ABSTRACT
Modern surgical procedures involve flexible instruments for both diagnostic and therapeutic purposes. The implementation of flexible instruments in surgery necessitates high motion and force fidelity, and good controllability of the tip. However, the positional accuracy and the force transmission of these instruments are jeopardized by the friction and clearance inside the endoscope, and the compliance of the instrument.

The objective of this paper is to set up a 3-D flexible multibody model for a surgical instrument inside an endoscope to study its translational and rotational behavior. The 3-D model incorporates all the deformations—axial, torsion, and bending—due to its interaction with the surroundings. The interaction due to the contact is defined along the normal and tangential direction at the contact point. The wall stiffness and damping are defined in the normal direction. Friction is defined along the tangential direction. The calculation of the interaction force and moment is explained with an example.

Various simulations were performed to study the behavior of the instrument inside a curved rigid tube. The simulations for the insertion into a 3-D tube defined in a plane were compared for both 2-D and 3-D model. The simulation results from the 3-D model give the same results as the 2-D model. A simulation was carried out for the insertion in a 3-D tube using the 3-D model and the total interaction force on the instrument was analyzed. A 3-D multibody model was set up for the simulation of fine rotation. A motion hysteresis of 5° was observed for the chosen configuration.

The 3-D multibody model is able to demonstrate the characteristic behavior of the flexible instrument under different scenarios. Both translational and rotational behavior of the instrument can be characterized for the given set of parameters. The developed model will help us to study the effect of various parameters on the motion and force transmission of the instrument.

INTRODUCTION
Minimally Invasive Surgery (MIS) has greatly reduced the unnecessary damage and trauma to healthy tissues, leading to faster recovery, reduced infection rates, and reduced post-operative complications. Most of the limitations imposed by the conventional laparoscopic system are well addressed by the surgical robotic system by increasing dexterity, restoring proper hand–eye coordination and an ergonomic working position, and improving visualization [1, 2]. Furthermore, the ability of inte-
The state-of-the-art robotic surgery systems employ rigid instruments [3]. However, with conventional colonoscopy and with the emergence of Natural Orifice Transluminal Endoscopic Surgery (NOTES) and Single Incision Laparoscopic Surgery (SILS) procedures, the use of flexible instruments is inevitable. These flexible instruments are fed through access channels provided in the endoscope or endoscopic platform. The instrument tip is remotely controlled. The inherent flexibility of the instrument, coupled with the friction inside the endoscope channel and the convoluted shape of the endoscope inside the body, makes the control of the instrument tip difficult and cumbersome. As the flexible endoscopy continues to evolve more into a therapeutic tool and as the endoscopic procedures are becoming more invasive, the surgical instruments require complex manipulations [4,5]. The instrument tip needs to deliver motion and force with a required precision and accuracy. The motion and force transmission of these instruments are critical for achieving good surgical outcomes.

In an endoscope-like surgical system, the instrument is controlled from the proximal end. Nonlinearities are introduced in motion transmission by the friction forces between the instrument and the access channel. Moreover, the shape of the endoscope changes depending on the location of the surgical site. There will be a change in the force/torque delivered which is dependent on the friction properties and the shape of the contacting surfaces. Since it is difficult to place the sensors at the distal end of the instrument, the actual position and the force delivered at the instrument tip are difficult to estimate and control. This makes the control of the instrument tip difficult and challenging.

A thorough understanding of the flexible instrument behavior inside the access channel of the endoscope can lead to a proper design of the controller and eventually can lead to the automatic control of the instrument tip for the desired motion or force. This also leads to the design of the instruments not only for the functionality but also for the control. In our previous study [6,7], we described the flexible multibody model to study the sliding behavior of the flexible instrument inside a curved endoscope in the presence of friction. A 2-D flexible multibody model was set up to study the effect of friction and bending stiffness of the instrument on motion hysteresis. However, the model was limited to planar cases and the model can address only the translational behavior. In a previous study [8], the motion of a slender and flexible beam in a rigid tube was considered. A 3-D flexible multibody model is required to address the dynamic behavior in rotation and translation.

