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Flexible Impulse Transfer using a Newton's Cradle Inspired Catheter: A Feasibility Study

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Abstract

A major challenge encountered during minimal invasive surgery is transferring high forces through small and flexible instruments, such as needles and catheters, due to their low buckling resistance. In this study we have determined the feasibility of using a Newton's Cradle-inspired catheter (patented) to transfer high force impulses. Exerting a high-force impulse on the tissue increases the critical buckling load and as such can prevent buckling. The system consisted of an input plunger onto which the impulse is given, a (flexible) shaft filled with Ø2 mm stainless steel balls, and an output plunger to transfer the impulse to the target tissue. In the proof-of-principle experiment, the effect of clearance (0.1, 0.2, and 0.3 mm), length (100, 200, and 300 mm), shaft type (rigid vs. flexible), curve angle (0, 45, 90, 135, and 180°), and curve radius (20, 40, 60, and 100 mm) on the efficiency was determined. The catheter was able to deliver forces of 6 N without buckling. The average impulse efficiency of the system was 35%, which can be further increased by optimizing the design. As such, this technology is promising for high force delivery in miniature medical devices during minimal invasive surgery in future.

Keywords: Buckling, Catheter, Feasibility, Force, Guidewires, Impulse, Medical Device Design, Momentum, Minimal Invasive Surgery, Newton's Cradle.

1   **1. Introduction**

2   In order to demonstrate the conservation of momentum and energy (see Eq. 1), Sir Isaac Newton developed the *Newton's*

3   *Cradle*. *Newton's Cradle* consists of a series of identical metal balls suspended in a metal frame by two wires in a v-

4   formation (Fig. 1). If one ball is pulled away from the other balls and is subsequently released, it will gain momentum p (the

5   product of its mass m [kg] times its velocity v [m/s]), which will subsequently be converted into an impulse J (the integral of

6   the force F [N] over the time dt [s] for which it acts) when it strikes the second ball (see Eq. 1). This second ball will

7   acquire, and subsequently transfer, the impulse to the third ball, and so on; producing a pressure wave that propagates

8   through the intermediate balls. When the pressure wave reaches the most distal ball, this ball is able to gain momentum and

9   will, subsequently reverse the direction of motion of the pressure wave. Assuming entirely elastic behavior, momentum loss

10   is zero, allowing for a successive oscillating motion of the two outer balls.

11

12   $$dp = m \cdot dv = \sum F \cdot dt \qquad (1)$$

13

14   with:

15   dp = change in momentum p [kgm/s]

16   m = mass [kg]

17   dv = change in velocity v [m/s]

18   F = force [N]

19   dt = time interval for which the force F acts [s]

20

21   A *Newton's Cradle* inspired mechanism, such as illustrated in Fig. 2, can be interesting for medical applications in

22   which high forces need to be transferred by flexible or ultrathin devices, such as guidewires, catheters, or needles, which are

23   susceptible to buckling. Buckling is characterized by a sudden sideways deflection of the slender device subjected to high

24   compressive loads, which is unwanted as it can cause damage to the surrounding tissues and decreases the force that can be

25   applied by the tip, e.g., when puncturing a coronary occlusion in the heart using a guidewire. Exerting an impulse on the

26   tissue can allow for applying high forces on tissues as it prevents buckling in two main ways. (1) The use of an impulse

27   increases the maximum compressive load, also known as the critical load, the device can withstand before buckling [1]. (2)

28   Using an impulse to penetrate tissues can decrease the penetration load by minimizing energy losses to the environment.

29   This effect can be mainly attributed to the inertia of the target tissue that acts as a counterforce to the incoming impulse, and
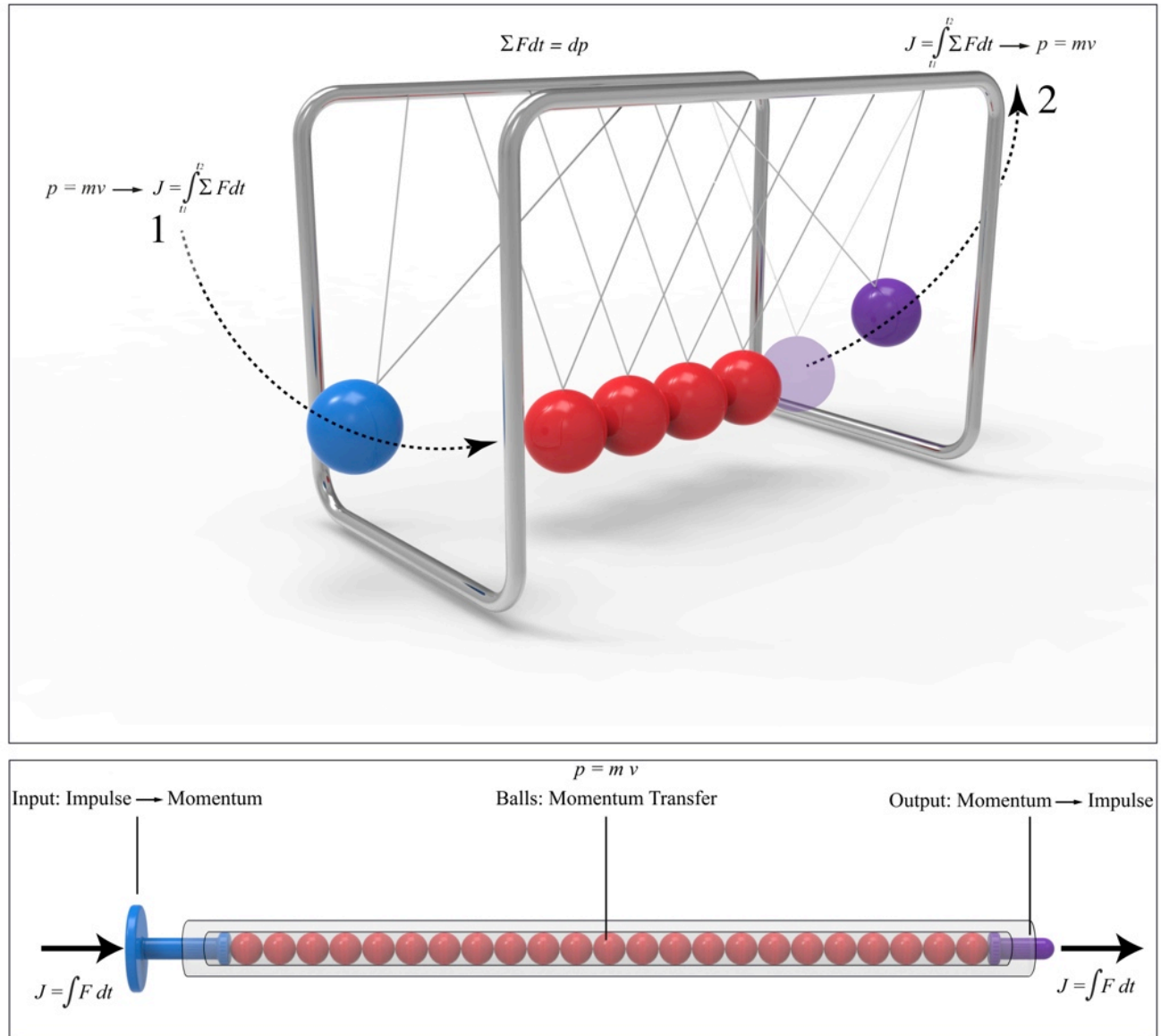
Fig. 1. *Newton's Cradle* mechanisms. Top: *Newton's Cradle*. Bottom: Schematic representation of the *Newton's Cradle's* inspired catheter. Color indications: blue = input ball, purple = output ball, and red = intermediate balls.

