

Mechanism and Constitutive Model of a Continuum Robot for Head and Neck Cancer Radiotherapy

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Abstract—In this work, we present a parallel soft robot mechanism and its model for the complete immobilization and motion correction of a patient’s deviation from target in head and neck cancer radiation therapy. Each soft robot within the mechanism contracts or expands in a radially symmetric fashion around the patient’s cranial region, deforming under the isochoric property of its incompressible material. Our design goal is to provide a real-time head and neck manipulation and motion correction that is needed during frameless and maskless cancer radiation therapy. We analyze the mechanical arrangement around the patient’s cranial region, derive the deformation model of these robots and then present simulation results. Our results show consistency of behavior in meeting AAPM positioning accuracy guidelines, with less than 1mm errors in test deformation cases for the soft robot using our proposed model. In a follow-up work, we provide the kinematics and dynamics of the mechanism with control results from test cases.

I. INTRODUCTION

Radiation therapy (RT) is an increasingly effective cancer treatment modality, with more than half of all cancer patients managed with RT having higher survival rates [1]. To assure optimal treatment, it is important for the patient to remain in a stable pose on the treatment machine. Therefore, the current clinical convention is to immobilize the patient with metallic frames and rigid masks in a process called frame-and mask-based immobilization (see Figure 1). However, these immobilization systems attenuate the radiation dose, lowering treatment quality, cause patient discomfort, and have been known to cause pain for the patient – especially when the scalp has to be opened in order to plant fiducials for patient tracking [2].

Systems such as the Cyberknife (see Figure 2), though ensuring complete non-invasive radiotherapy with implanted tiny gold fiducials to differentiate tumors from healthy tissues, are incapable of closed-loop, real-time head motion correction when the treatment beam is on as they are only capable of compensating motion with pre-calculated trajectories. Furthermore, they have limited effectiveness given their non-compliant parts that assume rigidity of the patient’s body.

Techniques devised for immobilization so far in clinics range from open-loop motion compensation systems, where the whole treatment is stopped whenever the patient deviates from target to closed-loop robot-in-the-loop compensation systems. Initial



Fig. 1: Mask/Frame Immobilization in Frame-based RT

F&M RT research consisted of semi-rigid and soft immobilizing devices around the patient’s head and neck (H&N) region without a closed-loop feedback controller. These techniques were mostly evaluative studies meant to investigate the feasibility of non-rigid immobilization techniques in treatment planning [3]–[8]. Closed-loop approaches achieve a higher real-time positioning accuracy, in a technique called frameless and maskless (F&M) RT mostly for treatment planning case studies. These mechanisms utilize rigid parallel robots which provide better positioning accuracy owing to their inherent defenses against the large flexure torques and error magnification of open-loop kinematic chains.

These approach consists in using rigid electro-mechanical links connected at discrete joints to correct motions in controlled settings. Example implementations include the steel-cast assembled 4-DOF robot of [9], the HexaPOD parallel manipulator of [10], or the in-house fabricated Stewart-Gough platform of [11]. These rigid parallel robots that have emerged as a means of immobilizing patients come with the hazards of rigid manipulation systems collocated with human beings being that

- they share their dexterous workspace with the patients’ body as well as tissues – a safety concern since these

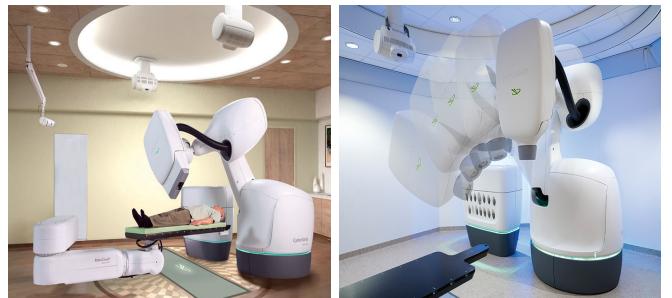


Fig. 2: The Cyberknife and 6-DOF robotic couch system.
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- robots' rigid mechanical components exhibit almost no compliance,
- they lack compliance in their structures that minimize actuator "hard shocks" on the patients' body, and
 - they are incapable of providing sophisticated motion compensation that may be needed for respiratory and internal organs displacement that often cause deviation from target.

In these implementations, the attenuation of the ionizing radiation owing to interference with the mechanical robot components is often left unresolved. It is noteworthy that recent developments such as [12] proposes moving the robot underneath the patient's body, away from the beam. However, it is not yet certain that this would fully resolve the radiation attenuation question. The human body is a natural system that needs to be manipulated with materials that can absorb much of the energy arising out of collisions, rather than rigid electro-mechanical components. To mitigate these issues, we proposed using inflatable air bladders (IABs) as motion compensators during F&M RT treatment [13]–[16]. Our IABs are continuum, compliant, and configurable (C3) soft robot (SoRo) actuators that provide therapeutic patient motion compensation during RT. These IABs inflate, deflate, extend or contract governed by their material moduli and stiffness, internal pressurization and incompressibility constraints when given a reference trajectory. To our knowledge, ours are the first to explore continuum compliant materials as actuation systems for cranial manipulation in robotic radiotherapy.

Here, we analyze and test the constitutive model of our new class of IABs. The rest of this paper is structured as follows: in § II, we present the overall continuum, compliant kinematic mechanism and analyze the deformation properties of the IAB in § III; we then provide and discuss simulation results in § IV. We conclude the paper in § V.