The objective of this paper is to set up a 3-D flexible multibody model to study both the translational and rotational behavior of the instrument in a 3-D environment. The flexible instrument is modeled as a series of interconnected two-noded spatial beam elements. The endoscope is modeled as a curved rigid tube of uniform circular cross-section. The shape of the curved rigid tube is defined by the center line of the tube using spatial geometric curves. The wall stiffness and damping is defined along the normal direction of the tube. The friction is defined in the tangential plane. The interaction forces are defined at the nodes of the instrument model. The axial, bending and rotational stiffnesses are defined along with the mass and inertia properties of the instrument. The calculation of interaction forces are explained subsequently. Simulations are performed for 2-D and 3-D cases and the results are compared.

In this paper, an endoscope refers to a flexible endoscope typically used for the examination of gastrointestinal tract, for example, during colonoscopy and gastroscopy procedures. However, the endoscope is modeled as a rigid curved tube. The instrument refers to the flexible instrument used for biopsy or for simple surgical procedures, which is fed through the access channel of the endoscope. The proximal end of the instrument is the base end from where the surgeon manipulates the instrument. The distal end is the tip of the instrument which interacts with the tissue directly.

The model of the flexible instrument and of the endoscope are explained in detail in the following section. The contact model and the calculation of interaction forces are explained subsequently. Various simulations were performed to study the behavior of the instrument. The simulation results are discussed thereafter.

**MODELING OF A FLEXIBLE SURGICAL INSTRUMENT**

The surgical instrument is modeled as a series of interconnected two-noded spatial beam elements. The endoscope is modeled as a rigid tube of uniform circular cross-section. The shape of the rigid tube is defined by a center line of the tube using spatial geometric curves. The contact between the beam and the tube is defined at the nodes of the beam elements.

A computer program SPACAR [9] is used for the modeling and simulation of the flexible surgical instrument. SPACAR is a modeling and simulation tool for multibody dynamic analysis of planar and spatial mechanisms with rigid and flexible elements.

**A Flexible Surgical Instrument As Flexible Beam**

The model of the flexible instrument together with the model of the tube is shown in Fig. 1. The origin of the global frame, O, is situated at the beginning of the tube and the initial tangential direction is the X-axis. The encircled number, , represents the \( n \)-th beam element. The nodes are represented by the numbers.

The surgical instrument is modeled as a series of interconnected flexible beam elements as available in the SPACAR program. Each beam element has a node at either end. The beam element has six degrees of freedom defined at each node—three
The number of degrees of freedom of the element as a rigid body is six. Consequently, the beam element has six deformation modes defined as [10, 11]:

\[
\begin{align*}
\text{elongation: } & \quad \epsilon_1 = l_0 - l_0, \\
\text{torsion: } & \quad \epsilon_2 = 2(\mathbf{e}_q^p \cdot (\mathbf{e}_q^p - \mathbf{e}_0^p))/2, \\
\text{bending: } & \quad \epsilon_3 = l_0\mathbf{e}_p^q - \epsilon_2^p, \quad \epsilon_4 = l_0\mathbf{e}_p^q - \epsilon_2^p, \\
& \quad \epsilon_5 = l_0\mathbf{e}_p^q - \epsilon_2^p, \\
& \quad \epsilon_6 = l_0\mathbf{e}_p^q - \epsilon_2^p, \\
\end{align*}
\]  

(2)

where \( l = \|\mathbf{x}^q - \mathbf{x}^p\| \) is the distance between the nodal points, \( l_0 \) is the reference length of the element, and \( \mathbf{e}_q = (\mathbf{x}^q - \mathbf{x}^p)/l \) is the unit vector directed from node \( p \) to node \( q \). The first and second deformation modes, \( \epsilon_1 \) and \( \epsilon_2 \), describe the elongation and torsion of the element. The other four deformation modes, \( \epsilon_3 - \epsilon_6 \), are related to the bending deformations of the element.

A distributed mass along the center line and lumped moments of inertia at the nodes for the rotational inertia are defined. The detailed description of the element and its stiffness and inertia properties can be found in [11, 12].

**An Endoscope As A Rigid Curved Tube**

The endoscope is modeled as a rigid curved tube of uniform circular cross-section. The shape of the tube is defined by the center line of the tube. The center line is defined by a straight line, a circular arc, a Bézier curve, or a combination of these. A planar case was explained in [6, 7], and can be referred to for the detailed description. A 3-D cubic Bézier curve is defined by a set of four control points. The control points, \( \mathbf{P}_1 \) and \( \mathbf{P}_4 \), coincide with the endpoints of the curve. The shape of the curve is determined by the interior control points, \( \mathbf{P}_2 \) and \( \mathbf{P}_3 \) [13, 14].