thus decreases energy loss by tissue movement and deformation [2, 3] (see also Sakes *et al.* [4] on this topic). In this study we will, therefore, investigate the feasibility of using a *Newton's Cradle* inspired mechanism for low-friction transfer of high-force impulses through slender shafts. For more information on the theory of the *Newton's Cradle* we would like to refer to the studies of Kinoshita *et al.* [5], Ceanga *et al.* [6], and Donahue *et al.* [7].

Depending on the specific medical application, the *Newton's Cradle* inspired mechanism might need to conform to different requirements. We will distinguish between two types of medical applications: (1) needle applications and (2)

40   catheter applications. In the needle applications, rigid, preferably hollow, shafts are needed, in order to administer medicine

41   or take a biopsy (amongst others). In catheter applications, on the other hand, a flexible shaft is needed to travel through the

42   vasculature from the incision point to the lesion site. The dimensions and flexibility of the shaft should conform to that of

43   currently available instruments. A specific catheter application that would benefit from high force delivery through a

44   flexible shaft can be found in the treatment of chronically occluded coronary arteries, in which forces of over 1.5 N need to

45   be delivered for puncture.

46   **2. Materials and Methods**

47   *2.1. Experimental Goal*

48   The goal of our experiment was to explore the possibility of using a *Newton's Cradle* inspired mechanism, and thus a series

49   of balls, to transfer an impulse through a long, slender, shaft. In order to achieve this goal, the quality of the impulse transfer

50   in terms of efficiency, as well as the ability of the *Cradle* prototype to exert high peak forces without buckling, needed to be

51   determined. For comparison, a Ø2 mm stainless steel needle (wall thickness 0.2 mm, L = 300 mm) can withstand

52   approximately 2.6 N before buckling and a similar size catheter only 0.003 N [8].

53   *2.2. Cradle Prototype*

54   The *Cradle* prototype consisted of a Ø2.5 mm shaft filled with Ø2 mm stainless steel balls (Fig. 2). The balls were

55   contained within a shaft with two polymer caps at either side. At the proximal and distal end, Ø1.5 mm plungers were

56   placed that transferred the input impulse to the balls and propagating pressure wave of the balls into an output impulse onto

57   the target, respectively. In order to ensure contact between the balls, a compression spring was placed distal to the output

58   plunger. By altering the pretension between the balls, the bending stiffness of the *Cradle* prototype could be altered. During

59   insertion, the pretension is removed to ensure the bending stiffness of the *Cradle* prototype does not exceed that of currently

60   available support catheter to prevent blood vessel wall damage. It must be noted, however, that the *Cradle* prototype is not

61   torsion stiff.

62   *2.3. Experimental Variables*

63   *2.3.1. Dependent Variables*

64   In order to determine the quality of the impulse transfer, the following dependent variables were measured at a fixed input

65   impulse:

66   **1.**   **The output impulse (J_o)** delivered by the output plunger.

67   **2.**   **The output peak force ($F_{peak\ out}$)** delivered by the output plunger.

68   *2.3.2. Independent variables*

69 The effect of the specific configuration of the *Cradle* prototype on the impulse transfer was researched by sequentially

70 altering the following independent variables:

71   **1.**   **Clearance (c).** The inner diameter of the shaft was varied to determine the influence of the clearance between the

72   balls and shaft. The clearance will most likely have an effect on the impulse and peak force efficiency due to two

73   main effects: (1) friction between the balls and the shaft and (2) misalignment of the balls. The clearance between

74   the balls and the shaft was set to 0.1, 0.2, and 0.3 mm.

75   **2.**   **Shaft type.** In order to determine the effect of the shaft material on the efficiency of the device, two different shafts

76   were tested: a rigid stainless steel shaft and a double-braided axially and radially stiff cardiac catheter (*Mach 1^{tm}*

77   8F, Boston Scientific, Marlborough, MA, USA).

78   **3.**   **Length (L).** During each collision, energy will be lost, mainly due to friction between the balls and the shaft. The

79   number of balls, and thus contact points, will thus likely have an effect on the impulse and peak force efficiency.

80   The effect of the number of balls on the energy loss was determined by setting the length of the shaft to 100, 200,

81   and 300 mm, resulting in a total of 50, 100, and 150 Ø2 mm balls in the shaft, respectively.

82   **4.**   **Curve angle (α)**. In endovascular interventions, it is a necessity to cross several intersections and curves in the

83   body. In order to determine the effect of these curves on energy loss, it was decided to subject the prototype to

84   single-constant radius curves ($r = 40$ mm) with four different curve angles: 45, 90, 135, and 180° and to compare

85   the impulse and peak force efficiency to that in the straight configuration.

86   **3.**   **Curve radius (r).** The effect of the curve radius was researched by subjecting the prototype to single-constant

87   radius curves with a curve angle $\alpha = 90°$ and four different curve radii: 20, 40, 60, and 100 mm.

