Cardiolock2: Parallel singularities for the design of an active heart stabilizer

Wael Bachta, Pierre Renaud, Edouard Laroche, Jacques Gangloff

Abstract—In this paper, the design of a new active cardiac stabilizer, Cardiolock2, is presented. Following the proof of concept Cardiolock [7], this device allows an active stabilization of the surface of a beating heart in two directions, which is considered sufficient for a complete stabilization. Piezoelectric actuation is combined with a compliant architecture to obtain high dynamics and accuracy. A remote center of motion is obtained with a serial architecture, and parallel mechanisms in configurations close to singularity are used to get a sufficient workspace. A kinematic analysis is first presented, before detailing the main properties of the device and the current development of the prototype.

I. INTRODUCTION

In the field of heart surgery, coronary artery bypass grafting (CABG) is one of the most widespread techniques. The surgeon has to suture arteries of 1-2 mm of diameter on the surface of the heart, using a suturing thread of 70 µm of diameter. The needed accuracy has first been obtained by stopping the heart, using a heart-lung machine, and achieving a sternotomy to get an easy access to the heart. The associated harmful effects [1] lead the surgeons to consider an operation on a beating heart, using a stabilizer. An even more challenging approach is to achieve the CABG on a beating heart in a minimally invasive surgery (MIS) context, i.e. by only introducing instruments through small incisions equipped with trocars.

Stabilizers are used to locally immobilize the beating heart surface, typically 5 × 5 mm². The lack of accuracy of these mechanical systems has been proved, when a sternotomy is performed [2], [3], and also in a MIS context [4]. In this paper, we consider this latter case: stabilizing the surface of a beating heart with a device compatible with MIS. The task is not easy, because of the heart dynamics, with an acceleration of the free surface in the order of 1 g [22], and because of the large forces that can be exerted by the heart. Using physiological similarity models, as proposed in [5], and in vivo experimental data acquired on pigs [6], we can expect the force of the human heart on the stabilizer to be approximately equal to 7 N, taking into account respiratory and cardiac motions.

In a MIS context, a cardiac stabilizer is introduced through a subxiphoid incision using a trocar. The stabilizer has a length of 250 mm for a correct access to the area of interest. The diameter of the device is limited by the trocar size, and is typically in the order of 10 mm. Using simple strength-of-materials considerations, the displacement of the end-effector due to the cardiac force is larger than the targeted accuracy, in the order of 0.1 mm. The heart forces also create significant torques on the base of the stabilizer that can cause additional deflections. A passive mechanical approach seems consequently of limited efficiency in such a context. We have therefore proposed [7] the concept of active heart stabilization by associating an active device called Cardiolock (Fig. 1), using piezoelectric actuation and a compliant structure, and high speed vision feedback to evaluate the end-effector displacements. The actuator is controlled to suppress the effect of the cardiac force on the stabilizer using the visual feedback.

The analysis of the heart forces conducted in [7] has demonstrated that the beam of the stabilizer is subjected to a force vector whose components are of the same order. Only the forces perpendicular to the beam induce significant beam deflections, and torques on the stabilizer base. To obtain a full compensation, we need therefore to provide a device with two degrees of freedom, whereas Cardiolock allows a compensation in a single direction.

Cardiolock does not exhibit a Remote Center of Motion (RCM), that would correspond to the incision point where a trocar is inserted. This is not fundamentally contradictory with a MIS use, since the device displacements are in the order of 1 mm at the end-effector location, and lower at the trocar position. However, the insertion point has a quite significant displacement, in the order of 5 mm in the direction perpendicular to the beam [8]. If we want to simply keep the base of the stabilizer in a static position during stabilization, to avoid the use of an additional actuated mechanism, the 5 mm displacement will constitute a perturbation on the device behavior, since it will create additional torques on
the device actuated joint. This can increase the control complexity and lower the performance of the device. A simple solution is to provide the active stabilizer with a RCM so that no work is developed by the force exerted by the trocar on the device during the compensation.