A part of the 3-D tube can be easily defined by selectively choosing the control points. Moreover, the straight tubes and circular tubes can be also used to define the overall length of the endoscope. A 3-D Bézier curve was considered for the study as it covers the typical steps required for the calculation of interaction forces and moments.

**Contact Model**

The contact between the beam and the wall is defined at the nodes of the beam elements. As a node approaches the inner wall of the tube, the node experiences a normal force depending on the depth of penetration and the rate of penetration. The wall stiffness and damping are defined normal to the surface. Friction at the contact point is also defined. Therefore, depending on whether there is any sliding motion at the contact point, the node experiences a friction force in the tangential direction.

**Normal Force.** Three contact regions are defined: no contact, full contact, and transition [6, 7]. Figure 3 shows the
cross-sections of the tube and the instrument at the contact. A transition zone is defined between \( r_a \) and \( r_b \) where the wall stiffness and damping varies continuously. The transition zone makes the normal reaction force continuous and makes the overall computation faster. The net normal reaction force \( F_n \), depending on the normal displacement \( x_n \) and the normal velocity \( v_n \), is given by

\[
F_n = \begin{cases} 
0 & \text{if } x_n < a \\
-(k/2)(b-a)\xi^2 - c_v(3-2\xi)\xi^2v_n & \text{if } a \leq x_n \leq b \\
-k(b-a)(\xi-1/2) - c_vv_n & \text{if } x_n > b
\end{cases}
\]

where \( \xi = (x_n - a)/(b-a) \), \( k \) is the wall stiffness, and \( c_v \) the wall damping coefficient. \( v_n \) is the velocity in the normal direction. \( a \) and \( b \) are the initial clearances from the \( r_a \) and \( r_b \), when the instrument lies on the center line of the tube, given by \( r_b - r_o \) and \( r_b - r_o \) respectively. If the value of \( F_n \) is positive, it is replaced by zero.

**Friction Force.** The friction force is calculated from a Coulomb friction force model that is made a continuous function of the sliding speed \( v_l \) at zero and is given by

\[
F_l = -F_c \tanh(c_vv_l)
\]

where \( F_c = \mu F_n \) is the Coulomb friction, \( \mu \) is the coefficient of friction between the contacting surfaces, \( F_n \) is the normal reaction force at the contact (Eqn. (3)), and \( c_v \) is the velocity coefficient which determines the width of the transition region near zero sliding velocity, \( v_l \) [7]. The friction force is directed to the opposite direction of the sliding speed as defined in (Eqn. (13)).

![Figure 3. Cross-sections of the tube and the instrument at the contact.](image)

**Calculation of Interaction Forces and Moments.** The calculation of interaction forces and moments requires finding the location of the interacting node \( P_o \) with respect to the base point \( P_c \) on the center line of the curved tube. Figure 4 shows one of the center lines of the tube defined by a Bézier curve. \( P(u) \) is the location of a point on the Bézier curve depending on the parameter \( u \) for \( 0 \leq u \leq 1 \) [13, 14]. The base point \( P_c \) is the point on the curve such that the vector \( (P_o - P_c) \) is perpendicular to the tangent vector \( P_c' \). In the case of straight and circular tubes, the base point can be obtained analytically. However, for a Bézier curve, the point \( P_c \) is obtained iteratively by solving

\[
(P_o - P_c) \cdot P_c' = 0 \quad (5)
\]

for \( 0 \leq u \leq 1 \). The value of \( u \), satisfying (Eqn. (5)), locates the point \( P_c \). In order to have a unique solution, \( P_o \) should be such that the normal distance \( d_{\text{norm}} \) is smaller than the minimum radius of curvature \( R_{\text{min}} \). The normal distance between \( P_o \) and \( P_c \) is given by

\[
d_{\text{norm}} = \|P_o - P_c\| \quad (6)
\]

**Force vector triad.** There are three orthogonal vectors defined at the interacting node \( P_o \), to calculate the forces acting at the node. Figure 4 shows the force triad attached to the node. It is assumed that the cross-sections of the tube and of the instrument are circular. As the instrument makes a contact with the inner wall of the tube, three directions are defined at the point of contact. The unit vector \( \hat{e}_1 \) is defined along the radial direction and expressed as

\[
\hat{e}_1 = \frac{(P_o - P_c)}{\|P_o - P_c\|} \quad (7)
\]

Clearly, there is no interaction if the interacting node lies on the center line of the tube. The second orthogonal unit vector \( \hat{e}_2 \) is
defined along the axial direction and determined by the tangent at the base point $P_c$ on the center line of the curved tube

$$\hat{e}_2 = \frac{P'_c}{\|P'_c\|}$$

(8)

The third orthogonal unit vector $\hat{e}_3$ is defined along the circumferential direction and is obtained from the cross product as

$$\hat{e}_3 = \hat{e}_1 \times \hat{e}_2$$

(9)

The vectors $\hat{e}_2$ and $\hat{e}_3$ are parallel to the tangent plane. The sliding velocity on the tangent plane determines the direction of the friction force.