88 The **input impulse (J_i)** and the associated **input peak force ($F_{peak\ in}$)** delivered onto the input plunger were set at 0.4 Ns

89 and 5 N, respectively, and were measured throughout the experiments.
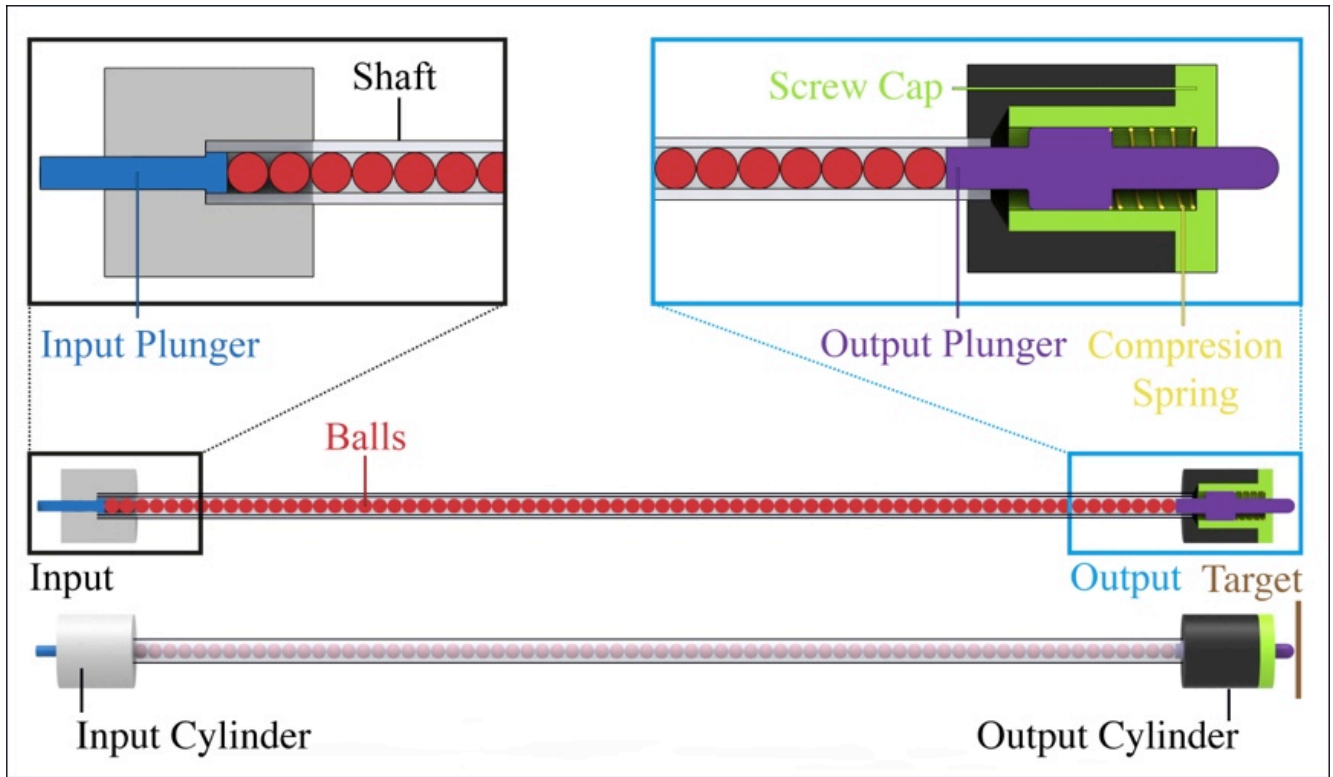
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91

92 **Fig. 2.** Schematic illustration of the *Cradle* prototype for testing purposes. Color indications: black = output housing, blue = input

93 plunger, green = output cap, grey = input housing, purple = output plunger, red = balls

94

95 *2.4. Experimental Facility*

96 For determining the feasibility of the *Cradle* prototype, an experimental facility was built (Fig. 3). The input impulse was

97 delivered by a solenoid actuator connected to a construction rail and was set to approximately 0.4 Ns using a voltage of 5.5

98 V (Power unit *ES 030-5*, Delta Elektronika, Zierikzee, the Netherlands) and an actuation time to 100 ms. The delivered

99 input and output impulses were measured using two miniature S-beam load cells (*LSB200 FSH00102* and *LSB200*

100 *FSH00104*; FUTEK Advanced Sensor Technology, Inc., Irvine, CA, USA) located at the in- and output, respectively. For

101 data acquisition, an analogue signal conditioner (*CPJ RAIL*, SCAIME, Annemasse, France), and a data acquisition system

102 with a sampling rate set to 10 kHz (*NI USB-6211*, National Instruments Corporation, Austin, TX) were connected to the

103 load cells. The load cells were controlled through *LabVIEW 2016* (National Instruments Corporation, Austin, TX).