II. MECHANISM DESCRIPTION

In previous works [13]–[15], we relied on a data-driven system identification approach to realize the overall system dynamics. Our resultant system model lumped that of the patient, and the treatment couch with the patient and IAB links. This meant that the model lacked high-fidelity and hence the reason for the memory-based adaptive control composite laws that we derived from inverse Lyapunov analysis. We could not utilize a switched systems approach that allowed to design resilience against potential hazards during operation with a distributed control mechanism. More so, the approximation model of the ensuing neural-network controller is a feature that required extensive training to realize a suitable controller for our H&N immobilization. Our goal here is to realize constitutive models for the IAB chains and peripheral actuation mechanisms – capable of manipulating the patient's H&N's complete DOF motion in real-time during RT. This would enable us write closed-form expressions for the complete kinematic equations of the IAB chains around the patient's body, which would be useful when we write out the closed-loop feedback control laws that govern the correction mechanism design for the patient's deviation from target.

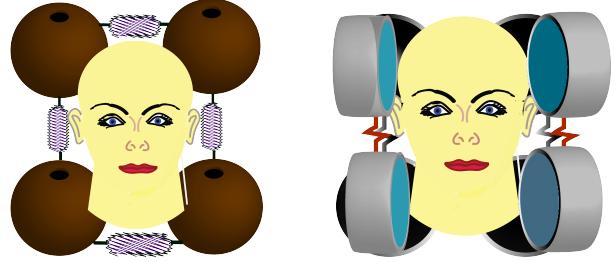


Fig. 3: An abstraction of the patient's position correction mechanism. In the left image, there are four IABs that constitute the base kinematic chain. They lift the head along the Z-axis as well as provide pitch motion corrections. On the right, the side kinematic chains provide roll and yaw motion corrections.

A. Immobilization System Setup

We now describe the mechanism of the complete immobilization system. We propose 3 IAB kinematic chains totaling 8 IABs around the patient's H&N region as illustrated in the abstraction diagram of Figure 3. There exists two IAB chains to the each side of the patient's H&N region. The side SoRo chains consist of two IABs, with each IAB making contact with the forehead and the chin/neck region at either symmetry of the patient's head. Underneath the patient's H&N is the third kinematic chain. This consists of four IABs, each positioned at four cardinal points beneath the patient's H&N region. We would like the interconnection among the IABs to have a passive actuation such that when an IAB in a chain deforms, the rest of the IAB in the chain maintain contact with the patient's H&N region. This is the purpose of the semi-plastic connectors attached between the IABs in each of the kinematic chains of Figure 3. Maintaining contact with the H&N region throughout deformation is a requirement for our the contact kinematics and H&N-IAB dynamics as described in [18]. The IABs change their configuration based on air that is conveyed into or out of their internal cavities when the sensed deviation of the patient from a target exceeds a setpoint or trajectory threshold. Each IAB is affixed to the H&N with radio-transparent semi-plastic connectors to prevent IAB free fall on the treatment table and around the patient's head. Alternatively, as in our previous design [19], the patient can wear the IAB chains using the encapsulating shield as reproduced in Figure 4.

The degrees of freedom of the mechanism of Figure 3 can be determined using *Gruëbler-Kutzbach's mobility condition*, wherein the number of degrees of freedom of a mechanism is given by (when the actuation results in a non-planar workspace configuration)

$$F = 6(N - g) + \sum_{i=1}^g f_i \quad (1)$$

where N is the number of links in the mechanism, g is the number of joints, f_i is the total number of degrees of freedom for the i th joint. When the deformation of the mechanism

results in a planar configuration, we have,

$$F = 3(N - g) + \sum_{i=1}^g f_i. \quad (2)$$

where through equations (1) and (2), the mechanism of Figure 3 have 16 DOFs since there are $N = 8$ links, $g = 8$ joints, and each joint has 2 DOFs .

B. Finite Elastic Deformation Model

Realizing a closed-form model for the immobilization mechanism requires understanding the constituent dynamics of each IAB within the overall robot mechanism. In this sentiment, we propose a finite elastic deformation model [17], [20], [21], based on the deformation invariants of the stored energy function of each IAB. In what follows, we briefly describe our motivation for devising a finite elastic deformation model for the IAB mechanism. The constant curvature approach for parameterizing the deformation of continuum robots [22]–[24] has played significant role in the kinematic synthesis of deformable continuum models over the past two decades. Under this framework, the configuration space of a continuum module is parameterized by the curvature of an arc projected on the soft robot's body, the arc's length, and the angle subtended by a tangent along that arc. By abstracting an infinite dimensional structure to 3D, large portions of the manipulator dynamics are discarded under the assumption that the actuator design is symmetric and uniform in shape. This makes the constant curvature model overly simplified so that it often exhibits poor performance in position control [25]. While the Cosserat brothers' beam theory has been relatively successful in modeling soft continuum dynamics [26], [27], its complexity, and sensing cost does not justify the alternatives. Finite element modeling methods, while accurate, require heavy computation so that they are not very useful for our real-time

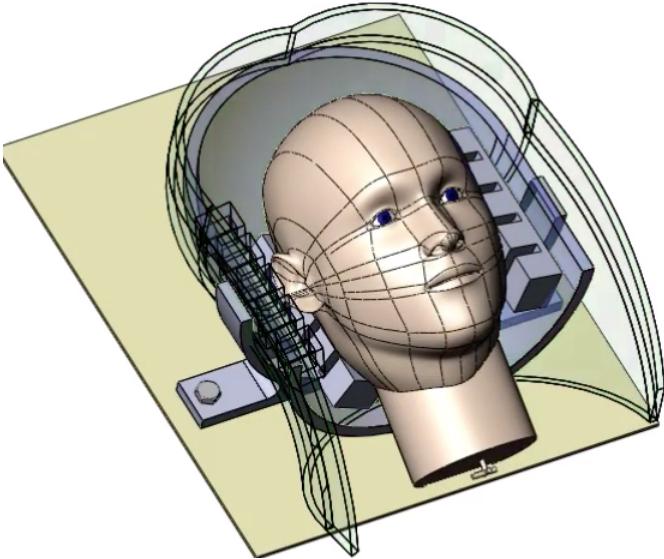


Fig. 4: Radio transparent wearable semi-plastic mask model that holds IABs in place around the patient's head and neck region.

