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Feasibility of Self-powering and Energy Harvesting using Cardiac Valvular Perturbations

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***Abstract*—In this paper we investigate the feasibility of harvesting energy from cardiac valvular perturbations to self-power a wireless sonomicrometry sensor. Compared to the previous studies involving piezoelectric patches or encasings attached to the cardiac or aortic surface, the proposed study explores the use of piezoelectric sutures that can be implanted in proximity to the valvular regions, where non-linear valvular perturbations could be exploited for self-powering. Using an ovine animal model, the magnitude of valvular perturbations are first measured using an array of sonomicrometry crystals implanted around the tricuspid valve. These measurements were then used to estimate the levels of electrical energy that could be harvested using a simplified piezoelectric suture model. These results were re-validated across seven different animals, before and after valvular regurgitation was induced. Our study shows that power harvested from different annular planes of the tricuspid valve (before and after regurgitation) could range from nano-watts to milli-watts, with the maximum power harvested from the leaflet plane. We believe that these results could be useful for determining optimal surgical placement of wireless and self-powered sonomicrometry sensor, which in turn could be used for investigating the pathophysiology of ischemic regurgitation (IR).**

***Index Terms*—Ultrasound, Energy Harvesting, M-scan Telemetry, Sonomicrometry, Cardiac Valvular Dynamics.**

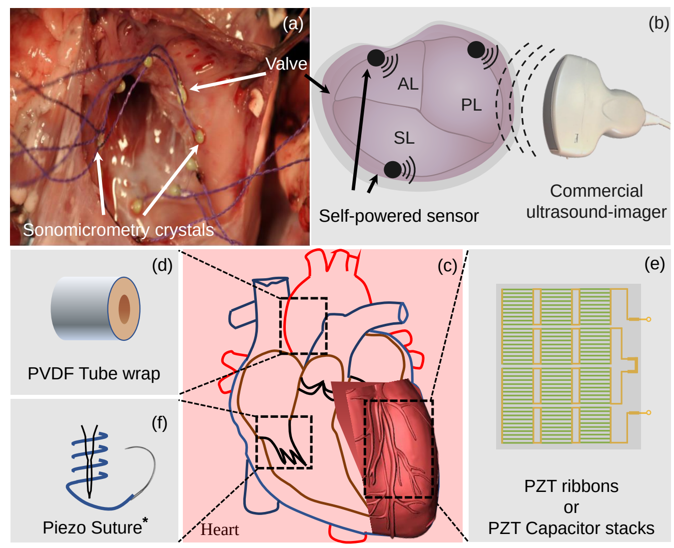

Figure 1. Vision of the proposed research for chronic monitoring of valvular dynamics and (a) overcoming the limitations of existing hard-wired sensing technology. (b) An array of wireless millimeter scale self-powered implants sutured on the valve and that can be interrogated using commercial diagnostic ultrasound scanner. (c) Anatomy of Heart and existing techniques for harvesting heart dynamics, (d) PVDF tube wraps for extracting energy from the outer annulus of the aorta, (e) PZT ribbons for harvesting from the surface of the heart which can power pacemakers (f) Proposed harvesting method to extract energy directly on the valve.

I. Introduction

A cardiac (tricuspid or mitral) valve is a complex structure with a dynamic physiology that relies on an intricate interplay between its individual components, which include the ventricles, papillary muscles, the annulus and the atrium. Failure in any one or all of these components, regardless of cause, can lead to significant mitral regurgitation (MR) and an inability to sustain normal cardiac performance. Specifically, ischemic mitral regurgitation (IMR) presents a vexing clinical problem where even a modest geometric perturbations of the valvular and sub-valvular apparatus could result in significant insufficiency [1]. Unfortunately, the pathophysiology of IMR is not well understood, and studies have suggested that the underlying cause lies in the annulus and the sub-valvular apparatus which need to be precisely tracked. However, tracking these 3D geometric perturbations presents a challenge for existing clinical imaging modalities like echocardiography or magnetic resonance imaging [2], hence requiring sensors (for example sonomicrometry crystals) to be implanted in proximity to the valvular apparatus, as shown in Fig. 1 (a). Current state-of-the-art sonomicrometry techniques [3] requires hard-wiring sensors to an external data acquisition and a power source which requires the animal to be intubated under general anesthesia. This prevents monitoring the precise three-dimensional changes in the mitral valvular complex during the evolution of IMR. For chronic and long-term monitoring, these sensors are desired to be wireless (untethered) and preferably self-powered by the cardiac activity, as shown in Fig. 1 (b).

Previous approaches of self-powering using cardiac activity have resulted in flexible piezoelectric transducers for harvesting energy from the cardiac surfaces [4]–[7], two special cases were shown in Fig. 1(d)-(e). In particular, [4], [5] proposed a flexible PZT transducer that was placed on the surface of the swine heart and was shown to harvest enough energy to power implantable devices like pacemakers. In [6], a cylindrical PVDF transducer is placed around the ascending aorta as was demonstrated for biomechanical energy harvesting. Unfortunately, these centimeter-scale surface transducers would require explicit wiring to the sensors implanted in proximity to

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the cardiac valve which could lead to surgical complications and also mechanically impede the dynamics of the valve. In this paper we investigate the feasibility of using piezoelectric sutures which are used in tethering sonomicrometry crystals to the surface of the valve for harvesting energy. The key concept here is to exploit strong valvular perturbations to excite non-linear modes in the piezoelectric suture. These non-linear or broadband modes could then be used to boost the energy harvesting capability of a piezoelectric transducer, as have been previously reported in literature [8]–[10].