104

105
106
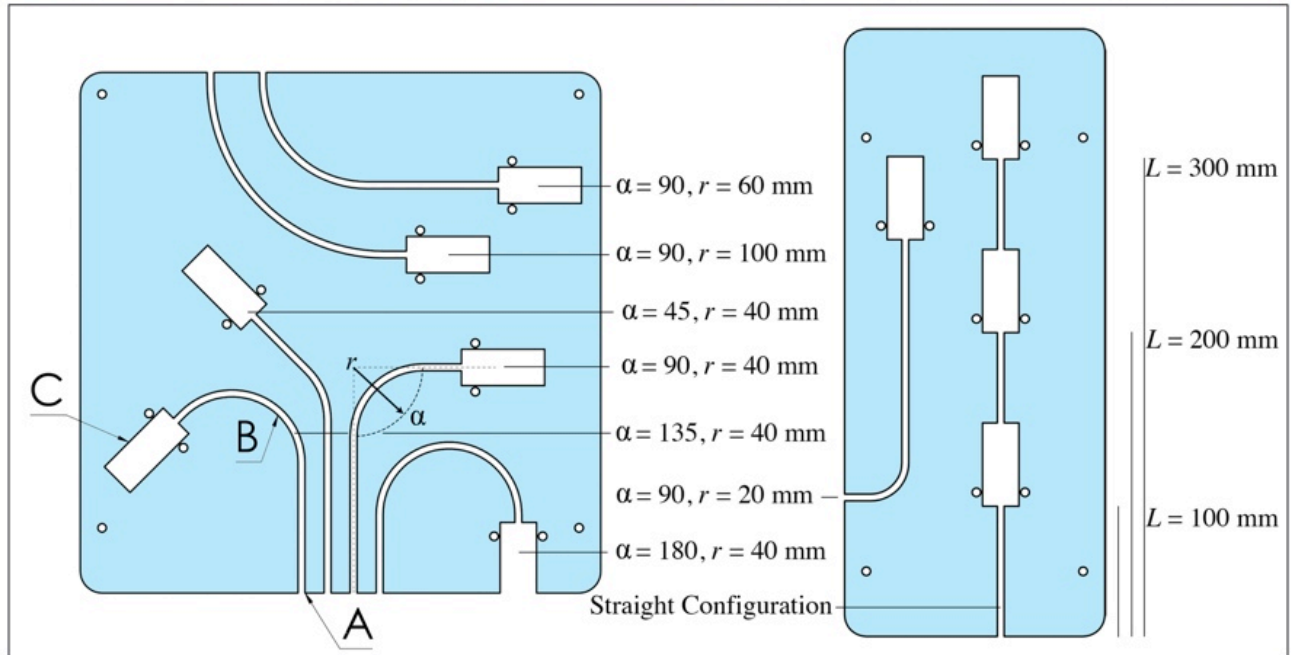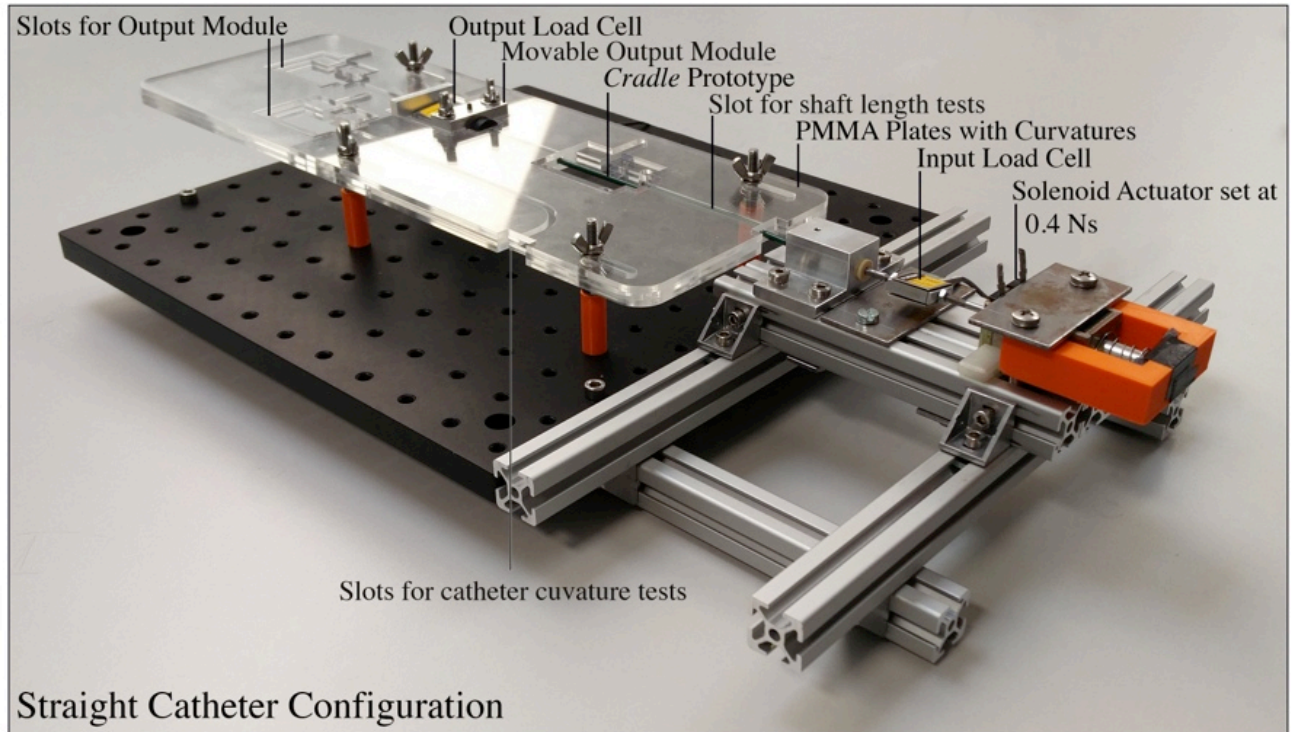107 **Fig. 3.** Experimental facility. Top: Complete Experimental Facility. Bottom: Schematic representation of the curvatures to which the

108 *Cradle* prototype was subjected. Letter indications: α = curve angle [°], A= slot for catheter, B= curvature to which the catheter is

109 subjected, C= insert for the output module containing the output load cell, L = length [mm] of the catheter, and r = curve radius [mm].

110

111   In order to test the effect of the curve angle and radius on the impulse and peak force efficiency and to confine the

112   prototype's shaft, three stacked transparent laser cut 3 mm thick PMMA plates were used: (1) a solid bottom plate, (2) a

113   middle plate containing the curvatures (see Fig. 3), and (3) a solid top plate, in between which the prototype was placed.

114   These plates were connected to a breadboard (*MB3030/M*, Thorlabs, Newton, NJ, USA) using four aluminums rods and M6

115   socket head screws. Since the solenoid and the input module were placed on two parallel construction rails, they could be

116   easily translated left or right to accommodate the different laser cut curves in the PMMA plates. It must be noted that the

117   position of the curves is different per configuration. The effect of the position of the curve was not researched in this study.

118   Additionally, in order to get a clearer understanding of buckling of the catheter during high-force impulse transfer, an

119   unconstrained test was performed.

120   *2.5. Experimental Protocol*

121   From the dependent variables, two efficiencies were calculated: (1) the impulse efficiency and (2) peak force efficiency.

122   First, the effect of the clearance between the balls and the shaft on the impulse and peak force efficiency of the *Cradle*

123   prototype was determined using the rigid stainless steel shaft of 200 mm in length. From these tests, the clearance with the

124   smallest impulse losses was determined. Subsequently, the effect of shaft type was determined using the stainless steel shaft

125   and catheter of 200 mm in length in the straight configuration and with a clearance of 0.3 mm. The clearance of 0.3 mm was

126   selected based on off-the-shelf catheter availability. Thirdly, the effect of shaft length on the efficiency was determined

127   using the flexible catheter shaft in the straight configuration. Fourthly, the effect of the curve angle on the efficiency was

128   determined for the 200 mm long flexible catheter shaft with a 40 mm curve radius. Fifthly, the effect of the curve radius on

129   the efficiency, was tested for the 200 mm long flexible catheter with a 90° curve angle. Each condition was tested 50 times.

130   Finally, the buckling resistance of the prototype was evaluated for the straight unconfined catheter shaft of 300 mm in

131   length and a clearance of 0.3 mm. This condition was tested 3 times.