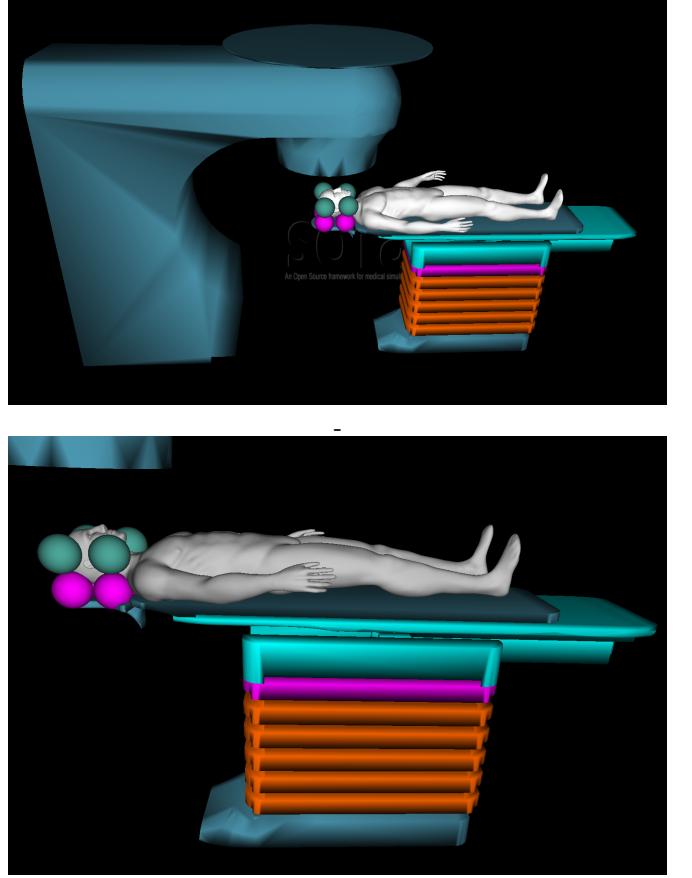


Fig. 5: System setup in the SOFA Framework Architecture. **Top:** Gantry, Turntable, Patient and IAB Chains around the patient's H&N Region. **Bottom:** Close-up view of compensating IABs around patient's H&N region with the patient lying in a supine position on the treatment couch.

control intentions. Therefore, we rigorously analyze the IABs' deformation.

Contributions:

- We propose an immobilization mechanism and construct a constitutive model for a new category of compliant, continuum and configurable (C3) IABs,
- We extend the principles of nonlinear elastic deformations [21] to isochoric strain deformations of these spherical IAB soft bodies, which are constructed out of materials with incompressible walls.
- We then analyze their deformation under stress, strain, internal pressurization, and an arbitrary hydrostatic pressure.

This kinematic model will then be used to develop the kinematics and Lagrangian dynamics of a multi-dof IAB SoRo motion correction mechanism as illustrated in Figure 5. Particularly, we analyze the deformation of a single soft robot body. In a follow-up paper [18], we address the kinematics, dynamics and control of the overall actuation system.

III. DEFORMATION ANALYSIS OF AN IAB

In this section, we present the spherical polar coordinate representation of IAB deformation, derive the deformation

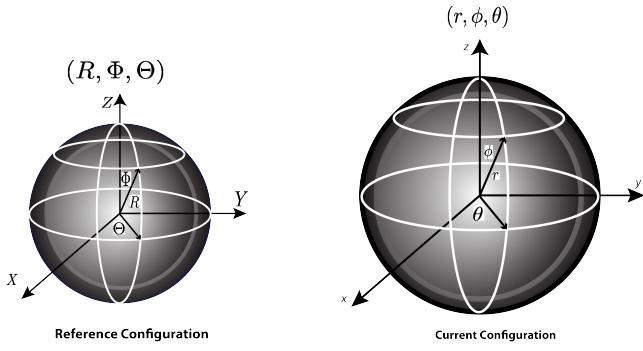


Fig. 6: Deformation in spherical polar coordinates.

gradient under assumptions of *isochoricity* and *incompressibility* of the IAB material skin. We then parameterize the configuration space of IAB kinematics based on a relationship between internal pressure, *Cauchy stress*, stored strain energy, and the radii of the IAB. The IABs are made out of rubber or elastomeric components which have the distinct property of incompressibility with a Poisson ratio of approximately 0.5 [28]. Our overarching assumption is that volume does not change locally during deformation at a configuration \mathbf{X} i.e. the deformation obeys *isochoricity*. Readers may refer to background reading materials in [17], [29] and [28].

A. Deformation Invariants

For a highly elastic, incompressible, and isotropic IAB under the action of applied forces, the deformation is governed by a stored energy function, W , which captures the physical properties of the material [30]. We choose two invariants, I_1 , and I_2 , described in terms of the principal extension ratios, $\lambda_r, \lambda_\phi, \lambda_\theta$, of the IAB's strain ellipsoids, defined as,

$$I_1 = \lambda_r^2 + \lambda_\phi^2 + \lambda_\theta^2, \quad \text{and} \quad I_2 = \lambda_r^{-2} + \lambda_\phi^{-2} + \lambda_\theta^{-2}. \quad (3)$$

Under the incompressibility assumptions of the IAB material body, we have that $\lambda_r \lambda_\phi \lambda_\theta = 1$ [21]. In spherical coordinates, the change in polar/azimuth angles as well as radii in the reference and current configurations are as illustrated in Figure 6. Forces that produce deformations are derived using the strain energy-invariants relationship, i.e., $\frac{\partial W}{\partial I_1}$ and $\frac{\partial W}{\partial I_2}$.