The main objective of this paper is to estimate the amount of electrical power that can be locally harvested (using a model of piezoelectric suture) from non-linear geometric perturbations of different annular regions of a cardiac valve. The study is based on the following three results that have been reported in our previous works: (a) Using implanted sonomicrometry crystals (as shown in Fig. 1(a)) it is possible to reliably measure fine geometric perturbations of the mitral-valve in real-time [3], [11]; (b) The implantation of the crystals in the proximity of the valve does not affect the valvular dynamics [12]–[14]; and (c) An FDA compliant ultrasound scanner can be used to image the ultrasonic pings simultaneously generated by the implanted sonomicrometry crystals and electrical power consumed by the crystals is less than a few microwatts [15]. The results of the feasibility study reported this paper would therefore facilitate development, optimization and surgical placement of an array of self-powered sonomicrometry sensors that can be used for real-time, wireless and long-term tracking of valvular structures.

This paper is organized as follows: Section II provides an estimate on the power requirements for a wireless SM sensor that can be interrogated with a diagnostic ultrasound imager. Also it describes the methods and experiments used for collecting and validating the SM data using an ovine model. Sections III and IV introduces the model of the piezoelectric suture and describes the methods used for estimating the levels of harvested power. Section V discusses limitations of the current work and the scope in future. In Section VI, we conclude the paper.

II. Methods and Experiments

Sonomicrometry based measurement involves tracking the distances between the different spatial locations that are marked using ultrasound crystals, which are millimeter scale implants and weigh less than 20 milligrams. For instance, Fig. 1 (a) shows a typical placement of the crystals around the tricuspid valve region [16]. Each of these crystals sequentially transmit an ultrasonic pulse which is then received by the other crystals. The time-of-delay between the transmitted and received pulses is then used to estimate the respective distances between the crystals and hence the perturbation and the dynamics of the valve. In this paper, we have used the sonomicrometry preparation in live ovine models to estimate the magnitude of mechanical power that can be harvested using local valvular perturbations.

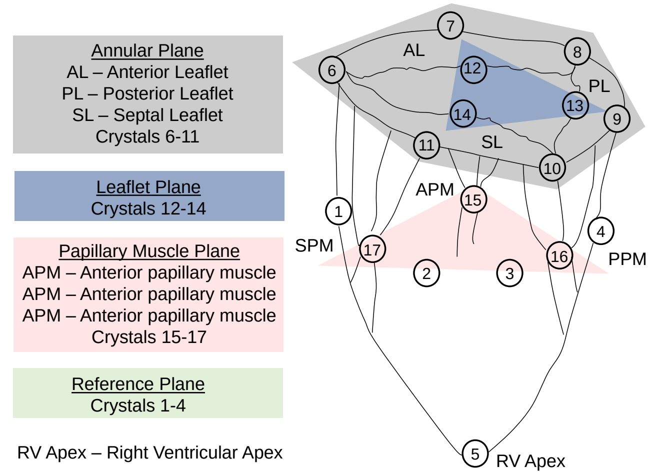

Figure 2. Geometric placement of sonomicrometry crystals and anatomy of tricuspid valve where crystals labeled (1-5), (6-11), (12-14), (15-17) were located around the epicardium, tricuspid annulus, tricuspid leaflet edges and papilliary muscle tips of the right ventricle respectively.

A. Ovine model preparation

The annular planes around the tricuspid valve are shown in Fig. 2 which also shows the placement of the sonomicrometry crystals (marked 1-17) [16]. The time-of-delay between the transmitted and received sonomicrometry pulses is then used to estimate the respective distances between the crystals and the relative mechanical strain at the marked locations. In-vivo experiments was performed on seven adult male sheep in a fully equipped and accredited animal facility at West Michigan Regional Laboratory which is a part of the Spectrum Health Delivery System. The clinically pertinent surgical procedures followed the protocols described in the published literature which supports the ovine model as the preferred animal model of human mitral valve pathophysiology.

Once the animals were sedated, intubated and under general anesthesia, a left thoracotomy was carried out in sterile fashion to expose the heart. After full heparinization, cardiopulmonary bypass was established and the heart subsequently arrested with standard crystalloid cardioplegic solution. Under cardiac standstill, the tricuspid valve was exposed through the right atrial appendage. Two sonomicrometry crystals were placed on the more dynamic posterior mitral annulus approximately 2 cm apart. Wire connections of the sonomicrometry crystals are connected to an outside data acquisition source(from SonoMetrics) through the open thoracotomy. The animals are subsequently weaned from cardiopulmonary bypass and stable hemodynamics achieved. Various physiologic conditions are introduced: 1) increased heart rate (30% above baseline) and contractility with epinephrine infusion 2) blood pressure increase (50% above baseline) with norepinephrine infusion 3) bradycardia (heart rate below 60) with esmolol infusion. Simultaneous data acquisition from crystals were performed before and during each intervention with at least a 5 minute stabilization period between interventions.

