts for solving the problems of long-term reliability and autonomy. They are created using silica-based MEMS technology and do not use onboard batteries, so there are no concerns related to them being replaced, leaking or degrading. This also makes the overall risks of device malfunctions lower and improves the time of device operation. Moreover, in general, such an approach decreases the overall level of environmental outcomes of spent batteries and makes maintaining devices quite simple. Moreover, the increasing flexibility and independence of implantable MEMS bladder pressure sensors are one of the major causes of biocompatible energy harvesters' development as investigated in [84] and [86]. In this dot, triboelectric as well as piezoelectric generators are the most common methods to convert physiological motion's mechanical energy into electrical energy. It should be added that no additional surgical procedures are necessary, and a wide variety of physiological events, such as muscle contractions or fluid flow, can be converted into electricity to guarantee enough power for the implantable sensors.

In conclusion, the recent breakthrough in implantable MEMS sensors for monitoring of bladder pressure represents a progressive change in precision medicine. These advancements promise enhanced diagnostic accuracy, proactive interventions, as well as patient-tailored treatments. Through interdisciplinary cooperation and adoption of state-of-the-art technologies, researchers are primed to tackle unmet patient needs in bladder-related disorders while propelling innovation in sensor design as well as healthcare delivery.

4. Conclusions

Biomedical applications of the MEMS sensor design pose a major challenge in the field of non-invasive and cost-effective healthcare. Preliminary works focused on BioMEMS design and fabrication for the purpose of monitoring bladder pressure, but there were some obstacles to be addressed in order to commercialize it. Obstacles like appropriate pressure range and frequency response for the human bladder, as well as biocompatibility of materials and telemetric requirements are meant for the implanted sensing application. Also, some other design issues should be considered such as suitable transduction mechanisms, structural shape, and optimum geometrical design of the sensor to achieve high performance of the sensor. From this review, it was discovered that various ranges of pressure had been used to measure pressure even for the same application to measure pressure in the bladder. This requires further studies to suggest the relevant range of pressure range as the right pressure range increases the sensor's precision. Similarly, in terms of size and shape, many researchers suggested a certain size and shape of sensor but not many of them optimized the structure and size to improve the performance, particularly for sensor's sensitivity and linearity. Likewise, the consideration of frequency response of the bladder is not given serious attention in design process of the sensors although it is an important requirement in designing an implanted MEMS sensor. Many researchers proposed various types of biocompatible materials and passive telemetry wireless capability of their MEMS sensor design. From this review, it is concluded that there is still a room for improvement. While suggesting improvements in the design of a high-performance implanted MEMS sensor specifically for the application of bladder pressure monitoring, it is hoped that the descriptions given and the references to the literature therein assist researchers to discover other applications.

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