B. Analysis of Strain Deformations

Suppose a particle on the IAB material surface in the reference configuration has coordinates (R, Φ, Θ) defined in spherical polar coordinates (see Figure 6), where R represents the radial distance of the particle from a fixed origin, Θ is the azimuth angle on a reference plane through the origin and orthogonal to the polar angle, Φ . Denote the internal and external radii as R_i , and R_o respectively. We define the following constraints,

$$R_i \leq R \leq R_o, \quad 0 \leq \Theta \leq 2\pi, \quad 0 \leq \Phi \leq \pi. \quad (4)$$

Now, suppose that the IAB undergoes deformation under the application of pressure on the internal IAB walls as depicted in Figure 7. Arbitrary points A and A' in the reference configuration become Q and Q' in the current configuration.

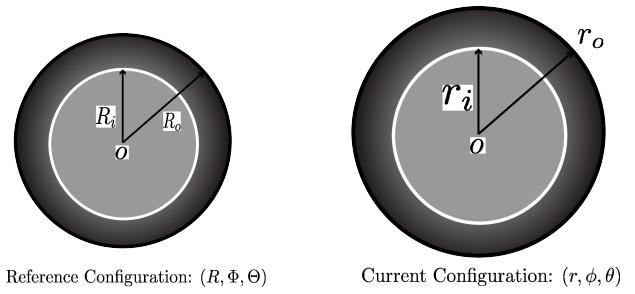


Fig. 7: Radii change under deformation.

Suppose that the vector that describes the *fiber* that connects points A and A' is $a = a_R e_r + a_\Theta e_\Theta + a_\Phi e_\Phi$ where e_R, e_Θ , and e_Φ are respectively the basis vectors for polar directions R, Θ , and Φ such that its axial length stretches *uniformly* by an amount $\lambda_z = \frac{r}{R}$. We assume that there are internal constraints such that spherical symmetry is maintained during deformation of the incompressible IAB material shell. With the spherical symmetry of the IAB preserved, we have the following constraints in the current configuration

$$r_i \leq r \leq r_o, \quad 0 \leq \theta \leq 2\pi, \quad 0 \leq \phi \leq \pi. \quad (5)$$

The radial vectors \mathbf{R} and \mathbf{r} are given in spherical coordinates as,

$$\mathbf{R} = \begin{bmatrix} R \cos \Theta \sin \Phi, \\ R \sin \Theta \sin \Phi, \\ R \cos \Phi \end{bmatrix} \quad \text{and} \quad \mathbf{r} = \begin{bmatrix} r \cos \theta \sin \phi, \\ r \sin \theta \sin \phi, \\ r \cos \phi \end{bmatrix}. \quad (6)$$

The material volume $\frac{4}{3}\pi(R^3 - R_i^3)$ contained between spherical shells of radii R and R_i remains constant throughout deformation, being equal in volume to $\frac{4}{3}\pi(r^3 - r_i^3)$ so that

$$\frac{4}{3}\pi(R^3 - R_i^3) = \frac{4}{3}\pi(r^3 - r_i^3) \\ r^3 = R^3 + r_i^3 - R_i^3. \quad (7)$$

The homogeneous deformation between the two configurations imply that

$$r^3 = R^3 + r_i^3 - R_i^3, \quad \theta = \Theta, \quad \phi = \Phi, \quad (8)$$

where the coordinates obey the constraints of equations (4) and (5). Owing to the preservation of spherical symmetry, the *Lagrangean* and *Eulerian* axes coincide, with one axis aligned to the radial axis of the sphere and the other pair oriented arbitrarily normal to it so as to form a mutually orthogonal triad. The principal stretch along the azimuthal and zenith axes is therefore $\lambda = \lambda_\theta = \lambda_\phi = r/R$. Since for an isochoric deformation, $\lambda_r \cdot \lambda_\theta \cdot \lambda_\phi = 1$, it follows that the principal extension ratios are

$$\lambda_r = \frac{R^2}{r^2}; \lambda = \lambda_\theta = \lambda_\phi = \frac{r}{R},$$

where λ is the *azimuthal* and *polar principal stretch*. It is noteworthy that in the reference configuration, the principal stretches are $\{\lambda_i\}_{i=1,2,3} = 1$ such that $I_1, I_2, I_3 = 3$. The Mooney-Rivlin strain energy for small deformations as a

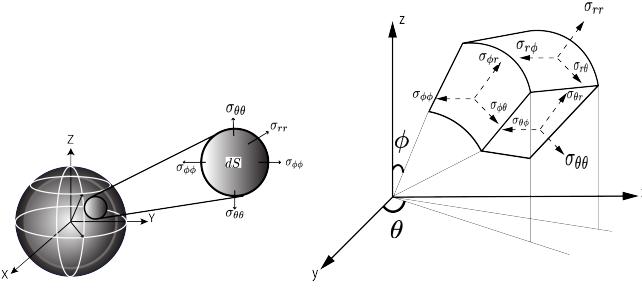


Fig. 8: Body stress distribution on continuum's differential surface, dS .