B. Data acquisition and Analysis

The sutured crystals on the tricuspid valve were clustered into four groups based on their location. As shown in Fig. 2,

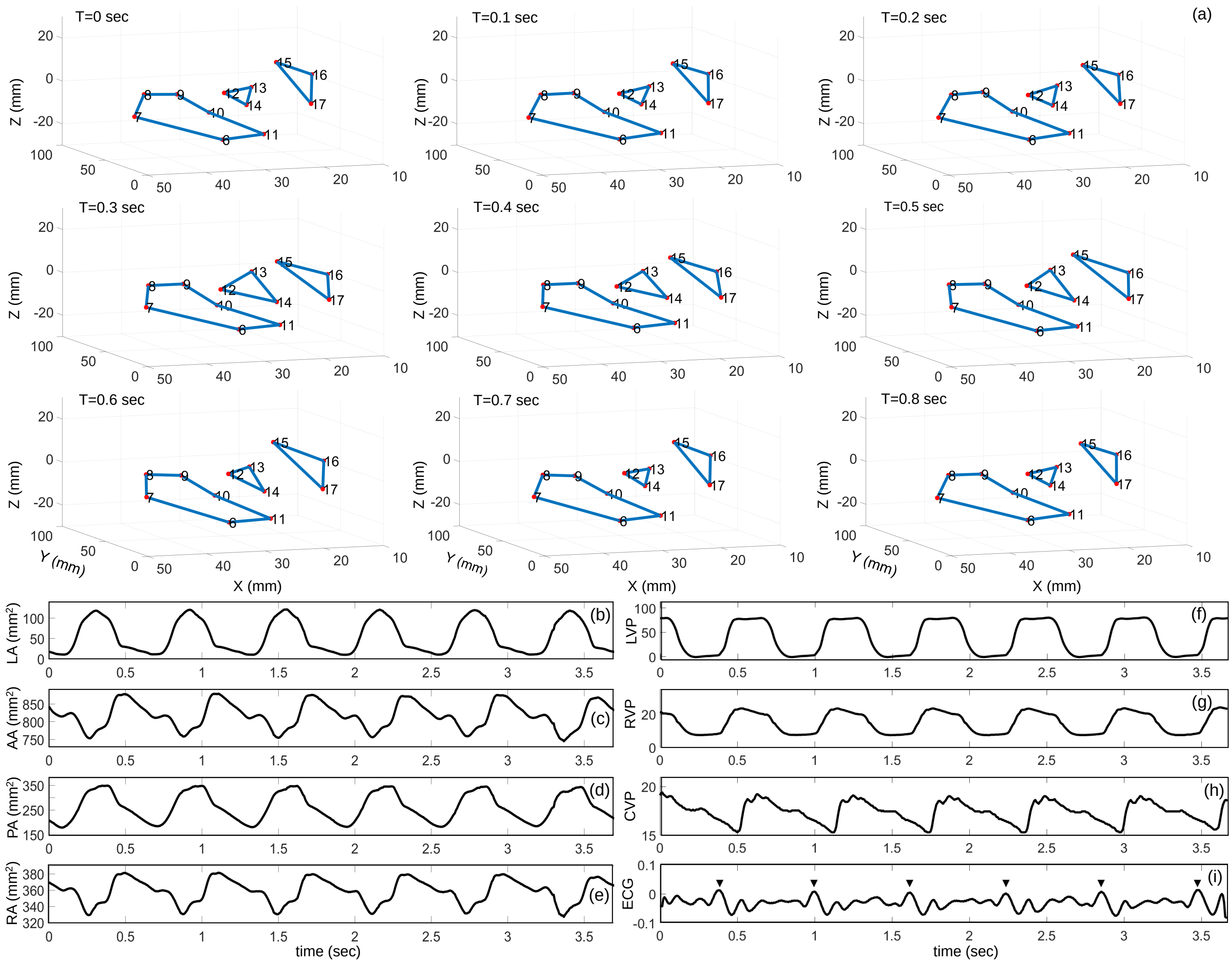

Figure 3. (a) 3D reconstruction of the valve using the raw data acquired using the sonomicrometry equipment at different instances showing the relative movement of the crystals. (b)- (e) shows the variations of the area enclosed by the polygon formed by the crystals in each plane respectively. (f) - (h) shows the pressure values LVP RVP and CVP respectively and (i) shows the processed ECG data which were all collected simultaneously and the peaks are marked manually indicating the duration of cardiac cycle.

the crystals 6-11 were placed on the tricuspid annulus, 12-14 were placed on the edges of the leaflets, 15-17 were placed on the tips of the papillary muscles while the crystals 1-5 were sutured on the epicardium of the right ventricle. The raw data from the SonoMetrics equipment gave us the recorded traces for every single crystal working both as a receiver and a transmitter. The data was recorded using 17 crystals which corresponds to 289 raw traces for every run (around 130 Hz). The data was initially processed using the software Sonosoft, provided by the manufacturer, and the 3D coordinates of the crystals at each time instant were computed assuming the location of crystal 5 as the origin. Post simulations were carried in MATLAB using a custom data analysis software.