132   *2.6. Data analysis*

133   Per measurement, the data from the S-beam load cell was processed with *MATLAB 2015b* (The Mathworks, Inc., Natick,

134   MA) to calculate the impact peak force and the impulses J_i and J_o [Ns] by integrating the force F [N] over the time dt [s] for

135   which it acted. Subsequently, the mean output peak force and mean output impulse with the associating standard deviations

136   were calculated per condition. The efficiency of the system [%] was determined in two ways: (1) dividing the output

137   impulse by the input impulse and multiplying this value by 100 and (2) by dividing the output peak force by the input peak

138   force and multiplying this value by 100, for the different clearances, lengths, curve angles, and curve radii. One-way

139   ANOVAs were conducted to determine whether the clearance, length, curve angle, and curve radius had a significant

140   influence on the efficiency. A T-test was performed to determine the effect of the shaft type.

3. Results

3.1. General Findings

143   Figure 4 illustrates an overview of the obtained data from the load cells. The highest peak force measured during the

144   experiment was 6 N in the straight configuration. As can be seen from Figure 4, the load cells have a block shape with a

145   slight peak at the start. It was also observed that the input peak force was not constant (see Table 1) for all the tests even

146   though the solenoid was calibrated and powered using the 5.5 V setting. Furthermore, the input impulse duration was also

147   slightly different per configuration; a longer impulse duration of approximately 70 ms was found for the straight catheter

148   configuration, while an impulse duration of approximately 30 ms was found for the rigid stainless steel tube. Furthermore, it

149   was noticed that the catheter slightly shifted during actuation. In Figure 5, an overview of the results is graphically

150   illustrated. In the unconstrained test no buckling was observed, see Figure 6.

3.2. Effect of Clearance on the Efficiency

152   Table 1 illustrates an overview of the effect of the clearance and input momentum on the efficiency. As can be seen from

153   Table 1, the peak force efficiency ranged from 32 to 57% and the impulse efficiency from 30 to 57%. The peak force
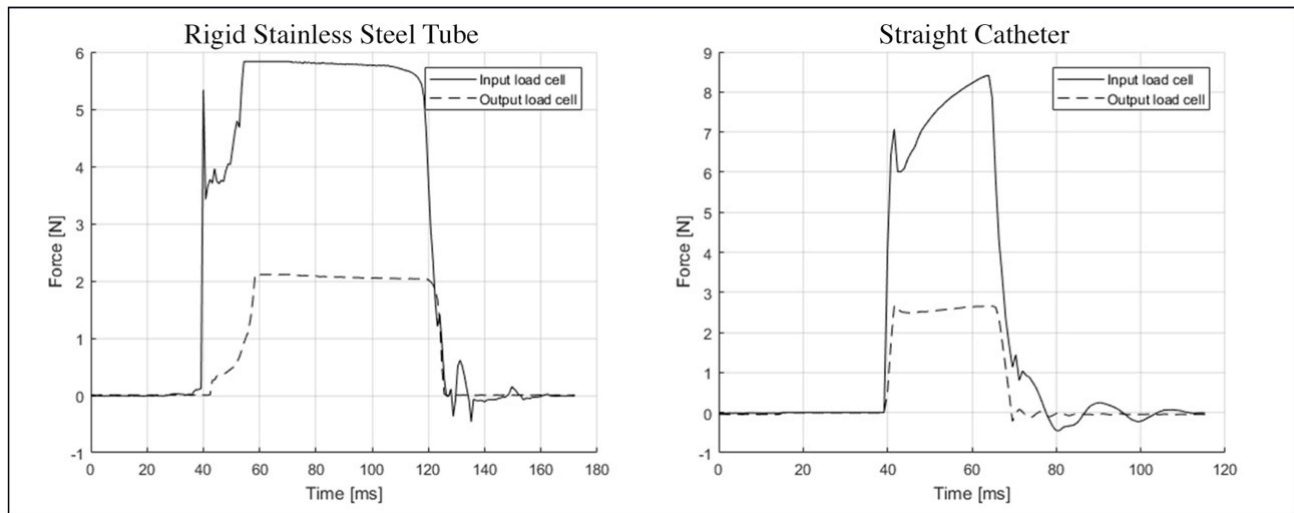
154

Fig. 4. Two examples of force versus time graphs of the input and output load cell for the straight configuration rigid stainless steel tube

157   (left) and catheter (right).

158 **Table 1. The Effect of the Clearance and Input Impulse of the Shaft on the Efficiency.** The effect of the clearance was tested using

159 the rigid stainless steel shaft. The length of the capillary tube was set to 200 mm.

| Clearance [mm]	Mean Input Peak Force [N]	Mean Input Impulse [Ns]	Mean Output Peak Force [N]	Mean Output Impulse [mNs]	Mean Peak Force Efficiency [%]	Mean Impulse Efficiency [%]
0.1	9.9 ± 0.5	$2.4 \cdot 10^{-1} \pm 14.0 \cdot 10^{-3}$	3.7 ± 0.3	$8.8 \cdot 10^{-2} \pm 8.2 \cdot 10^{-3}$	36.9 ± 1.6	37.3 ± 2.0
0.2	4.0 ± 0.3	$1.5 \cdot 10^{-1} \pm 9.6 \cdot 10^{-3}$	2.3 ± 0.1	$8.6 \cdot 10^{-2} \pm 5.6 \cdot 10^{-3}$	57.1 ± 3.9	56.9 ± 3.6
0.3	7.5 ± 0.1	$1.7 \cdot 10^{-1} \pm 5.8 \cdot 10^{-3}$	2.4 ± 0.2	$5.4 \cdot 10^{-2} \pm 3.9 \cdot 10^{-3}$	32.0 ± 2.4	30.1 ± 2.1

160

161 and impulse efficiency were negatively, and statistically significantly, affected by the clearance as determined by two one-

162 way ANOVAs (F (2, 237) = 1796, p = 6.6·10^{-144} & F (2, 237) = 2067, p = 10·10^{-161}, respectively). A clearance of 0.2 mm

163 resulted in the highest efficiency.