As a consequence, we propose in this paper the design of Cardiolock2, the second generation of our active stabilizer. It enables a compensation in two directions, with a RCM mechanism. To do so, parallel singularities are considered to amplify the actuator displacements. In the second section, the general architecture of the device is first presented, with the kinematic decomposition and the proposal of the use of singularities to control the amplification of the actuator displacement. In the third section, the main properties of the device are given, with an emphasis on the design constraints. In the fourth section, the current development of the prototype is detailed, before concluding.

II. STABILIZER KINEMATICS

A. Selection of an architecture

Similarly to Cardiolock, we consider an architecture based on the use of compliant joints [9] to avoid the presence of backlash and friction. In this case, the Pseudo Rigid Body Model (PRBM) [10] approach enables to use a rigid-body kinematics analysis.

1) Kinematic constraints: The device must provide two end-effector degrees of freedom, in directions perpendicular to the stabilizer beam. It must also have a RCM to limit the influence of the trocar forces. Since the end-effector displacements needed to compensate for deflections are in the order of 1-2 millimeters, the 250 mm distance between the RCM and the end-effector enables us to approximate the needed device displacements by two rotations with respect to the RCM. The device must therefore be a spherical two degrees-of-freedom (DOF) mechanism with a RCM, in a serial or parallel arrangement.

2) Serial or parallel architecture?: Several parallel mechanisms with a RCM have been proposed in the literature. Sphericity can be obtained by using several legs, each having a spherical movement [11]. For some other mechanisms, the property is obtained by the arrangement of the joints [12], [13]. For the cited architectures, the center-of-motion could coincide with the trocar. For all these architectures, revolute and sometimes spherical joints have however to be manufactured with a spatial arrangement. The manufacturing complexity is thus very high. When sphericity is due to special conditions on the joint positions, manufacturing errors and flexibilities can furthermore cause the loss of sphericity. In that case, it is difficult to establish which displacements are finally controlled.

Additionally, experimental evaluation of the Cardiolock device has shown that some significant differences between the device dynamic behavior and its estimation from Finite Element Analysis (FEA) come from the influence of linkage assembly. The choice has therefore been made to minimize the complexity of the architecture assembly and promote the simplicity by considering a 2 DOF serial architecture.

3) Proposition: Since the rotations of the stabilizer beam are of small amplitude, it is interesting to consider the special arrangement represented in Fig. 2, that allows us to get a decoupled behavior in the represented nominal configuration. In Fig. 2, one can see indeed that the first revolute joint $J_1$ axis lies with the beam axis in a horizontal plane, and provides an end-effector velocity in the $x_{ee}$ direction. The second joint $J_2$ axis lies with the beam in a vertical plane, so that the $y_{ee}$ direction velocity is obtained. This decoupling will simplify the dynamic control of the system.

The Jacobian matrix that relates the joint velocities $(\dot{\theta}_1, \dot{\theta}_2)^T$ to the end-effector velocities $(\dot{x}, \dot{y})$ is thus only a function of the angles $\alpha_1$ and $\alpha_2$, since the beam length $L$ is constrained by medical requirements:

$$
\begin{pmatrix}
\dot{x} \\
\dot{y}
\end{pmatrix} = 
\begin{pmatrix}
L \sin(\alpha_1) & 0 \\
0 & L \sin(\alpha_2)
\end{pmatrix}
\begin{pmatrix}
\dot{\theta}_1 \\
\dot{\theta}_2
\end{pmatrix}
$$

(1)

The Jacobian matrix provides also the relationship between small revolute joint rotations and end-effector displacements. For a given set of end-effector displacements, the actuated joints rotations decrease when the angles $\alpha_1$ and $\alpha_2$ are increased.

From a dynamical point-of-view, the parameters $\alpha_1$ and $\alpha_2$ should be minimized in order to obtain a more compact structure, with lower inertias. The values of the two parameters must however be chosen by considering also the actuation of each joint since the piezo actuators which are to be integrated to control the revolute joints have a limited stroke.