Fig. 3(a) shows an experimentally determined three-dimensional map of the annular regions of an ovine tricuspid valve. X, Y and Z refers to the three co-ordinates of the crystals measured in millimeters (mm). Each of the crystals labeled 12-14 in Fig.3(a) track the relative size of the leaflet plane during each opening and the closing of the valve. Other crystals in Fig. 3(a) are used for tracking auxiliary regions (annular edges and papillary muscle tips) in the proximity of the valve. In our previous studies [16] we have shown that the sonomicrometry data could be also used for estimating changes in clinically relevant valvular parameters, for example the changes in subvalvular distances and the annular area, which could be useful for clinical diagnostic purposes. To validate the data obtained using our experimental setup we first compute simple geometric area to verify the data with the well accepted ECG and pressure recordings which were also collected simultaneously. Fig. 3 (b)-(e) show the estimated change in the planar area corresponding to different tricuspid valvular regions based on the raw data collected from an ovine model (LA, AA and PA refers to the area computed in leaflet plane, annular plane and papillary muscle plane respectively). Fig. 3 (f)-(h) show the plots for left-ventricular pressure (LVP), right-atrial pressure (RVP) and central venous pressure (CVP)

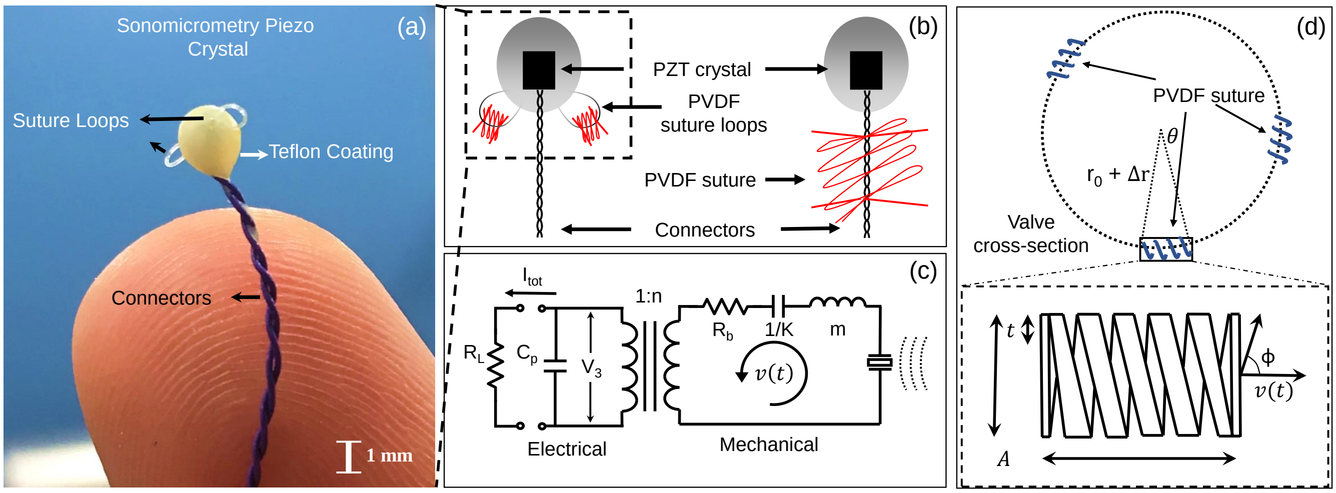

Figure 4. (a) Sonomicrometry crystal and (b) mechanism for harvesting energy either by replacing the suture loops or by using a piezoelectric suture to attach the crystal to the valve. (c) Suture model and parameters used in the analysis. (d) Cross-section of a valve assuming circular contour and sutured crystals tied to the boundary. At the bottom shows the model a multi-loop piezoelectric suture which reduces the effective local strain level $d\delta$ by a loop factor N.

for the same duration. Also Fig. 3 (i) shows the filtered ECG data. The raw data of ECG collected using an ADC is processed by a Savitzky-Golay least-square polynomial filter [17] to extract the epochs which indicate the cardiac cycle. The results validate the accuracy and the potential of SM measurements in studying the valve dynamics.

The measured geometric data was found to be consistent with published reports from other groups, [18], [19] solidifying the reliability and validity of wired sonomicrometry technology. Another observation that can be inferred from the wired sonomicrometry experiments is that the rate of change in mechanical strain (change in dimension normalized by the original dimension) due to valvular dynamics is in the order of 0.5 ϵ/s. This is also consistent with strain and strain-rate measurements obtained using cardiac ultrasound [20]. This magnitude of strain-variations could potentially be exploited to harvest microwatts of energy using a millimeter-scale piezoelectric transducer.