164 *3.3. Effect of Shaft Type and Length on the Efficiency*

165 The shaft type significantly influenced the peak force and impulse efficiency of the system as determined by two t-tests (t

166 (198) = 17.1, p = 5.5·10^{-41} & t (198) = 18.2, p =2.8·10^{-44}, respectively). A slightly higher efficiency was found for the

167 catheter in the straight configuration. The impulse efficiency was $30.1 \pm 2.1\%$ for the rigid stainless steel tube and $34.2 \pm$

168 0.8% for the catheter (see Table 2). The length of the catheter negatively, and statistically significantly, affected the peak

169 force and impulse efficiency as determined by the one way ANOVAs (F (2, 237) = 22020, p = 6.9·10^{-270} & F (2, 237) =

170 33933, p = 4.8·10^{-294}).

171 *3.4. Effect of Curvature on the Efficiency*

172 The curve angle negatively affected the peak force and impulse efficiency of the system (see Table 3) as determined by two

173 one-way ANOVAs (F(4, 395) = 106.6, p = 1.2·10^{-32} & F(4, 395) = 214.8, p = 7.6·10^{-72}). The impulse efficiency was

174 approximately 7.1% lower in the highly curved situation (α = 135° and r = 40 mm). The curve radius also statistically

175 significantly affected the efficiencies, with the lowest impulse and peak force efficiency found for the lowest radius of 20

176 mm (F (3, 316) = 299.2, p = 5.8·10^{-92} & F (3, 316) = 380.8, p = 1.5·10^{-104}).

177
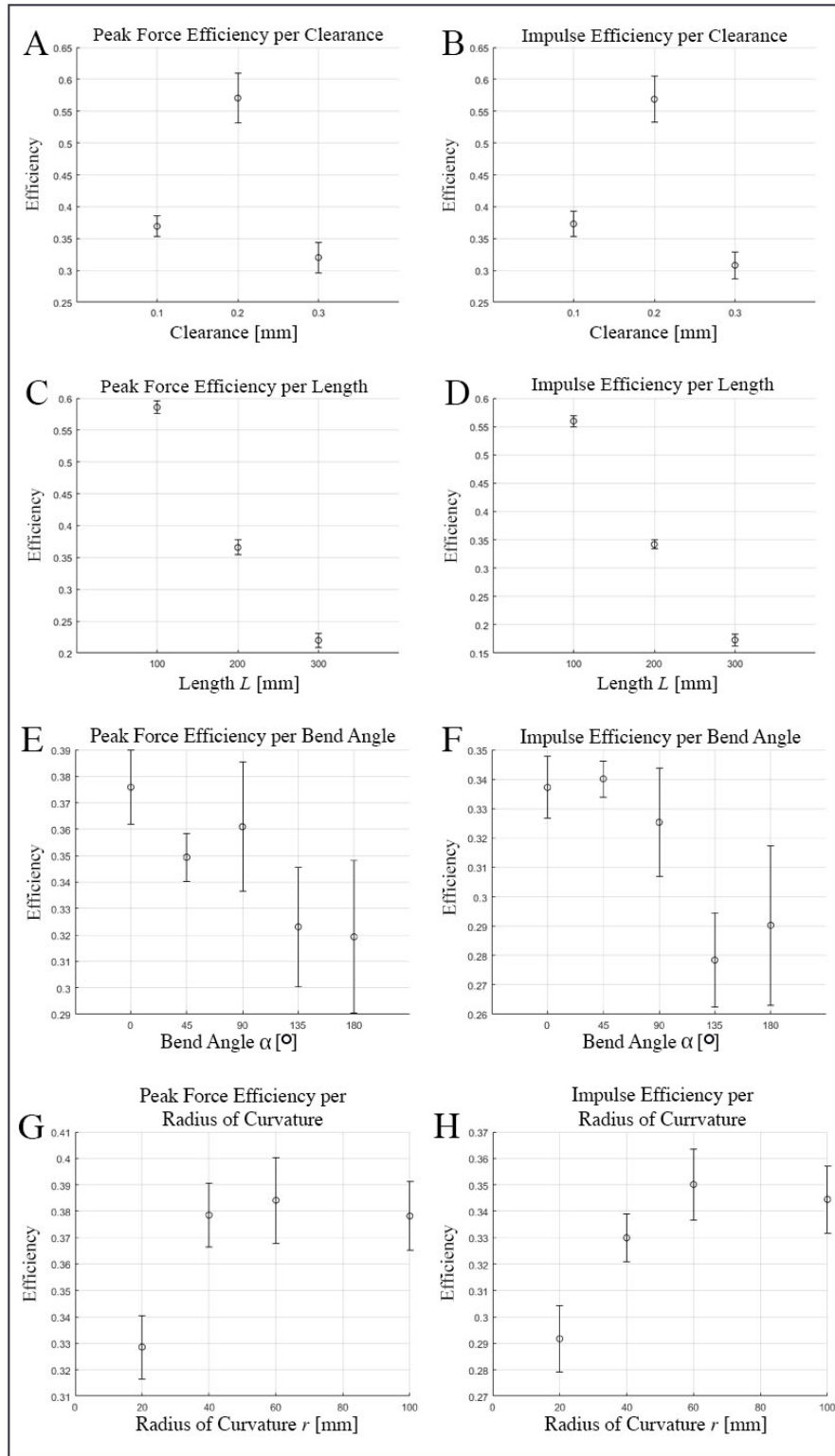
Fig. 5. Impulse and peak force efficiency results. (A+B) Peak force and Impulse efficiency for the different clearances of the rigid stainless steel shaft. (C+D) Peak force and Impulse efficiency for the different lengths for the catheter. (E+F) Peak force and Impulse efficiency of the catheter in different curve angles ($r = 25$ mm). (G+H) Peak force and Impulse efficiency of the catheter in different curve radii ($\alpha = 90°$).