B. Singularities for amplification

In this section, the actuation of the two revolute joints $J_1$ and $J_2$ is considered. As explained before, we adopt a totally compliant structure. Therefore stack piezo actuators have been chosen: this technology does not introduce any backlash or friction. Since this type of actuators delivers a linear motion, an actuation stage has to be designed in order to transform this motion in a rotation of the joints of the serial spherical architecture. The actuation stage must provide a high ratio rotation/translation to lower the angles $\alpha_1$ and $\alpha_2$ introduced in section II-A.3.

A simple slider-crank mechanism enables such a transformation, and was used for Cardiolock (Fig. 3). In such a configuration, the ratio rotation/translation is linked to the length $\lambda$ (Fig. 3). If we want to increase the rotation for a constant actuator translation, the parameter $\lambda$ has to be lowered. However, for small values, the mechanism stiffness becomes limited, if we consider for instance a force in the Y direction.

Indeed, some closed-loop mechanisms, or more generally parallel architectures can provide at the same time a stiff structure, to minimize any uncontrollable displacements, and a large rotation from the displacement provided by the piezo actuator. For a parallel mechanism, the actuator velocities $\dot{q}$ and the end-effector velocity $\dot{X}$ are linked by the relationship:
plane, and equiprojectivity properties: its orientation \( \theta \) end-effector pose is defined by the position \( u \) prismatic joints. Their directions are defined by unit vectors. More particularly, we can consider a 3P selected to obtain the desired out-of-plane stiffnesses. And think to a planar parallel structure, whose thickness can be equivalent stiffness in all the directions, we can immediately actuate and end-effector velocities.

Out-of-plane stiffness is controlled by the width of the platform is an equilateral triangle, for symmetry reasons. The rotation \( \epsilon \) only avoids to be completely in singularity. The platform is an equilateral triangle, for symmetry reasons.

The points \( A_i \), \( i \in [1,3] \), are controlled by actuated prismatic joints. Their directions are defined by unit vectors \( u_i \), \( i \in [1,3] \), and \( q_i \) denotes the prismatic joint position. The end-effector pose is defined by the position \( (x, y) \) of \( E \) and its orientation \( \theta \). For such a mechanism, the expression of \( J_X \) and \( J_q \) (Eq. 2) can be easily derived, using for instance equiprojectivity properties:

\[
J_q \dot{q} = J_X \dot{X}
\]  

(2)

where \( J_X \) is rank deficient in a parallel singularity [14]. That means \( \dot{X} \neq 0 \) can be obtained with \( \dot{q} = 0 \) and, in the vicinity of that singularity, one can tend to increase the ratio between actuators and end-effector velocities.

Since we need the equivalent of a revolute joint, with equivalent stiffness in all the directions, we can immediately think to a planar parallel structure, whose thickness can be selected to obtain the desired out-of-plane stiffnesses. And more particularly, we can consider a 3PRR in a configuration close to singularity (Fig. 4). In the represented configuration, the rotation \( \epsilon \) only avoids to be completely in singularity. The platform is an equilateral triangle, for symmetry reasons.

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\[
J_X = 
\begin{pmatrix}
A_1 B_1 |_x & A_1 B_1 |_y & (A_1 B_1 \times B_1 E) |_z \\
A_2 B_2 |_x & A_2 B_2 |_y & (A_2 B_2 \times B_2 E) |_z \\
A_3 B_3 |_x & A_3 B_3 |_y & (A_3 B_3 \times B_3 E) |_z
\end{pmatrix}
\]  

(3)

and

\[
J_q = 
\begin{pmatrix}
u_1 \cdot A_1 B_1 & 0 & 0 \\
0 & u_2 \cdot A_2 B_2 & 0 \\
0 & 0 & u_3 \cdot A_3 B_3
\end{pmatrix}
\]  

(4)

with \( z \) the unit vector perpendicular to the mechanism plane, \( (\dot{x}, \dot{y}, \dot{\theta})^T \) the vector of operational velocities.
mechanism, considering a compliant mechanism.