III. Piezoelectric Suture Model and Energy Estimation

Our modeling study is based on two techniques for integrating piezoelectric transducers to the sonomicrometry crystal, as shown in Fig. 4(a). Either the crystals could be tethered directly to the valve annulus by means of a piezoelectric suture, or the crystal could be first integrated by a piezoelectric suture loop which could then be tethered to the valve using a standard surgical suture, as shown in Fig. 4 (b). Connecting piezoelectric loops to the crystal could be the easiest way for designing an integrated implant but it might be less efficient in energy harvesting because of its poor mechanical coupling (contact) to the tissue. Whereas using piezoelectric suture could provide good mechanical coupling with the tissue, however, fabricating a flexible suture might present a challenge. In our estimation study we abstract these two possible configurations while noting that future research would be needed to choose between the two options.

A. Power estimation procedure

For our energy estimation model, we approximated the shape of the valve to be circular, with each crystal located along the perimeter of the circle, as illustrated in Fig. 4(d). Ideally suture can be assumed as a helical loop with a limited number of turns N, but in our analysis we approximated the shape to a rectangular strip with thickness t and cross-section area A. Note that the use of a helical loop with N turns (as shown in Fig. 4(d), effectively reduces the local strain level by N and ensures that the piezoelectric suture can operate within the material fracture compliance limits. We model the radius of the circular contour as a function of time and thus it can be considered as $r = r_0 + \Delta r(t)$ where r_0 is the mean radius during a cardiac cycle. First, a normal vector to the plane passing through clustered set is estimated based on the minimum mean squared error (MMSE). These coordinates were then rotated and translated so that the crystals were located on the XY plane with the center at origin. The radius r of the circle passing through the points then is estimated based on the MMSE. The rate of change in radius $v(t)$ at time instant t at each cross-sectional plane is estimated using a first order approximation as

$$\begin{aligned} v(t) &= \frac{d}{dt}(r(t)) \\ v(n) &= \frac{r(n) - r(n-1)}{\Delta t} \end{aligned} \tag{1}$$

where the radius $r(n)$ is estimated for every cycle and the sampling time-interval Δt is chosen to be around 75 ms. The velocity profile for each cross-sectional plane is shown in Fig. 5 (a) which is the average value across the data recorded from seven ovine models. Although the direction of the strain in the suture would be along ϕ, as shown in Fig. 4 (d), in our current analysis we consider $v(t)$ to be the estimate of the strain-rate of the suture since our objective is to estimate the maximum power levels.

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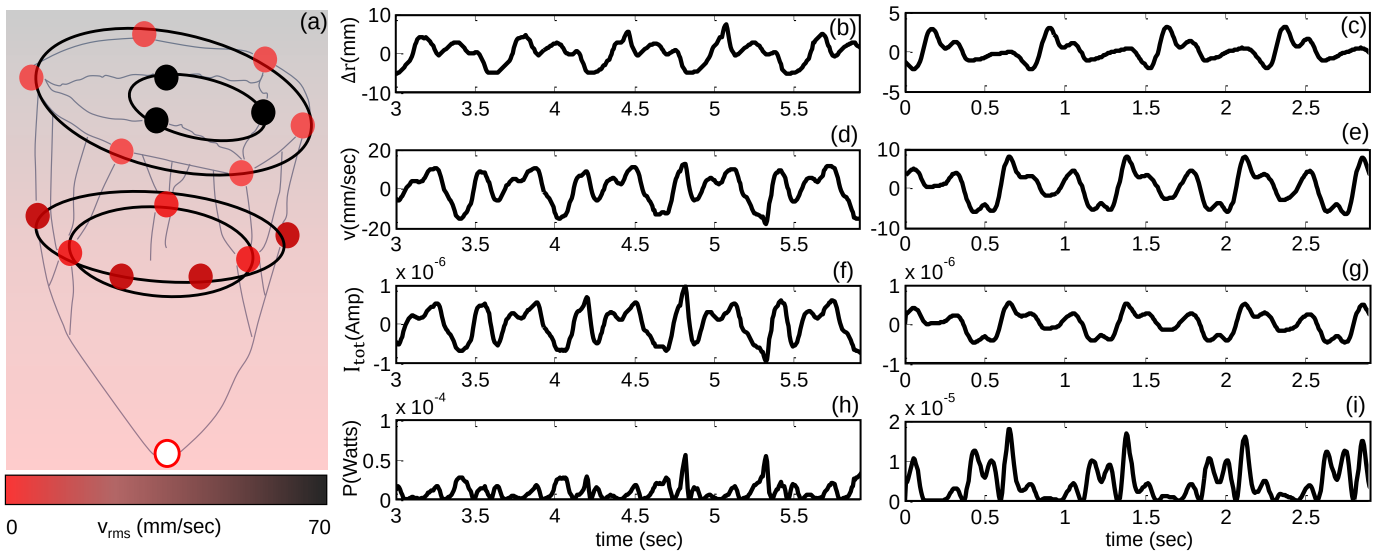

Figure 5. (a) Average velocity profile at each cross section which is obtained by analyzing the data recordings from seven ovine models (black indicating the maximum and red corresponding to zero velocity). (b)-(i) Estimated change in radius, velocity, current and power on the leaflet edges based on the data collected before and after inducing tricuspid-regurgitation (TR).