**Velocity at the point of contact.** If the interacting node has a translational velocity $v_o$ and angular velocity $\omega_o$, the resultant velocity $v$ at the contact point is given by

$$v = v_o + r_o \omega_o \times \hat{e}_1$$

(10)

where $r_o$ is the radius of the flexible instrument.

The velocity vector $v$ can be decomposed into normal and tangential components

$$v = v_n \hat{e}_n + v_t \hat{e}_t$$

(11)

where $v_n = v \cdot \hat{e}_1$ is the velocity component normal to the wall and the normal direction is given by $\hat{e}_n = \hat{e}_1$. The tangential velocity component $v_t$ is given by

$$v_t = \| (v \cdot \hat{e}_2) \hat{e}_2 + (v \cdot \hat{e}_3) \hat{e}_3 \|$$

(12)

and the tangential direction $\hat{e}_t$ is given by

$$\hat{e}_t = \frac{(v \cdot \hat{e}_2) \hat{e}_2 + (v \cdot \hat{e}_3) \hat{e}_3}{\| (v \cdot \hat{e}_2) \hat{e}_2 + (v \cdot \hat{e}_3) \hat{e}_3 \|}$$

(13)

Figure 5 shows the velocity components on the tangent and normal plane at the point of contact $P$. The directions of force components due to the interaction are also shown.

**Resultant force and moment acting at the node.** The resultant force $F$ acting at the contact point $P$ of the instrument by the inner wall of the tube is given by:

$$F = F_n \hat{e}_n + F_t \hat{e}_t$$

(14)

where $F_n$ and $F_t$ are the normal and tangential forces given by Eqn. (3) and Eqn. (4) respectively. Figure 6 further illustrates the force components acting on the instrument at the point of contact.

Therefore, the resultant force $F_o$ and moment $M_o$ acting at the node $P_o$ due to the total interaction force $F$ acting at the contact point $P$ are given by

$$F_o = F$$

(15a)

$$M_o = r_o \hat{e}_n \times F$$

(15b)

where the direction of the moment is determined by the vectors $\hat{e}_n$ and $F$, and will be perpendicular to the normal plane.

**Discussion**

The model of the instrument is defined using spatial beam elements. It is assumed that the deformation in the individual elements are small. However, the entire instrument can show large deformation. The instrument model encompasses all the deformations possible due to the axial, bending, and torsional...
interactions. The calculation of the interaction force and moment has included the resultant effect due to the thickness of instrument. The actual interaction force acts on the side of the instrument and the point of contact is offset by the radius of the instrument cross-section. This offset causes an extra bending moment and also results into a twisting moment.

The instrument remains inside the curved tube. It is assumed that the deformation on the wall at the point of contact is small as compared to the total deformation of the instrument. The tangent planes at the point of contact on both contacting surfaces are the same. Stiffness and damping are defined at the inner wall of the tube. Damping reduces the oscillating behavior of the instrument and allows faster convergence to the stable solution.

**SIMULATION**

The interaction force between the instrument and the curved tube is implemented in the SPACAR program through a user-defined routine. Some of the dynamic DOFs can be suppressed to reduce the computation time. For example, the axial and torsional deformations of the beam element can be suppressed while simulating for the beam insertion without rotation inside the planar tube. The axial deformation is suppressed in all the simulation cases as the ratio of stiffness along the axial direction in the planar tube. Damping reduces the oscillating behavior of the instrument and ensures that the instrument does not buckle outwards respectively. The force is directed towards the normal of the circular section of the curved tube. The axial deformation of the beam elements is suppressed. The coefficient of friction used in the simulation is zero.