Finally we can therefore consider Cardiolock2 as a serial spherical architecture, each actuated revolute joint being obtained by means of a planar parallel structure in a configuration close to parallel singularity. The parallel structures are controlled with piezo actuators, and designed as compliant mechanisms. For such mechanisms, the Pseudo Rigid Body Model (PRBM) [10] approach enables to use the previously introduced kinematic analysis. The synthesis of the planar structure must however be achieved by taking into account the material elasticity in the joints. This is achieved in the next section.

The use of singularities has been proposed a few times for the design of robotic systems [15]–[17]. It is interesting to note that a very close problem, which deals with the amplification of the linear displacement provided by a piezo actuator, has been at the same time widely studied in the field of actuator design. The proposed mechanical amplifiers [19]–[21], are then designed using continuum mechanics or sometimes PRBM approach. In the latter case, rigid body kinematics are used to investigate architectures that could create amplifiers. The finally proposed solutions are based on the use of parallel singularity, by instance for bridge-type mechanisms [21], but without explicit identification of the phenomenon.

III. DESIGN OF CARDIOLOCK2

A. Description

The device (Fig. 5, 6 and 7) consists in one actuated subsystem and a distal beam, which is the only element that needs to be sterilized. The asepsy of the actuated subsystem can be simply obtained by wrapping it in a sterile bag. The distal beam is a stainless steel tube, so that it is possible to provide a suction capability by inserting aspiration tubes inside the beam.

It is worth noticing that the use of a tubular beam increases the beam deflection by comparison with a plain cylinder, since its quadratic moment is 24% lower for the selected geometry wrt the beam of Cardiolock. But in the same time, it allows 21% higher beam eigenmodes.

The actuated subsystem is based on the architecture represented in Fig. 2, obtained using two parallel mechanisms given in Fig. 4. The two parallel platforms are identical, and made from 7075T6 aluminum alloy. For sake of compactness, the piezo actuators, Cedrat Technologies APA120ML, are not located in the plane of the parallel structures but at their back. Such a position in the back of the parallel structures leads to an additional bending moment on the parallel structure. To limit the induced stress increase, a compliant prismatic joint has been added (Fig. 6).

The selected piezo actuators are components with integrated mechanical amplifiers. Their maximum stroke when not constrained is equal to 130 microns, and the maximum blocked force is equal to 1400N. These actuators are of the same type as the one used for Cardiolock. The performance obtained with such actuators is discussed in section 3.3. We present first the main elements taken into account to select the design of the parallel platforms.

B. Parallel structures design

As mentioned in section 2, the geometric parameters describing the parallel structures have to be chosen simultaneously with the parameters linked to the serial spherical mechanism.

The selection of the parameters has been achieved in an iterative way, using analytical results from the kinematic analysis and numerical results from FEA. In this paper, we
only present the main properties of the selected design. The value of the main geometric parameters is given in Table I.

From FEA, it is possible to determine the stiffness of the equivalent spring acting against each piezo actuator. This stiffness governs the maximum displacement that can be obtained with the actuator, due to the piezoelectric material elasticity. In our case, the piezo actuator can achieve a 100 microns excursion. This allows end-effector displacements of 1.5 mm in the two directions. The heart forces induce end-effector displacements of 0.8 mm. We can therefore compensate for these deformations, and possible additional flexibilities of the structure which supports the device.

The compliant joints are based on circular flexure joints, but with a modified geometry (Fig. 6), composed of circular arcs, that enables a more compact design, while obtaining a very close behavior. Stresses in the joints are only increased by 11% by the geometry modification. The minimum thickness of the joints is 0.5 mm, a value chosen to allow CNC machining.

The maximum stresses in the joints have been evaluated using FEA. The Von Mises equivalent stress is in the order of 240 MPa, when the maximum displacement of the piezo is reached and the heart forces are applied on the device. Due to the periodic nature of the load, with a frequency equal to the heartbeat, the device has been checked considering the fatigue of the compliant joints. Using a Haigh diagram [18] of 7075T6, the device is expected to have around 280 hours of lifetime. Since the platforms can be obtained by CNC machining, one can think of regular standard replacement to go beyond that duration. An alternate way could be to use in the future a material with a higher fatigue strength/elasticity modulus ratio, such as titanium alloy Ta6V, which is also machinable.