function of the strain invariants of (3), is,

$$W' = C_1(I_1 - 3) + C_2(I_2 - 3), \quad (9)$$

where C_1 and C_2 are appropriate choices for the IAB material moduli. The Mooney form (9) has been shown to be valid even for large elastic deformations, provided that the elastic materials exhibit incompressibility and are isotropic in their reference configurations [20]. For mathematical scaling purposes that will soon become apparent, we rewrite (9) as $W = \frac{1}{2}W'$ so that

$$W = \frac{1}{2}C_1(I_1 - 3) + \frac{1}{2}C_2(I_2 - 3). \quad (10)$$

Note that equation (9) or (10) becomes the neo-Hookean strain energy relation when $C_2 = 0$. The deformation gradient \mathbf{F} in spherical polar coordinates, becomes

$$\begin{aligned} \mathbf{F} &= \lambda_r \mathbf{e}_r \otimes \mathbf{e}_R + \lambda_\phi \mathbf{e}_\phi \otimes \mathbf{e}_\Phi + \lambda_\theta \mathbf{e}_\theta \otimes \mathbf{e}_\Theta \\ &= \frac{R^2}{r^2} \mathbf{e}_r \otimes \mathbf{e}_R + \frac{r}{R} \mathbf{e}_\phi \otimes \mathbf{e}_\Phi + \frac{r}{R} \mathbf{e}_\theta \otimes \mathbf{e}_\Theta. \end{aligned} \quad (11)$$

The invariant equations, in polar coordinates, are therefore a function of the right Cauchy-Green and finger deformation tensors *i.e.*,

$$I_1 = \text{tr}(\mathbf{C}) = \frac{R^4}{r^4} + \frac{2r^2}{R^2} \quad (12a)$$

$$I_2 = \text{tr}(\mathbf{C}^{-1}) = \frac{r^4}{R^4} + \frac{2R^2}{r^2}. \quad (12b)$$

C. Stress Laws and Constitutive Equations

We are concerned with the magnitudes of the differential stress on the IAB walls in spherical polar coordinates from a mechanical point of view. Since the IAB deforms at ambient temperature, we assume that thermodynamic properties such as temperature and entropy have little to no contribution. The IAB material stress response, \mathbf{G} , at any point on the IAB's boundary at time t determines the Cauchy stress, σ , as well as the history of the motion up to and at the time t . The constitutive relation for the nominal stress deformation for an elastic IAB material is given by

$$\sigma = \mathbf{G}(\mathbf{F}) + q\mathbf{F}\frac{\partial\Lambda}{\partial\mathbf{F}}(\mathbf{F}), \quad (13)$$

where \mathbf{G} is a functional with respect to the configuration χ_t , q acts as a Lagrange multiplier, and Λ denotes the internal

constraints of the IAB system. For an incompressible material, the indeterminate Lagrange multiplier becomes the hydrostatic pressure *i.e.* $q = -p$ [29]. The incompressibility isotropic assumption of the IAB material properties imply that $\Lambda \equiv \det \mathbf{F} - 1$. As such, we find from (13) that

$$\begin{aligned} \sigma &= \mathbf{G}(\mathbf{F}) - p\mathbf{F}\text{adj}^T(\mathbf{F}) \\ &= \mathbf{G}(\mathbf{F}) - p\mathbf{F}\mathbf{F}^{-T}\det(\mathbf{F}) \\ &= \mathbf{G}(\mathbf{F}) - p\mathbf{I} \end{aligned} \quad (14)$$

where the last part of the equation follows from the isochoric assumption *i.e.*, $\det(\mathbf{F}) = 1$. In terms of the stored strain energy, we can rewrite (14) as

$$\sigma = \begin{bmatrix} \sigma_{rr} & \sigma_{r\phi} & \sigma_{r\theta} \\ \sigma_{\phi r} & \sigma_{\phi\phi} & \sigma_{\phi\theta} \\ \sigma_{\theta r} & \sigma_{\theta\phi} & \sigma_{\theta\theta} \end{bmatrix} = \frac{\partial W}{\partial \mathbf{F}} \mathbf{F}^T - p\mathbf{I}, \quad (15)$$

where \mathbf{I} is the identity tensor and p represents an arbitrary hydrostatic pressure. A visualization of the component stresses of (15) on the walls of the IAB material is illustrated in Figure 8. It follows that

$$\begin{aligned} \sigma &= \frac{\partial W}{\partial \mathbf{I}_1} \cdot \frac{\partial \mathbf{I}_1}{\partial \mathbf{F}} \mathbf{F}^T + \frac{\partial W}{\partial \mathbf{I}_2} \cdot \frac{\partial \mathbf{I}_2}{\partial \mathbf{F}} \mathbf{F}^T - p\mathbf{I} \\ &= \frac{1}{2}C_1 \frac{\partial \mathbf{I}_1}{\partial \mathbf{F}} \mathbf{F}^T + \frac{1}{2}C_2 \frac{\partial \mathbf{I}_2}{\partial \mathbf{F}} \mathbf{F}^T - p\mathbf{I} \\ &= \frac{1}{2}C_1 \frac{\partial \text{tr}(\mathbf{FF}^T)}{\partial \mathbf{F}} \mathbf{F}^T + \frac{1}{2}C_2 \frac{\partial \text{tr}([\mathbf{F}^T \mathbf{F}]^{-1})}{\partial \mathbf{F}} \mathbf{F}^T - p\mathbf{I} \\ &= C_1 \mathbf{FF}^T - C_2 (\mathbf{F}^T \mathbf{F})^{-2T} - p\mathbf{I} \\ &= C_1 \mathbf{B} - C_2 \mathbf{C}^{-2} - p\mathbf{I} \end{aligned} \quad (16)$$

where as before, $\mathbf{C} = \mathbf{F}^T \mathbf{F}$ and $\mathbf{B} = \mathbf{FF}^T$. Thus, the constitutive law that governs the Cauchy stress tensor is

$$\sigma = C_1 \mathbf{B} - C_2 \mathbf{C}^{-2} - p\mathbf{I}. \quad (17)$$

It follows that the normal stresses are

$$\sigma_{rr} = -p + C_1 \frac{R^4}{r^4} - C_2 \frac{r^8}{R^8} \quad (18a)$$

$$\sigma_{\theta\theta} = \sigma_{\phi\phi} = -p + C_1 \frac{r^2}{R^2} - C_2 \frac{R^8}{r^8}. \quad (18b)$$