As the instrument advances through the curved rigid tube, the interaction force at the various nodes is calculated based on the contact model defined and applied at the interacting nodes. Figure 7 shows the plot of forces acting at the first two distal nodes. It can be observed that the node 1 makes contact with the outer wall of the tube, whereas the node 2 with the inner wall as expected. Accordingly, the forces $F_1$ and $F_2$, acting at the first two distal nodes when the instrument advances through the circular tube. $X_1$ and $X_2$ are the nodal positions of the respective nodes. It can be observed that the node 1 makes contact with the outer wall of the tube, whereas the node 2 with the inner wall as expected. Accordingly, the forces $F_1$ and $F_2$ are directed inwards and outwards respectively. The force is directed towards the normal at the point of contact. The center line of the tube is also shown. As the instrument is confined in the tube and the cross-section of the tube is much smaller than the radius of curvature of the tube, the nodal positions are not discernible from the center line. The forces at the first two distal nodes are shown for clarity and explanation. Similar plots can be obtained for all the interacting nodes.

**Insertion in a Curved Rigid Tube Defined in xy-plane**

In this simulation, a 3-D model is set up in order to validate it with the previously developed 2-D model [6, 7]. A curved rigid tube is defined in xy-plane. The shape of the curved tube is defined by an arc of 90° in xy-plane with straight sections at both ends of the arc. The straight sections at the entry and exit are along the x- and y-axes respectively. The radius of the arc is 0.5 m. The arc together with the straight lines at the two ends defines the center line of the tube. The straight section in the beginning of the curved tube provides a guide way to the instrument and ensures that the instrument does not buckle under the influence of interacting forces in the beginning of the insertion. The properties of the beam element and of the curved rigid tube are given in Table 1.

The total length of the instrument considered is 1.0 m, and 10 equal length beam elements are used to model the instrument. The proximal end is constrained: its rotation is zero and a translation motion in the longitudinal direction is prescribed along the x-axis. The velocity profile for the translation is trapezoidal, with an initial acceleration of 0.01 m/s², a constant velocity of 0.01 m/s, and a deceleration of 0.01 m/s² until the velocity is zero again. It remains at rest after the insertion. A 0.90 m length of the instrument is inserted inside the curved tube. After the insertion of the instrument, the distal end is already out of the circular section of the curved tube. The axial deformation of the beam elements is suppressed. The coefficient of friction used in the simulation is zero.

<table>
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<th>Description</th>
<th>Unit</th>
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<td>Density, $\rho$</td>
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Comparison with the 2-D model simulation results.

The forces at the distal node are compared for both the 2-D model and the 3-D model. Figure 8 shows the plot of the force exerted on the instrument at the distal end. The force components $F_{x1}$, $F_{y1}$, and $F_{z1}$ are shown as the instrument advances through the tube. The distal end of the instrument makes contact with the inner wall of the tube and experiences a force normal to the contacting surface. At 39.7 s, $F_{y1}$ is equal to $F_{x1}$ as the distal end reaches the arc length corresponding to 45°. At 79.0 s, $F_{y1}$ is zero as the distal end is out of the circular tube. The distal end further moves in the straight section of the tube until the 0.90 m length of the instrument is inserted. $F_{x1}$ is decreasing in this section. As the distal end reaches the straight section along the y-direction, there is a force only along the x-direction. $F_{z1}$ is zero as expected. The plots for $F_{x1}$ and $F_{y1}$ from the 2-D and 3-D models are overlapping and there is no discernible difference between the two plots.

Figure 9 shows the plot of total force exerted on the instrument during insertion in the circular tube. The force components $F_x$, $F_y$ and $F_z$ are shown and the simulation results are compared for the 2-D and 3-D models. $F_z$ is zero as the instrument and the tube model is confined in $xy$-plane. $F_x$ and $F_y$ are same and there is no difference in the two simulation results.

Insertion in a 3-D Curved Rigid Tube

In the second simulation, we consider a 3-D curved rigid tube. The rigid tube is defined by a Bézier curve, selecting four control points, $P_1(0,0,0,0,0.0)$, $P_2(0.6,0.0,0,0.0)$, $P_3(0.0,−0.4,−0.2)$, and $P_4(0.6,−0.4,0.0)$. Straight sections are defined at the entry and exit of the curved tube. The friction is assumed to be zero. A stainless steel wire of diameter 0.5 mm is used to calculate the parameters for the instrument model. The data used for the simulation are given in Table 1. The velocity profile used for the insertion is same as in the previous simulation.