Unconfined *Cradle* Prototype
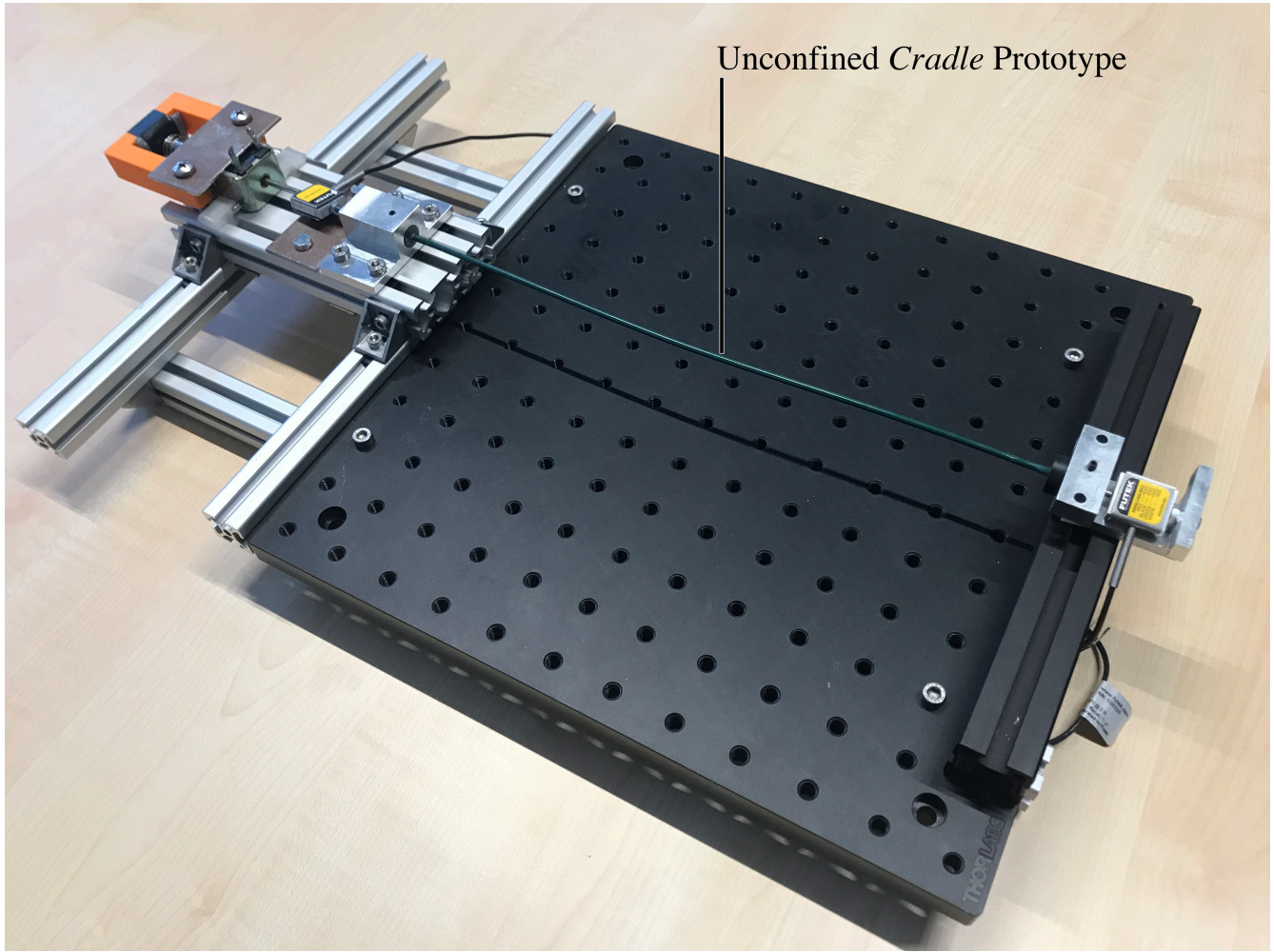
- **Fig. 6. Buckling Resistance Proof-Of-Principle Experiment with an unconfined *Cradle* Prototype.** For this experiment, a catheter length of 300 mm and clearance of 0.3 mm was used. No buckling was observed in this experiment.

Table 2. The Effect of the Shaft Type (Rigid Stainless Steel Tube versus Flexible Catheter) and Length on the Efficiency. The effect of the shaft types was tested in the straight configuration and a clearance of 0.3 mm between the balls and the shaft.

Shaft Type:	Length [mm]	Mean Input Peak Force [N]	Mean Input Impulse [Ns]	Mean Output Peak Force [N]	Mean Output Impulse [mNs]	Mean Peak Force Efficiency [%]	Mean Impulse Efficiency [%]
Rigid	200	7.5 ± 0.1	$1.7 \cdot 10^{-1} \pm 5.8 \cdot 10^{-3}$	2.4 ± 0.2	$5.4 \cdot 10^{-2} \pm 3.9 \cdot 10^{-3}$	32.0 ± 2.4	30.1 ± 2.1
Catheter	100	5.6 ± 0.1	$5.5 \cdot 10^{-1} \pm 16.3 \cdot 10^{-3}$	3.3 ± 0.08	$3.1 \cdot 10^{-1} \pm 9.9 \cdot 10^{-3}$	58.6 ± 1.0	60.0 ± 1.0
	200	4.9 ± 0.1	$4.9 \cdot 10^{-1} \pm 6.7 \cdot 10^{-3}$	1.8 ± 0.04	$1.7 \cdot 10^{-1} \pm 3.6 \cdot 10^{-3}$	36.6 ± 1.2	34.2 ± 0.8
	300	6.9 ± 0.2	$6.0 \cdot 10^{-1} \pm 16.1 \cdot 10^{-3}$	1.5 ± 0.06	$1.0 \cdot 10^{-1} \pm 7.1 \cdot 10^{-3}$	22.0 ± 1.1	17.3 ± 1.0

Table 3. The Effect of the Curvature of the Shaft on the Efficiency. The effect of the curvature was tested using the flexible catheter shaft with a clearance of 0.3 mm between the balls and shaft.