Table I
PIEZOELECTRIC PROPERTIES AND PARAMETERS USED IN THE SUTURE MODEL.

Parameter	Symbol	Value (PVDF)	Value (PZT)
Strain coefficient	d_{31} (m/V)	23×10^{-12}	110×10^{-12}
Dielectric constant	ϵ_{33} (F/m)	11×10^{-11}	198×10^{-11}
Young's modulus	Y_{11} (N/m^2)	3×10^9	6×10^{10}
Transducer thickness	t_3	10 μm	10 μm
Electrodes area	A	10 mm^2	10 mm^2
Load resistance	R_L	60 $M\Omega$	60 $M\Omega$

Assuming the two electrodes of the piezo are electrically neutral, the total charge (Q_{tot}) generated on each side of the electrode due to deformation is given by,

$$Q_{tot} = \int D_3 dA \quad (2)$$

where, D_3 is the electrical displacement along the polarization direction.

D_3 is determined by the amount of stress (T_1) and the electric field generated across the electrodes and is given by,

$$\begin{aligned} D_3 &= -d_{31}T_1 + \epsilon_{33}E_3 \\ &= -d_{31}Y_{11}\eta_1 + \epsilon_{33}\frac{V_3}{t_3} \end{aligned} \quad (3)$$

Equation. 3 is the general constitutive equation for the piezo-electric materials, where the parameters of the material are summarized in Table I.

Defining the strain to be the ratio between relative change in radius($\Delta r = r - r_0$) and initial radius (r_0), the rate of change in charge is given by,

$$\begin{aligned} Q_{tot} &= -\frac{d_{31}Y_{11}A}{r}\Delta r + \epsilon_{33}\frac{A}{t_3}V_3 \\ \frac{dQ_{tot}}{dt} &= -\frac{d_{31}Y_{11}A}{r}\frac{dr}{dt} + \epsilon_{33}\frac{A}{t_3}\frac{dV_3}{dt} = I_{tot}. \end{aligned} \quad (4)$$

and A is the area of suture strip. Based on the equivalent circuit model of the piezoelectric transducer as shown in Fig. 4 (c) the voltage (V_3) and current I_{tot} flowing through the load resistor is given by,

$$\begin{aligned} \frac{dV_3}{dt} + \frac{V_3}{RC_p} - \frac{d_{31}Y_{11}A}{C_p r}\frac{dr}{dt} &= 0 \\ \frac{dV_3(t)}{dt} + \frac{V_3(t)}{RC_p} - \frac{d_{31}Y_{11}A}{C_p}\frac{v(t)}{r(t)} &= 0 \end{aligned} \quad (5)$$

where $C_p = \frac{\epsilon_{33}A}{t_3}$ is the intrinsic capacitance as shown in Fig. 4 (c). Note that this equation is consistent with the earlier reported works [4]. The differential equation can be discretized for simulation as

$$V_3(n+1) = V_3(n) - \Delta t \times (\frac{V_3(n)}{RC_p} - \frac{d_{31}Y_{11}A}{C_p}\frac{v(n)}{r(n)}) \quad (6)$$

and the instantaneous power (P) across a resistor load of R can be estimated as

$$P(n) = \frac{V_3(n)^2}{R} \quad (7)$$

It can be seen that the voltage generated across the transducer is a function of the history of the rate of change in strain, as shown in equation. 6, which is directly proportional to instantaneous rate ($v(t)$) and inversely proportional to radius ($r(t)$). Table. I shows typical values of a piezo material which were reported in literature [21] and used here in our simulations and analysis.

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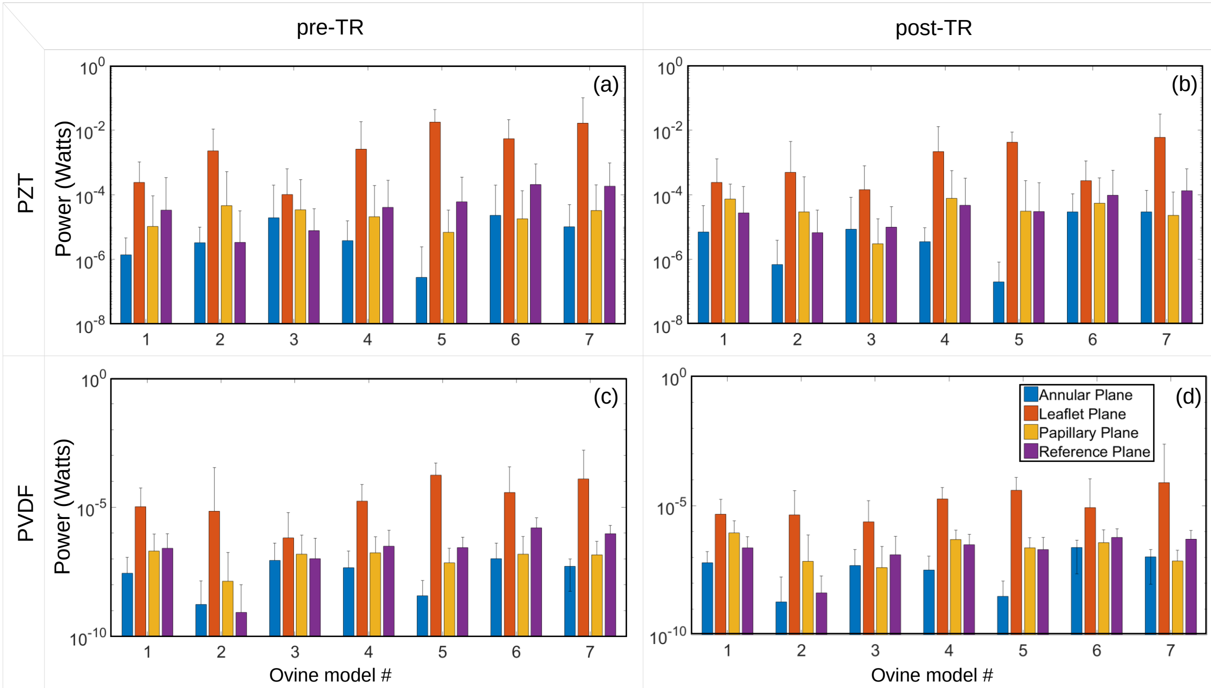