D. IAB Boundary Value Problem

Here, we analyze the stress and internal pressure equations of the IAB at equilibrium. Consider the IAB with boundary conditions given by,

$$\sigma_{rr}|_{R=R_o} = -P_{\text{atm}}, \quad \sigma_{rr}|_{R=R_i} = -P_{\text{atm}} - P \quad (19)$$

where P_{atm} is the atmospheric pressure and $P > 0$ is the internal pressure exerted on the walls of the IAB above P_{atm} *i.e.*, $P > P_{\text{atm}}$. Suppose that the IAB stress components satisfy hydrostatic equilibrium, the equilibrium equations for the body force \mathbf{b}' s physical component vectors, b_r, b_θ, b_ϕ are

$$\begin{aligned} -b_r &= \frac{1}{r^2} \frac{\partial r^2 \sigma_{rr}}{\partial r} + \frac{1}{r \sin \phi} \frac{\partial \sin \phi \sigma_{r\phi}}{\partial \phi} + \frac{1}{r \sin \phi} \frac{\partial \sigma_{r\theta}}{\partial \theta} \\ &\quad - \frac{1}{r} (\sigma_{\theta\theta} + \sigma_{\phi\phi}) \end{aligned} \quad (20a)$$

$$\begin{aligned} -b_\phi &= \frac{1}{r^3} \frac{\partial r^3 \sigma_{r\phi}}{\partial r} + \frac{1}{r \sin \phi} \frac{\partial \sin \phi \sigma_{\phi\phi}}{\partial \phi} + \frac{1}{r \sin \phi} \frac{\partial \sigma_{\theta\phi}}{\partial \theta} \\ &\quad - \frac{\cot \phi}{r} (\sigma_{\theta\theta}) \end{aligned} \quad (20b)$$

$$-b_\theta = \frac{1}{r^3} \frac{\partial r^3 \sigma_{\theta r}}{\partial r} + \frac{1}{r \sin^2 \phi} \frac{\partial \sin^2 \phi \sigma_{\theta\phi}}{\partial \phi} + \frac{1}{r \sin \phi} \frac{\partial \sigma_{\theta\theta}}{\partial \theta} \quad (20c)$$

(see [31]). From the equation of balance of linear momentum (*Cauchy's first law of motion*), we have that

$$\operatorname{div} \boldsymbol{\sigma}^T + \rho \mathbf{b} = \rho \dot{\mathbf{v}} \quad (21)$$

where ρ is the IAB body mass density, div is the divergence operator, and $\mathbf{v}(\mathbf{x}, t) = \dot{\chi}_t(\mathbf{X})$ is the velocity gradient. Owing to the incompressibility assumption, we remark in passing that the mass density is uniform throughout the body of the IAB material. When the IAB is at rest, $\dot{\mathbf{v}}_t(\mathbf{x}) = 0 \forall \mathbf{x} \in \mathcal{B}$ such that equation (21) loses its dependence on time. The assumed regularity of the IAB in the reference configuration thus leads to the steady state conditions for Cauchy's first equation; the stress field $\boldsymbol{\sigma}$ becomes *self-equilibrated* by virtue of the spatial divergence and the symmetric properties of the stress tensor, so that we have

$$\operatorname{div} \boldsymbol{\sigma} = 0, \quad (22)$$

Equation 22 is satisfied if the hydrostatic pressure p in (17) is independent of θ and ϕ . Therefore, we are left with (20a) so that we have

$$\frac{1}{r} \frac{\partial}{\partial r} (r^2 \sigma_{rr}) = (\sigma_{\theta\theta} + \sigma_{\phi\phi}). \quad (23)$$

Expanding, we find that

$$\begin{aligned} \frac{1}{r} \left[r^2 \frac{\partial \sigma_{rr}}{\partial r} + \sigma_{rr} \frac{\partial(r^2)}{\partial r} \right] &= (\sigma_{\theta\theta} + \sigma_{\phi\phi}) \\ \frac{1}{r} \left[r^2 \frac{\partial \sigma_{rr}}{\partial r} + 2r \sigma_{rr} \right] &= (\sigma_{\theta\theta} + \sigma_{\phi\phi}) \\ r \frac{\partial \sigma_{rr}}{\partial r} &= \sigma_{\theta\theta} + \sigma_{\phi\phi} - 2\sigma_{rr} \\ \frac{\partial \sigma_{rr}}{\partial r} &= \frac{1}{r} (\sigma_{\theta\theta} + \sigma_{\phi\phi} - 2\sigma_{rr}). \end{aligned} \quad (24)$$

Integrating the above equation in the variable r , taking $\sigma_{rr}(r_o) = 0$, and carrying out a change of variables from r to R , we find that

$$\begin{aligned} \sigma_{rr}(\delta) &= - \int_{\delta}^{r_o} \frac{1}{r} (\sigma_{\theta\theta} + \sigma_{\phi\phi} - 2\sigma_{rr}) dr, \quad r_i \leq \delta \leq r_o \\ &= - \int_{\Delta}^{R_o} \frac{1}{r} (\sigma_{\theta\theta} + \sigma_{\phi\phi} - 2\sigma_{rr}) \frac{dr}{dR} dR, \quad R_i \leq \Delta \leq R_o \\ &= - \int_{\Delta}^{R_o} \frac{R^2}{r^3} (\sigma_{\theta\theta} + \sigma_{\phi\phi} - 2\sigma_{rr}) dR, \quad R_i \leq \Delta \leq R_o \\ &= - \int_{\Delta}^{R_o} \left[2C_1 \left(\frac{1}{r} - \frac{R^6}{r^7} \right) + 2C_2 \left(\frac{r^5}{R^6} - \frac{R^{10}}{r^{11}} \right) \right] dR. \end{aligned} \quad (25)$$