Shaft Type:	Curve Angle [°]	Curve Radius [mm]	Mean Input Peak Force [N]	Mean Input Impulse [Ns]	Mean Output Peak Force [N]	Mean Output Impulse [Ns]	Mean Peak Force Efficiency [%]	Mean Impulse Efficiency [%]
Flexible Catheter	0°	∞	4.9 ± 0.1	$4.9 \cdot 10^{-1} \pm 6.7 \cdot 10^{-3}$	1.8 ± 0.04	$1.7 \cdot 10^{-2} \pm 3.6 \cdot 10^{-3}$	36.6 ± 1.2	34.2 ± 0.8
	45°	40	6.2 ± 0.1	$5.9 \cdot 10^{-1} \pm 9.6 \cdot 10^{-3}$	2.2 ± 0.04	$2.0 \cdot 10^{-1} \pm 3.7 \cdot 10^{-3}$	35.0 ± 0.9	34.0 ± 0.6
		20	3.1 ± 0.08	$3.3 \cdot 10^{-1} \pm 6.9 \cdot 10^{-3}$	1.0 ± 0.05	$9.5 \cdot 10^{-2} \pm 5.0 \cdot 10^{-3}$	32.9 ± 1.2	29.1 ± 1.3
	90°	40	4.6 ± 0.07	$4.6 \cdot 10^{-1} \pm 6.2 \cdot 10^{-3}$	1.7 ± 0.05	$1.5 \cdot 10^{-1} \pm 3.9 \cdot 10^{-3}$	36.1 ± 2.5	32.5 ± 1.8
		60	3.6 ± 0.2	$3.7 \cdot 10^{-1} \pm 14.5 \cdot 10^{-3}$	1.4 ± 0.04	$1.3 \cdot 10^{-1} \pm 3.8 \cdot 10^{-3}$	38.4 ± 1.6	35.0 ± 1.3
		100	4.4 ± 0.1	$4.5 \cdot 10^{-1} \pm 10.9 \cdot 10^{-3}$	1.7 ±0.06	$1.6 \cdot 10^{-1} \pm 5.5 \cdot 10^{-3}$	37.8 ± 1.3	34.5 ± 1.3
	135°	40	5.6 ± 0.1	$5.4 \cdot 10^{-1} \pm 8.3 \cdot 10^{-3}$	1.8 ± 0.1	$1.5 \cdot 10^{-1} \pm 7.4 \cdot 10^{-3}$	32.3 ± 2.3	27.9 ± 1.6
	180°	40	4.9 ± 0.1	$4.9 \cdot 10^{-1} \pm 10.9 \cdot 10^{-3}$	1.6 ± 0.1	$1.4 \cdot 10^{-1} \pm 13.4 \cdot 10^{-3}$	31.9 ± 2.9	29.0 ± 2.7

4. Discussion

4.1. Summary of Main Findings

In this study, we have determined the feasibility of using a series of balls to transfer an impulse through a flexible shaft. The catheter was able to deliver high forces up to 6 N without buckling. In the current set-up we could not increase the input impulse due to the capacity of the load cells, however higher values can be achieved in future by increasing the input impulse. The impulse efficiency ranged in between 17.3 and 60.0%, with an average of 35%. The length had the greatest influence on the efficiency, while the curve angle and radii had only a minimal influence of in between 5–7%, which is beneficial for clinical situations where tortuous paths need to be followed towards the operation area. A slightly higher efficiency was found for the catheter shaft in the straight configuration than for the stainless steel tube, which is most likely due to a decrease in the friction coefficient between the balls and shaft. A clearance of 0.2 mm between the balls and the shaft resulted in the highest efficiency. However, the optimal clearance is yet to be determined and will most likely lie somewhere in the 0.1 and 0.3 mm range.

During the experiments it was observed that the input peak force was not constant throughout the tests even though the solenoid actuator was powered at an equal setting. This difference in input peak force is most likely caused by the difference in contact stiffness during impact in the different configurations, which in turn might be caused by the clearance, flexibility of the catheter shaft, or the curve. Also, due to the difference in contact stiffness the duration of the impulse changed for the

210   different configurations, resulting in an increase of approximately 40 ms for the more flexible catheter configuration. It must

211   also be noted that the difference in magnitude and duration of the impulse may have affected the efficiency of the system.

212   *4.2. Limitations of this Study*

213   In this study we have not determined the feasibility of using this system in a clinical situation. An application that would

214   benefit from a flexible tool that can deliver high forces without buckling is the endovascular treatment of Chronic Total

215   Occlusions (CTOs; see also Sakes *et al.* [9] on this topic). Furthermore, as there is a clear trend towards even smaller,

216   needle-like, instruments, buckling will become a more important failure mode that needs to be taken into account in the near

217   future. In order to prevent buckling and as such increase the success rate of the endovascular treatment of CTOs, as well as

218   percutaneous needle procedures, a *Cradle*-inspired catheter could be beneficial.

219   *4.3. Recommendations*

220   The true working principle of *Newton's Cradle* is still under debate. Multiple theories exist that describe the working

221   principle of *Newton's Cradle*, none of which are completely correct. Therefore, more research is necessary in determining

222   the true nature of *Newton's Cradle*. Furthermore, in an effort to understand the working principle of our system better,

223   research should be conducted into accurately determining the energy losses, due to friction and collisions in the shaft.

224   With optimization of the current design and facility, a higher efficiency could be obtained in future. For this purpose,

225   research should be conducted to determine the most optimal clearance and material combination of the balls and shaft.

226   Furthermore, it is also of importance to determine the optimal pretension, and thus contact stiffness, between the balls to

227   minimize energy loss. If the pretension is insufficient, gaps may form between the balls, which will negatively affect the

228   impulse transfer. Finally, it is also of importance to test the *Cradle* prototype in an environment that more closely resembles

229   the clinical situation. The experimental facility in this study is rigid and does not account for the flexible environment the

230   *Cradle* prototype will encounter during minimal invasive surgery. The effect of the flexible environment on the impulse

231   transfer, as well as the effect of catheter movement during actuation, will need to be researched in a follow-up experiment.

232   In order to integrate the catheter in a laparoscopic or endovascular procedure in the near future it is a necessity to allow

233   for single-handed control and adjusting of the output peak force. The safety of the device could be improved by

234   incorporating the catheter shaft with the output plunger. This will minimize the chance of loss of the indenter or the balls.

235   Additionally, the ability to actively steer the tip during the procedure should be researched, as this can increase the

236   effectiveness of the procedure and decrease the chance of unwanted tissue damage. Finally, in order to navigate through

237   narrow tortuous pathways and decrease the invasiveness of the procedure, it is a necessity to further reduce the outer

238   diameter of this catheter. Due to the simplicity of the design, further diameter reduction should be possible towards a sub-

239   millimeter scale. However, it must be noted that both the impulse and peak force efficiency will most likely decrease when

240   miniaturizing the outer diameter, as the number of balls, and thus contact points will increase. The same also goes for

241   increasing the length, which increases the required number of balls. Therefore, other ball-shapes should be researched, such

242   as cylinder-types, to minimize the number of contact points. Additionally, by redesign of the balls, the *Cradle* prototype

243   could be made torsion stiff. For example, instead of using balls, wedged-shaped one degree of freedom elements can be

244   used.

5. Conclusions

246   This study has illustrated that it is feasible to transfer high-force impulses through a series of balls confined within a rigid

247   shaft or flexible catheter. This technique could be beneficial for medical instruments that suffer from buckling, such as

248   guidewires and needles, especially since there is a trend for further miniaturization. More research is needed to allow for

249   high efficiency impulse transfer in even smaller and longer devices. In the future we will continue this investigation and will

250   develop this tool into a handheld clinical prototype.

Competing Interests

255   None declared.

Ethical approval

261   Not required

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