Figure 6. (a)-(b) shows the variations in power values estimated on each contour across the seven datasets (ovine sheep models) before and after inducing TR respectively (assuming PZT as the material of suture). Similarly (c)-(d) shows the variations for the case of PVDF suture.

Table II
MAXIMUM AND AVERAGE ESTIMATED POWER THAT CAN BE HARVESTED AT DIFFERENT ANNULAR PLANES, UNDER DIFFERENT CONDITIONS

Condition; piezo material	**pre-TR; PZT**		**post-TR; PZT**		**pre-TR; PVDF**		**post-TR; PVDF**	
	Max. Power	Avg. Power	Max. Power	Avg. Power	Max. Power	Avg. Power	Max. Power	Avg. Power
Annular plane (6-11)	46 μ W	6.5 μW	30.3 μW	3.8 μW	0.21 μ W	39.3 nW	0.1 μW	17.8 nW
Leaflet plane (12-14)	**21 mW**	**2 mW**	**5 mW**	**0.6 mW**	**0.24 mW**	**11 μW**	**0.19 mW**	**8.9 μW**
Papillary Muscle plane (15-17)	0.14 mW	22.1 μW	0.1 mW	15.4 μW	0.84 μ W	0.17 μW	0.6 μW	0.1 μW
Reference plane (1-4)	0.23 mW	27.8 μW	0.1 mW	17.2 μW	0.78 μW	0.13 μW	0.38 μW	61.5 nW

IV. RESULTS

Based on the radius estimation procedure, different approximations of the cross-sectional area were made and the rate of change in the radius was estimated using the equation 1. Fig. 5 (b),(c) shows the variations in the change in radius at the leaflet plane with respect to time for a particular experiment case before and after inducing tricuspid-regurgitation (TR) respectively. Fig. 5 (d),(e) shows the instantaneous velocity values for either cases based on the method mentioned in section III-A. The power that can be harvested from the suture is estimated using the equation 7, for two specific cases: (a) before TR is induced (labeled as pre-TR); and (b) after TR is induced (labeled as post-TR). The results for both these cases are shown in Fig. 5(f)-(g). The results show that even though the change in the radii is periodic there is a large difference in power estimated at each cycle, as shown in Fig. 5(f). This can be attributed to the natural deviations in the ovine cardiac function and intrinsic non-linear dynamics of valvular perturbations. These non-linear effects result in energy bursts which can be readily harvested for self-powering. Note that for surface piezoelectric transducers, like the ones reported in [4], the non-linear dynamics (like buckling) gets filtered out resulting in smooth and periodic strain-variations. Therefore, transducers with large surface area have to be used to harvest sufficient energy.

Fig. 6 shows the bar plots corresponding to the estimated power that can be harvested from each annular plane, for different animals and for pre and post TR conditions. We

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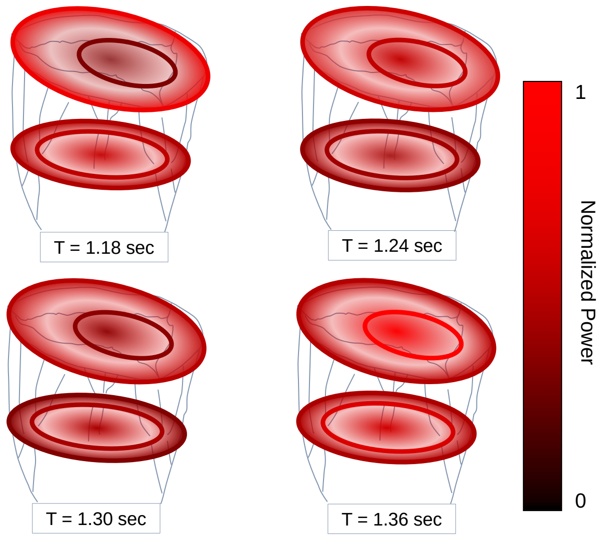

Figure 7. Power profile across the tricuspid valve showing the variations, in the normalized power, with respect to time. The power levels change rapidly with crystals located at different locations harvesting maximum power at different time-instants.