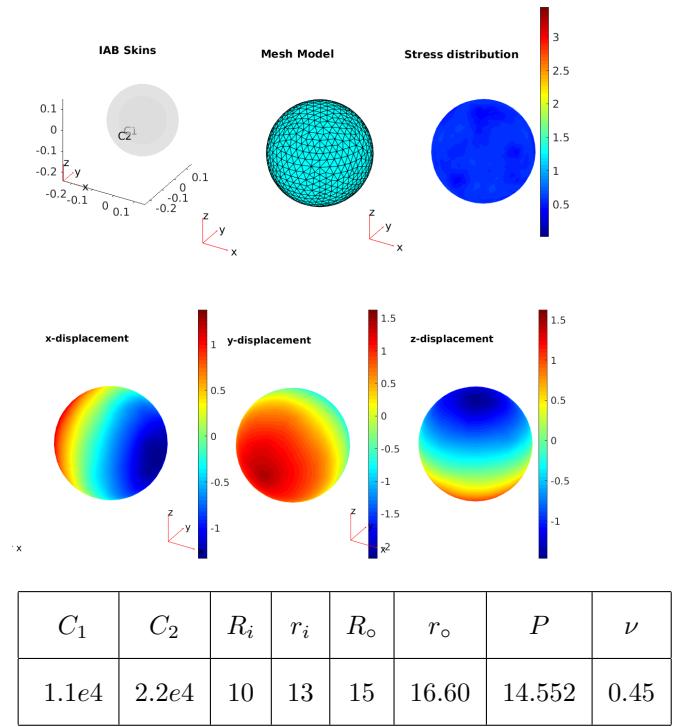


Fig. 9: IAB Expansion I

In the same vein, using the boundary condition of (19)|₂ and taking the ambient pressure $P_{\text{atm}} = 0$, we find that

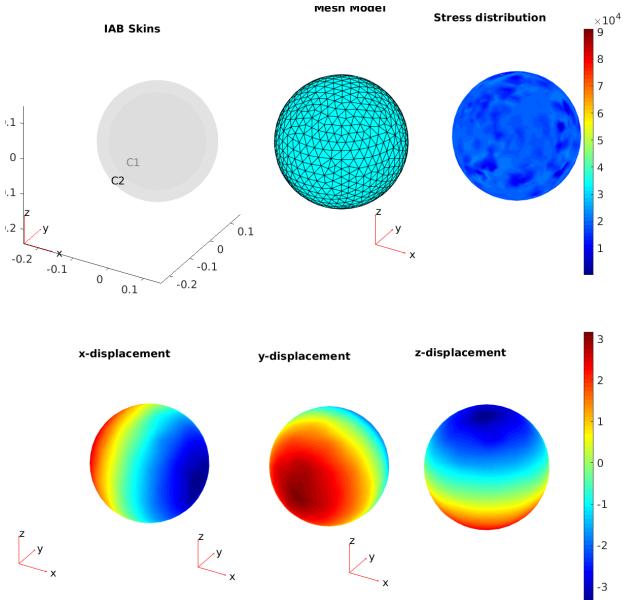
$$\begin{aligned} P &= \int_{R_i}^{R_o} \frac{R^2}{r^3} (\sigma_{\theta\theta} + \sigma_{\phi\phi} - 2\sigma_{rr}) dR, \quad R_i \leq R \leq R_o \\ &= \int_{R_i}^{R_o} \left[2C_1 \left(\frac{1}{r} - \frac{R^6}{r^7} \right) + 2C_2 \left(\frac{r^5}{R^6} - \frac{R^{10}}{r^{11}} \right) \right] dR \\ &\equiv \int_{r_i}^{r_o} \left[2C_1 \left(\frac{r}{R^2} - \frac{R^4}{r^5} \right) + 2C_2 \left(\frac{r^7}{R^8} - \frac{R^8}{r^9} \right) \right] dr \end{aligned} \quad (26)$$

where $r_i \leq r \leq r_o$. Equations (25) and (26) completely determine the deformation kinematics of the IAB material at rest. In [18], we relate the head and neck force to the contact forces on the IAB surface boundary using the component stress laws just derived.

IV. SIMULATION AND RESULTS

For a Cauchy-Elastic IAB material we would like to find the pressure using (26) that deforms the IAB under the isochoric principle from the reference configuration radius, R_i to, $r_i = R_i \pm \delta \text{ cm}$ in a current configuration for a given δ with a corresponding change in r_o as given by $r_o = \sqrt[3]{R_o^3 + r_i^3 - R_i^3}$. The results shown in Figures 9, 10, 11 and 12 respectively represent the deformation behavior of the IAB material body under extension and compression respectively with different realistic material properties (stated in the tables beneath the figures). Note that all IAB radii dimensions are in cm, the pressure is given in psi, C_1 and C_2 denote the IAB material moduli, and ν is the Poisson ratio of the body in the tables.