Figure 10 shows the plot of forces $F_1$ and $F_2$, acting at the first two distal nodes when the instrument advances through the 3-D curved tube. As the curvature and torsion of the tube changes, the force direction also changes. The force is directed towards the normal at the point of contact. $X_1$ and $X_2$ are the nodal positions of the respective nodes. Figure 11 shows the magnitude of the forces acting at the first two distal nodes as the instrument advances through the tube. The forces at the first two distal nodes are shown for clarity and explanation.

Figure 12 shows the plot of the total force acting on the instrument while advancing through the tube. The total force was obtained by summing all the forces acting at the contacting nodes as the instrument progresses through the tube. The force components $F_x$, $F_y$ and $F_z$ are shown. The curved rigid tube will experience the same force with opposite sign.
Simulation of Fine Rotation

A 3-D multibody model is set up for the simulation of fine rotation. A rigid tube consists of a straight part in the beginning, a circular section of 90° with a radius of 300 mm, and another straight section at the exit. A stainless steel wire of diameter 0.5 mm and length 500 mm is modeled using 10 equal length spatial beam elements. It is assumed that the wire is straight in the beginning. The total length of 485 mm is inserted through the circular section of the tube so that the distal end is in the straight section of the tube. The initial deformation is obtained from the simulation of wire insertion.

A sinusoidal rotation motion with an amplitude of 180° and a frequency of 1 Hz is given to the proximal end and the rotation of the distal end is observed. The value of coefficient of friction used for the simulation is 0.2. Figure 13 shows the plot of motion hysteresis in rotation. The rotation of the distal end along the longitudinal axis was compared with the rotation of the proximal end. The simulation result showed a motion hysteresis of 5°.

Figure 14 shows the plot of the reaction moment at the proximal end. The reaction moment along the z-axis is constant and equal to $1.42 \times 10^{-3}$ Nm. This is true as the wire is confined to the curved rigid tube in xy-plane. A bending moment of $2.05 \times 10^{-3}$ Nm is required to deform the wire in a circular arc of radius 300 mm. The observed value from the simulation is less than the calculated value as the wire ends are in the straight sections of the tube and there is a clearance in the tube also. The moment in x-direction is due to the total friction moments acting at the nodes.
DISCUSSION

The 3-D flexible multibody model of the instrument incorporates all the deformations possible due to its interaction with the surroundings. The mechanical properties of the instrument and of the interacting surface can be varied; and the effect of various parameters on the characteristic behavior of the instrument can be studied. The shape of the tube can be varied according to the need. However, the shape of the tube does not change due to its interaction with the instrument. In many cases, this assumption can be valid as the endoscope is supported by the surrounding organs or part of the anatomy where it is inserted. The endoscope eventually provides the rigid support to the instrument. Nonetheless, the interaction of the instrument tip with the tissues exerts higher forces on the endoscope and the endoscope cannot provide the rigid support as intended. Therefore, a flexible tube model will be required to address such interactions. The developed model is limited to the interaction with the rigid tube.

The simulation results from the 2-D and 3-D models are same, but at the cost of higher computation time. Though the 3-D model is a generic model and all types of motions and interactions can be defined, the 3-D model should be used at one’s own discretion. The simulation results also give us the confidence in characterizing the instrument’s behavior. The effect of friction and the rotational stiffness on the motion hysteresis in rotation and force transmission can provide an insight into designing the flexible instrument for surgical intervention. The results of the study will be reported subsequently.

An experimental set-up was designed to validate the model and simulation results. The experimental validation of the model will be reported in the subsequent article.

CONCLUSION

A 3-D flexible multibody model of the instrument was set up. The contact model and the 3-D model of the curved rigid tube was appropriately defined to address the translational and rotational behavior of the instrument in the presence of friction. The developed 3-D model incorporates all the deformations possible—axial, torsion, and bending—due to its interaction with the surroundings. Various simulations with the 3-D model provide an insight into the characteristic behavior of a flexible instrument inside a curved rigid tube. The simulation results with the 3-D model are in good agreement with the previously developed 2-D model for the planar case. The developed 3-D model can be given different input motion and the dynamic behavior of the instrument can be analyzed for different shapes of the 3-D curved tube. The simulation for fine rotation shows a motion hysteresis of $5^\circ$ for the chosen configuration. The developed 3-D model will help us to study the effect of various parameters on the motion and force transmission of the instrument under different working environments. The model will be further validated by comparing simulation results with the experimental data.

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