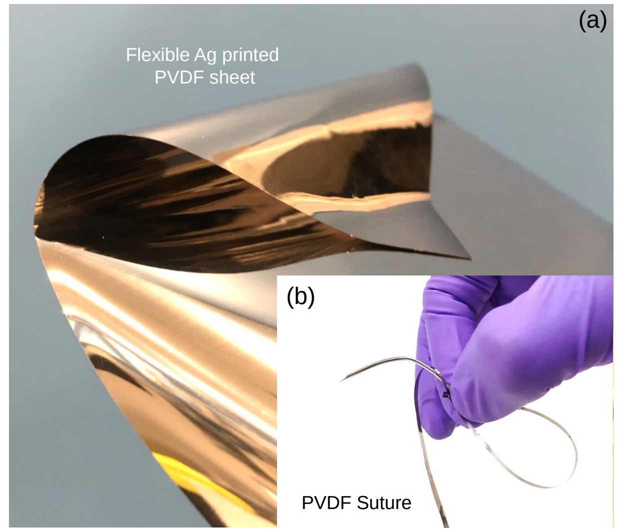

Figure 8. Flexible silver-coated PVDF sheet as a potential candidate for constructing piezoelectric sutures.

estimated the power levels assuming two types of suture models incorporating two different piezoelectric materials. The material properties and parameters used in our simulations were summarized in Table. I. Note that while PZT sutures could potentially provide more power, they need to be properly encapsulated to ensure biocompatibility. The Fig. 6 shows that the estimated power values are consistent across different animals showing the scalability of the proposed approach. Also, the instantaneous voltage and current values range between 0-20 V and 0-1 μA respectively. In all cases, the maximum amount of power can be harvested from the leaflet plane. Although there is a slight decrease in power for the case of pre-TR and post-TR, as shown in Fig. 6(a)-(d), it can be seen that the power levels across different cross-sections in all the cases are almost similar. This implies that self-powering is still feasible even after TR is induced. Fig. 7 shows the power profile for a particular case of ovine model (pre-TR, using PVDF suture model) where the variations of the normalized estimated power at different instances in time. It can be seen that variations of power level are rapid (the time-scale of the change is in order of 60 msec) when compared to that of the change in radius or velocity. Also, the distribution of power level show bursts where different sections of the valvular planes deliver maximum power at different time instants. Thus, placement of the sutures and hence the sonomicrometry sensors is important to ensure efficient and consistent levels of harvested power.

Although the analysis presented in this paper are based on the simulations of the suture models, earlier studies have shown a good match between the simulation studies and experimental results [4]. Thus, the average power that can be harvested, as summarized in Table. II should scale. Also, note that model assumed the piezoelectric suture to be operating in the transverse mode, however, in practical implementation we need to consider the longitudinal and shear effects. Therefore, the numbers summarized in Table. II represent the upper limit and needs to be moderated based on practical suturing conditions. Also the power density is around $10mW/cm^2$ in case of PZT and $100\mu W/cm^2$ in case of PVDF for a piezoelectric suture.

V. Discussion and Future work

The feasibility analysis presented in this paper is based on ideal assumptions on the suture orientation and its material properties, and therefore should be considered as the upper-limit. While the results presented in this paper have been estimated using suture parameters summarized in Table I the results should scale with any variations in these parameters. It can be seen from equation. 6 that with the values of voltage scale linearly with the change in area where as the power levels vary quadratically, however, corner effects need to be considered when translating these results. Also, a suture with dimensions 0.5 mm x 2 cm might be reasonable for simulations but practical feasibility of such a configuration needs to be further explored.

In practice, however, the results may differ and the non-linear valvular dynamics might boost or damp the level of harvested energy. In particular, mechanical properties like buckling and torsional strain in case of helical suture, which is not captured by the linear model given in equation. 5, could enhance the results where as the properties like damping and mechanical coupling can deteriorate the performance of the transducer needs to be validated with experimental results. Future work will also include verifying an integrated sonomicrometry suture in-vivo. Fig. 8 shows one possible configuration to construct the suture cut out of a PVDF sheet. The mechanical stability, bio-compatibility and packaging of the suture is a topic of future research. Also, future work will involve integrating piezoelectric energy harvesting circuits [22] and sonomicrometery telemetry circuits [23] with the piezo-electric suture.

VI. Conclusion

In this paper, we estimated the levels of energy energy that can be harvested from non-linear mechanical perturbations of a cardiac valvular appartus. Our analysis demonstrate the feability of of harvesting 0.1-10mW/cm^2 which is orders of magnitude higher than the previous configurations that harvested energy from the surface of the heart or the aorta. For instance, the time average power density was reported to be $1.2\mu W/cm^2$ when harvesting energy from the surface of the heart [4], where as the power density was $170nW/cm^3$ when the energy was harvested from the aorta [6]. Experimental verification of this improvement would be the subject of future work. However, higher levels of harvested power opens the possibility of designing self-powered sonomicrometry based sensors that can be used for chronic monitoring of mitral-valve dynamics and post-surgical IMR.

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