Figures 9 and 10 illustrate the uniform expansion of the IAB material skin when the calculated pressure is applied into their internal cavities. For the given deformation cases, the



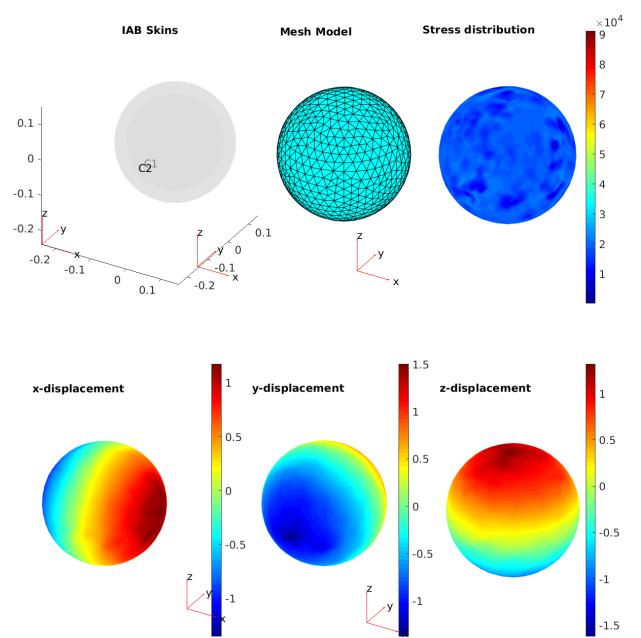
C_1	C_2	R_i	r_i	R_o	r_o	P	ν
5e5	1e6	7.5	12	10	13.21	14.5193	0.4995

Fig. 10: IAB Expansion II

calculated pressure, and associated physical properties of the IAB are as given in the tables beneath the figure. To simulate the soft body material, we apply a triangular mesh to the rigid IAB model using finite element modeling. We then compute the pressure that is required to uniformly expand the IAB from a reference to a current configuration (given by the values of R_o and r_o in the tables). The computed mesh model is shown in the top middle of the figures while the stress distribution after the application of the calculated pressure is shown in the upper-right corner of the figures. We choose a Poisson ratio of $0.45 - 0.5$ to model the incompressibility material properties of the IABs.

In the compression case of Figure 9, our goal was to uniformly expand the IAB from an outer radius of 15cm to 16cm in the current configuration. Notice a uniform expansion of 1.6cm along the three Cartesian axes of the spherical IAB in the lower charts of Figure 9. The deformation is a consequence of (26). We notice a similar consistency in Figure 10 with a required expansion of 3.21cm. The IAB extends by $\sim 3.18\text{cm}$ along all three axes – exhibiting a positioning error of 0.3mm .

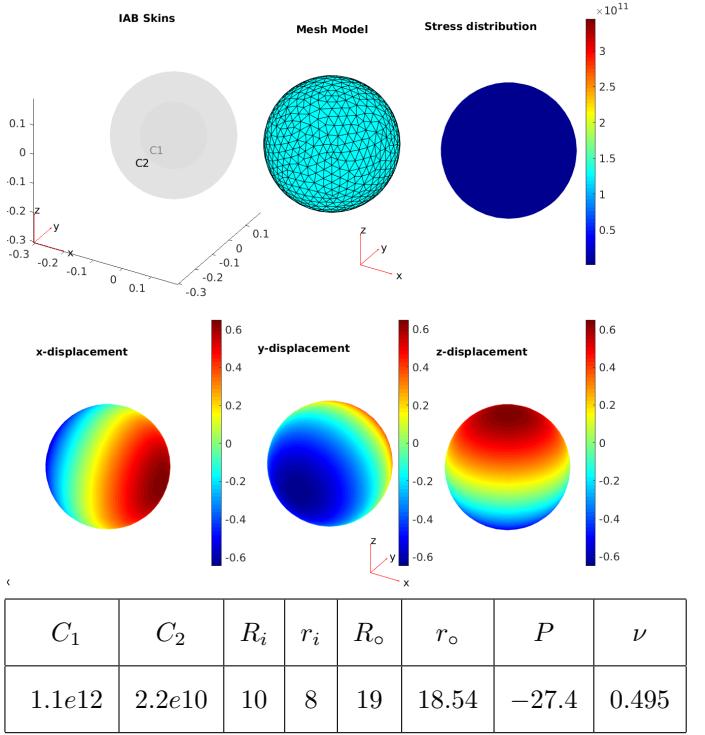
In the deformation cases of Figures 11 and 12, we present a uniform contraction of representative volumes of the isotropic and incompressible IAB under the application of the derived internal pressure for a desired radial shrink. The calculated negative pressures shown in the tables signify that air is being drawn out of the bladders. Again, we noticed an almost uniform Misses stress distribution on the IAB walls and an equal shrinking along the Cartesian axes of the IAB. For example, for a required shrink of 1.17cm from the reference configuration of Figure 11, we notice a corresponding shrinkage along the desired directions by the required amount in the lower charts



C_1	C_2	R_i	r_i	R_o	r_o	P	ν
5e5	1.2e6	12	10	15	13.83	-27.4	0.45

Fig. 11: IAB Compression I

of the figure. For both figures of Figure 11 and Figure 12, notice that the deformation precision is well below the 2mm clinical accuracy recommended by the AAPM task group guidelines [32].



C_1	C_2	R_i	r_i	R_o	r_o	P	ν
1.1e12	2.2e10	10	8	19	18.54	-27.4	0.495

Fig. 12: IAB Compression II

V. CONCLUSION

We have presented the deformation analysis of a single inflatable air bladder using a constitutive model based on Truesdell's determinism from stress principle [33]. The deformation results presented in this work are well within the positioning accuracy requirements for head and neck immobilization in cancer RT as defined in the guidelines of the American Association for Physicists in Medicine Task Group 42 [32]. In a follow-up paper [18], we analyze the contact and deformation harmonics behavior of the IABs when they come in contact with the cranial region of the head. We then proceed to construct the kinematic model for full head and neck positioning control in real-time frameless and maskless radiation therapy. Future work would include physical IAB and IAB chains build as well as the implementation of the model described in this work on test patient cases.

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