It has been confirmed by simulation that the prismatic joint significantly lowers the maximum stress in the joints, by limiting the influence of the bending moment induced by the actuator in the compliant joints of the parallel structure. Furthermore, it avoids to suppose that the piezo actuator is equivalent to a prismatic joint, which is delicate due to the small connecting surfaces that could allow non desired displacements.

C. Dynamic behavior

Two aspects have been considered for the analysis of the dynamic behavior of the device. First of all, eigenmodes of the structure have been numerically evaluated. The whole structure has been considered, including the actuators. The first two eigenmodes have been identified at 57Hz and 75Hz, which is compatible with the dynamics of the application. The first mode corresponds to the resonance of the first actuator with the associated parallel mechanism. The second mode is a structural mode that does not affect the actuators: it corresponds to the resonance of the stabilizer beam and the torsion of the end-effector of the second parallel structure.

In a second step, a dynamic model is derived to evaluate the maximum accelerations that can be obtained with the device. The force \( F_p \) delivered by a piezo actuator is linked to its displacement \( \delta u_p \):

\[
F_p = NV - K_p \delta u_p
\]

(6)

\( N \) is a characteristic of the actuator, \( V \) the control voltage and \( K_p \) the actuator stiffness. The model is valid below the component resonance. For each parallel structure that constitutes a revolute joint, the dynamic model describing the piezo actuator displacement can be obtained using eq. (6) and Newton-Euler equations:

\[
\frac{I}{\lambda^2} \ddot{\delta u}_p + (K_p + K_{eq})\delta u_p = NV - \frac{\Gamma_h}{\lambda}
\]

(7)

where \( I \) is the inertia of the system mounted on the parallel structure end-effector, \( \lambda = R\sin(\epsilon) \) is the ratio between the actuator displacement and the rotation of the parallel structure end-effector, and \( \Gamma_h \) is the torque induced by the heart force. \( K_{eq} \) represents the equivalent stiffness of the compliant parallel structure acting on the actuator.

From this equation we can evaluate the resonance frequencies of the system (actuator+ transmission mechanism) and also the maximum accelerations for a given position and heart force applied on the system. The computed resonance frequencies are equal to 68Hz and 160Hz respectively for the first and second stabilizer joints. We obtain for the first axis a frequency close to the one estimated from FEA. The difference may come from the more detailed analysis obtained with the FEA, performed using a CAD definition of the mechanism.

Finally, the APA120ML actuators chosen for the mechanism allow the compensation of the device flexibilities, with end-effector accelerations of at least 6m/s², the acceleration of the free heart surface evaluated in [22], and maximum heart forces of 7N. It is also possible to compensate additional flexibilities of 0.4 mm of amplitude.

IV. Prototype development

A. Current development

The device is currently finalized. The two compliant stages have been manufactured, and the assembly of one degree-of-freedom accomplished. In Fig. 8 and 9, one can see a global view of the current development of the device and a close-up view on the compliant structure.

The device will be fully functional within a short period of time, only one part still needs to be manufactured.
B. Compensation capability

A first static test has been achieved on the current release of the prototype. A camera is used to evaluate the end-effector displacements when applying the whole range of voltage input to the actuator. In Fig. 10, one can see that the end-effector amplitude is equal to 1.36mm, i.e. the evaluated value with an error of 10%. A very good linearity is observed between the control voltage and the stabilizer tip displacement.

V. CONCLUSION

In this paper, a new active stabilizer for beating heart surgery has been introduced. The device allows a 2 DOF stabilization, using piezo actuation and a compliant architecture. A RCM mechanism is used, associated to parallel mechanisms in configurations close to singularity, to obtain a sufficient end-effector displacement amplitude. Dynamic behavior has been analyzed numerically and using a simplified dynamic model to check for the performance of the stabilizer. The prototype will now be finalized for first in vivo experiments.

REFERENCES